Effects of Lifting Posture on the Lumbar Spine under an Unexpected Unloading Condition

by

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ABSTRACT

Abstract of thesis entitled "Effects of Lifting Posture on the Lumbar Spine under an Unexpected Unloading Condition"

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Nurses and physical therapists working with patients (such as athetoid) are frequently exposed to sudden loads. Many researchers have investigated the effects of sudden load application. However, little is known about the effects of sudden release of load. It has been proposed that a sudden release of load, for instance, when a load slips, can generate an unexpected acceleration and very large muscle forces, which may reach a level sufficient to damage the soft tissues of the spine. In addition, when the externally applied load is suddenly released, the exerted muscle force created to maintain equilibrium would generate an unexpected acceleration and unbalance the body. The purpose of this study is therefore to investigate the response of trunk and leg muscles under sudden release during lifting and to determine whether a proper lifting posture could reduce the risk of injury to the back.

Ten normal and healthy male volunteers aged from 22-35 were recruited in this study. None of the subjects had any history of back injuries or significant back pain for the last two years. It was predicted that a subject would lose balance under a sudden unload condition if squat lifting with the lifting weight over 114N, asymmetric stoop lifting with the lifting weight over 173N, or symmetric stoop lifting posture with the lifting weight over 205N. Squat lifting was found to have longer response duration and longer co-contraction time compared to symmetric and asymmetric stoop lifting, meaning less stability under sudden unloading and also imposing larger forces on the spine. It was found that asymmetric stoop lift

had the highest peak axial rotation moment and peak lateral bending moment in comparison with symmetric squat and stoop lift; however, symmetric stoop had highest peak flexion-extension moment in comparison with symmetric squat and asymmetric stoop lift. Results also showed that squat lift had the highest absolute rate of change of moment, implying that it may be more difficult to preserve balance when encountering sudden release of load in squat lifting than in the other two lifting postures.

The safety limit for squat lifting is lower than others, but it has been commonly adopted as the preferred posture for lifting, and therefore the current ergonomic guidelines for proper lifting should be carefully addressed if sudden release conditions are to be taken into account. Nevertheless, the results of this study are valuable for establishing guidelines for manual material handling workers, who are exposed to sudden unload such as garbage collectors, truck unloaders and luggage dispatchers. Further biomechanical studies are required to determine the force generated by each muscle, and therefore the load on the lumbar spine under different lifting conditions can be studied quantitatively.

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LIST OF ABBREVIATIONS

LD	Latissimus Dorsi
ES	Lumbar Erector Spinae
EO	External Oblique
IO	Internal Oblique
RA	Rectus Abdominis
BF	Biceps Femoris (long head)
TA	Tibialis Anterior
RF	Rectus Femoris
G	Gastrocnemius (lateral head)
L	Left
R	Right
Sq	Symmetric squat lift
St	Symmetric stoop lift
A	Asymmetric stoop lift
AR	Axial Rotation
LB	Lateral Bending
FE	Flexion Extension

CHAPTER 1 INTRODUCTION

1.1 Background

Although back pain cannot be classified as a disease, it is one of the most common physical ailments in industrialized nations and is the largest epidemic society confronts. Each year, disability attributed to such pain increases significantly. Statistics show that low back pain affects all occupations with about the same rate of occurrence. Resulting disability is not associated with occupation, race or sex and is equally common among all kinds of people (Raschke, U., 1996).

More working days are lost because of back pain than because of any other disease or injury (Spengler et al., 1986). Low back pain is the leading cause of industrial disability in those younger than 45 years. The total cost of low back pain to the United States economy is $90 billion a year. In the United States, one million back injuries occur per year, 100 million work days are lost each year, and low back pain accounts for 20% of all work related injuries. (Bureau of Labour Statistics, USA, 1992).

There is a strong correlation between an increase in the occurrence of low back injury and increased manual load handling activity. Epidemiological studies have frequently reported the association between lifting and low back pain (Andersson 1997). During daily activities, each body segment such as the head, neck, trunk, hands, arms, pelvis and legs are subjected to forces of different magnitudes. The extensive rate of back injury linked with manual lifting has been focused on in terms of the stresses on the low back. As people lift and perform material handling tasks significant loads are placed on the spine. Although there is disagreement in the literature with regard to the etiology of low back disorders, it seems evident that mechanical factors are a principal cause (Frymoyer and Pope, 1978; Nachemson, 1978).

Many physical causes of low back pain such as bending and twisting were sudden maximal efforts incidentally carried out at the moment of incident (Magora, 1973). It has also been suggested that sudden maximal efforts could strain the soft tissues, especially when the worker is assuming an unfavourable posture for this type of effort. Many epidemiological studies also suggest that sudden loading can have an adverse effect on the spine (Marras et al., 1987; Lavender et al., 1989 & 1993; Oddsson et al., 1999). Although the effect of unexpected sudden unloading on the back can also be a source of low back pain, little is known about it and only a few related studies have been conducted to date.

Both sudden loading and unloading can result in slips and falls and it was found that the most frequently injured area in these accidents is the lumbosacral region. It was suggested that injury was due to the sudden unexpected nature of fall and slip, which cause trunk muscles to over-response and overload the trunk (Shannon, 1981). Sudden release of load can happen in many manual handlings such as luggage dispatching, refuse collecting or truck unloading. However, the possible adverse effects are not well documented. In order to establish an understanding of the effects and significance of an unexpected release of load, a series of experimental lifting studies was carried out to investigate the possible adverse effects of sudden release and gain a better understanding of the effects of different parameters such as lifting technique, lifting symmetry and weight lifted.

The purposes of this study were to determine the effects of different lifting conditions during the lumbar spine on sudden release of load situations and to investigate whether the risk of injury to the back due to such unexpected unloading conditions could be reduced or even minimized by the adoption of a proper lifting posture.

1.2 Hypothesis

It is hypothesized that adopting a preferred lifting posture could alleviate the potentially adverse effects of an unexpected decrease in load during lifting.

1.3 Objectives

The overall purpose of this study is to investigate the mechanics of lifting and the effects of different parameters (lifting symmetry, lifting technique and lifting weight) on the lumbar spine under sudden release of load. The specific objectives are:

1. To study the ability to preserve balance following sudden unexpected release of a range of loads while adopting different lifting symmetries and lifting techniques by measuring the centre of pressure excursion in both the antero-posterior and medio-lateral directions;
2. To qualitatively examine the activities of various trunk and lower limb muscles, including bilateral latissimus dorsi, lumbar erector spinae, external oblique, internal oblique, rectus abdominis, biceps femoris (long head), tibialis anterior, rectus femoris and gastrocnemius (lateral head), in response to sudden release of load;
3. To investigate low back loading during lifting by quantifying the net moment at the lumbosacral joint (L5/S1) and compare the effect of sudden unexpected decrease in the load being lifted on the net moment at L5/S1 while adopting different symmetric and asymmetric lifting techniques and lifting weights of various magnitudes;
4. To examine whether a specific lifting posture could reduce the possible risk of injury arising from unexpected unloading condition; and
5. To establish new guidelines for manual materials handling to take account of sudden unexpected decrease of load.

CHAPTER 2 LITERATURE REVIEW

2.1 Prevalence And Etiology Of Low Back Disorders

Low back pain is the main cause of industrial disability in those younger than 45 years. A study conducted by McCoy in 1993 estimated that 60-80% of the population in the USA were affected by low back pain at some time in their lives (McCoy, 1997). The total cost of low back pain to the United States economy is $90 billion a year. In the United States, one million back injuries occur per year, 100 million work days are lost each year, and low back pain accounts for 20% of all work related injuries. (Bureau of Labour Statistics, USA, 1992). In Hong Kong, although full statistics are lacking, the cost of occupational-related injury in terms of lost productivity is also large.

Extensive research on the back found that low back disorders arise from unknown sources. Pope et al. (1998) found only 50% of the cases relating low back disorders had a definable cause in their study. Kelsey (1988) also concluded that there were different types of back disorders, which undoubtedly have different etiologies, and it was difficult to know exactly what is meant by low back disorders because of different diagnostic criteria. White and Panjabi (1990) commented that the exact cause of low back disorders was still uncertain despite enormous efforts of researchers in the past two decades.

The search for the etiology of low back pain has been largely restricted to the degeneration of the intervertebral disc; however, it may not contain a complete answer to the problem. In particular, not everyone who has degenerated discs is affected by low back pain. Until the recent past, it seems that a majority of the investigators in biomechanics have most often implicated physically heavy work, static work postures, bending, twisting, lifting, forceful movement, repetitive work and vibrations in the genesis of low back disorders. Although there is disagreement in the literature with regard to the etiology of low back disorders, it seems evident that mechanical factors are a principal cause.

In fact, many epidemiological studies have indeed shown that workers exerting sudden unexpected maximal efforts are particularly vulnerable to low-back disorders (Magora, 1973; Andersson, 1981). Due to the relatively short moment arms of the muscles about the centre of rotation, the internal muscle actions generated must be quite large to respond to the external load and achieve equilibrium and stabilize the spine (Magnusson et al. 1996). It has been proposed that a sudden release of load, for instance, when a load slips, can generate an unexpected acceleration and very large muscle forces, which may reach to a level sufficient to damage the soft tissues of the spine. It is believed that these large muscle forces are responsible for the majority of the compressive and shears loads on the spine, and are consequently responsible for resulting back injuries when these loads become extreme. In addition, when the externally applied load is suddenly released, the exerted muscle force created to maintain equilibrium would generate an unexpected acceleration putting the body out of balance. Immediately, the body will respond by generating a very large muscle force, possibly larger than needed to regain the balance, and these postural reactions required to regain balance could be hazardous to the low-back musculoskeletal system, as suggested by Oddsson (1990), Magnusson et al. (1996) and Lavender et al. (1993).

Also, many physical causes of low back pain such as bending and twisting are sudden maximal efforts carried out at the moment of accident (Magora, 1973). It has also been suggested that sudden unexpected maximal efforts could strain the soft tissues, especially when the worker is assuming an unfavourable posture for this type of effort.

The risk of low back injury during slips and falls has been well documented by Manning & Shannon (1981). In their study, they found that the most frequently injured area in falling and slipping accidents was the lumbosacral spine. They also suggested that injury was due to the sudden unexpected nature of fall and slip, which cause trunk muscles to over-respond and overload the trunk.

2.2 Sudden Load And Sudden Unload

A sudden load is an unexpected load, which can take many forms and can be a sudden load application or release. These loading types can be found not only in leisure pursuits, but also in the workplace. Slipping and tripping are examples of sudden load application through the lower limbs. Unexpected slipping of an object being held in the hands or lifted is an example of sudden unloading. Nurses handling patients, physical therapists working with patients, drivers unloading their trucks and workers like refuse collectors, luggage dispatchers and movers are likely to be exposed to sudden unloading. Falls, slips and trips that occur while lifting a load are often associated with sudden jerking or twisting actions and can result in low-back injuries. Back injuries caused by falls are followed by longer sickness-absence and a higher rate of recurrence than back pain associated with other accidental causes (Troup 1981; Magnusson et al. 1996).

An experimental study was conducted by Oddsson et al. in 1999 to evaluate the effect of an unexpected postural perturbation during a lifting task by investigating the electromyographic responses in the erector spinae, during simulated slipping in a voluntary lifting movement. An increased erector spinae activity was found to be superimposed on the background activation present during the lift with forward perturbation, indicating that both the voluntary and postural motor programs caused activation of the erector spinae. However, during backward perturbation, there was a sudden termination of erector spinae activity followed by an extended period of rapid electromyographic amplitude fluctuations while the trunk was flexing, indicating an eccentric contraction of the erector spinae. This erratic behaviour with large electromyographic amplitude fluctuations in the erector spinae after a backward slip during lifting may indicate a rapid switch between voluntary and postural motor programs that require conflicting functions of the back muscles. This may cause rapid force changes in load-carrying tissue, especially in those surrounding the spine, thus increasing the risk of slip-and fall-related back injuries.

Lavender at al. (1989) studied the effect of varying amounts of preview time and task symmetry on trunk muscle response to sudden loading. The mechanical loading on the spine was quantified by the EMG activities of bilateral erector spinae, rectus abdominus, latissimus dorsi, and a generalized oblique muscle. He found that there was a linear relationship between the preview time and the peak normalized EMG signal, the mean normalized EMG signal, the onset rate of EMG activity and the lead and lag times of muscle response to the loading. Results indicate that the full preview condition (400ms) placed the least strain on the musculoskeletal system. For peak EMG activity, it was 21% greater than that observed in static conditions. The peak activity was found to be inversely proportional to the preview time. This indicates that any increase in the preview of loading should decrease peak muscle forces and thereby reduce the loading on the spine. The change in peak EMG activity seen under the asymmetric condition indicates that the system is under even greater stress because of the increased shear components that would be generated. It is this imbalance in the trunk loading that leads to high shear forces and possible back injuries. These findings were supported by Lavender et al.'s study in 1993. They tested the role of task experience in the development of preparatory strategies included muscle pretensioning, postural changes and intra-abdominal pressure in minimizing the postural disturbance to the body and the mechanical loading on the spine under sudden unexpected loading conditions. They found that even if the overall postural disturbance was not consistently reduced; the trunk flexion was significantly reduced in most subjects and the estimated spinal compression due to muscle loading was significantly reduced in all subjects. They also found that each subject developed a different preparatory strategy and while erector spinae pretensioning was always involved in the preparation, coactivation of the other trunk muscles was quite inconsistent among subjects. In addition, the use of intra-abdominal pressure during the preparatory period was found to be negligible and thus not likely to be a part of the preparatory strategy.

In 2000, De Looze et al. compared the reactions of the back and abdominal muscles and low back torque with presence or absence of load knowledge. They found that for a 6.5 kg weight, the back muscle activation and the peak L5/S1 torque in grasping the box without the load knowledge was 16% and 10% higher respectively. For a 16.5 kg weight, the back muscle activation was 10% lower during grasping and 10% higher during lifting in the absence of load knowledge situation. The significant deviations in back muscle activation, both in grasping and lifting, indicated that there was less optimal muscular control without the knowledge of the load and this may locally and instantaneously increase the mechanical loads. They also found that the knowledge of load had no effect on the activation of the abdominal muscles.

Marras et al. (1987) tested if trunk muscles overcompensation after sudden unexpected loading would result in excessive forces upon the trunk by asking subjects to hold a box in a static lifting position while weights ranging from 2.27 to 9.07kg were dropped into the box from a constant height under expected and unexpected conditions. They found that the mean and peak muscle forces for the unexpected condition were respectively about 2.5 times and 70% larger than the expected condition. Also, longer periods of force exertion and more rapid rises in trunk force development were found under the unexpected condition. Therefore, they concluded that trunk muscles would overcompensate under sudden unexpected condition and thus increase the load on the spine. Similar experiments were conducted by Magnusson et al. (1996) to study the effects of sudden load on the erector spinae. A weight of 20N was dropped from a constant height, applying a sudden forward bending moment to the trunk through a harness around the shoulders. They found that the spinal compression force was 4 times larger than in the expected condition and more exaggerated in asymmetric postures. Their findings were in contrast to those of Marras et al. (1987) and they attributed the discrepancy to the different experimental apparatus since the load was directly applied to the trunk in their study while the load was applied through the hands in Marras's study. Therefore, the force of upper limb overcompensation was transferred to the trunk. Also, the proprioceptive system of the back is comparatively slower than the arm-shoulder and so the trunk acts as an inertial damper and minimizes trunk motion whereas the upper limbs, which are of smaller mass, move more and activate its relatively faster proprioceptive system. Furthermore, 20N was not heavy enough to activate type II muscle fibres. Magnusson et al. (1996) also stated that, subjects minimize the postural disturbance under expected conditions by increasing the internal muscle reaction force but could no do so under unexpected conditions, putting the spine at risk since greater shear stresses were very likely be applied to the soft tissues such as the capsule and ligaments, resulting in low back injury.

2.3 Lifting Posture

Posture is defined as the configuration the body assumes to initiate an activity. Different material handling activities require different body postures. The body, however, may assume different configurations for the same activity. For example, loads can be lifted in squat, stoop or free-style postures.

2.3.1 Symmetric and asymmetric lifting

Conditions of asymmetry have been studied both in vitro (by simulation) and in vivo for lifting tasks; verifying that these conditions may be harmful for the musculoskeletal system of the trunk. Asymmetrical postures combining forward bending and twisting of the trunk increase intradiscal pressure (Andersson, 1985), shear and compression forces on the spine (Mital & Kromodihardjo, 1986), net bending moments about the trunk (Gagnon et al. 1993), and the electromyographic activity of paraspinal muscles (Kumar, 1980; McGill, 1991). In vitro studies, have predicted that twisting and bending are likely to cause disc prolapse (Adams & Hutton, 1982, 1985; Hickey & Hukins, 1979) and possibly damage the capsular ligaments (Adams & Hutton, 1985). Simulation studies confirmed that lateral bending and twisting coupled with flexion significantly increase the tensile strain in the disc fibres and increase the risk of disc rupture (Shirazi-Adl, 1989). Dynamometric studies showed that the capability of the trunk extensors decreases with asymmetrical trunk angle (Marras & Mirka, 1989). These results demonstrate the importance of asymmetric posture in relation to handling tasks.

Gagnon (1995) investigated the load exerted on the spine by sudden falling object in symmetric, asymmetric and full absorption positions by studying the maximal moments, maximal rates of loading of these moments and the integral of these moments. He found that the asymmetric posture imposed extra exertion on the contralateral trunk muscles especially the extensors and lateral flexors. Contrary to the hypothesis, a significant increase in muscular exertion was also found in the full absorption posture, where flexion of the elbows and the lower limbs was allowed. Gagnon attributed the result to the increase in the horizontal distance between the load and the lumbosacral joint because of the trunk flexion in the absorption posture. Therefore, they concluded that proper training for load absorption mechanism is necessary to reduce the risk of injuries (Lavender, 1989; Gagnon et al. 1995).

2.3.2 Squat lift and stoop lift

The squat posture is biomechanically less stressful than the stoop posture (Garg and Herrin, 1979; Leskinen et al. 1983), but the stoop posture leads to lower metabolic energy expenditure (Garg and Herrin, 1979; Kumar, 1984; Zuxiang and Xhijun 1990; Welbergen et al. 1991). Among squat, stoop and free style postures, the free-style posture is considered to be the least stressful (Brown, 1976; Garg and Saxena, 1979; Kumar and Magee, 1982; Kumar, 1984) or least tiring. However, there are mixed results in terms of metabolic energy expenditure. According to Brown (1973) and Garg and Saxena (1979), the free-style method of lifting is least expensive, but Kumar (1980) found the stoop lifting posture to be less expensive.

Hagen at al. (1993) studied physiological and subjective responses to maximal repetitive lifting employing stoop and squat techniques and found there was large variation in neuromotor activity among the subjects. Of the four muscles investigated, erector spinae showed the highest electrical activity for both lifting techniques. All of the subjects obtained higher peak amplitude for the vastus lateralis muscle during squat lifting than during stoop lifting. Except for one, all subjects showed higher peak amplitude for the biceps femoris muscle during stoop lifting than during squat lifting. A higher activity in the biceps femoris muscle during straight knee lift compared to lifting with flexed knees has also been reported (Nemeth et al., 1984).

Hagen et al. (1993) also studied the power output, O_2 consumption, heart rate and ventilation from ten experienced forestry workers under maximal repetitive lifting employing stoop and squat lifting postures. EMG activity of four muscles was recorded and the perceived central, low back and thigh exertion were also assessed. The two types of repetitive lifting were compared with maximal treadmill running. It was found that there was no significant difference between the two lifting postures regarding power output, however, the mean O_2 consumption was found to be significantly greater for squat than stoop. Also, the O_2 consumption during maximal squat lifting was found to be highly correlated with that of maximal treadmill running and stoop lifting, but no significant correlation was found between O_2 consumption of maximal treadmill running and stoop lifting. The perceived low back exertion was found to be significantly lower during squat lifting than stoop lifting, however, the result was not supported by EMG measurements.

In general, the conventional view of lifting postures is proper when considering compression on the lumbar spine during a lifting task and the recommended posture for lifting has traditionally been the squat lift. The stoop lifting posture has been avoided since it places greater stress on the back muscles, which is the prime force in producing compression on the spine and therefore, it is believed to present greater risk of injury during lifting.

2.4 Muscle Physiology

The act of lifting is believed to be associated to the development of low back pain (Frank et al. 1996) since it imposes high mechanical loads on musculoskeletal structures in the lower back region. The majority of the muscle groups, which act as effective movers or stabilizers of the spine have at least a single attachment on the spinal column or skull and therefore, the trunk musculature plays a crucial role in lifting and the understanding of the functions of trunk musculature is essential.

Anterior to the spine, the abdominal musculature of the thoracic and lumbar regions plays a vital physiological role. This group included four primary components, each described briefly as follows (El-Bohy, 1988):

1. The obliques externus abdominis act to flex the thoracic and lumbar spine against eternal resistance such as gravity. In concurrence with the action of

other local musculature, it flexes the spine laterally, and when combined with other spinal rotators, rotates the spine to the opposite side of muscular action.

2. The obliques internus abdominus lies beneath the external oblique and functions similarly. However, it rotates the spine to the same side of contraction.
3. The rectus abdominus is the most superficial of the abdominal muscles and acts to flex or laterally flex the spine depending on contraction symmetry.

Posterior to the spine, there are three major muscle groups, which influence the lumbar, cervical and thoracic regions. The most important group is the erector spinae, which starts as a large mass of muscle tissue in the lumbar region but separates into three distinct branches consisting of the iliocostalis, longissimus, and spinalis. Reported results of EMG studies indicate that the erector spinae is not very active in the erect posture, unless a deliberate effort is made to extend the thoracic spine more completely (Luttgens and Wells, 1982). In forward flexion of the spine, the erector spinae undergoes eccentric contraction until the weight of the upper body is supported by the vertebral ligaments (Basmajian and Deluca, 1985). However, the most forceful engagement of the muscle occurs during extension, hypertension and lateral flexion when these movements are performed against gravitational influences or other external resistances. In conclusion, the function of the erector spinae, depending on contraction symmetry and coordination with other anterior and lateral muscles, is to extend and/or rotate the spine and cause lateral flexion.

Latissimus dorsi is one of the superficial muscles of the back, which connect the upper limbs to the trunk and are concerned with movements of these limbs. The latissimus dorsi extends, adducts, and medially rotates the humerus at the shoulder joint.

It is obvious that any component of the complex structure of the spine is a possible source of low back pain, providing this component has sensory nerve endings. In addition, knowledge of the innervations of the lumbar spine is essential for a better diagnosis and treatment of low back disorders. The principal aim of all recent research concerning the anatomy of the intervertebral joint is to track the nerve endings and branches that supply each component. Pain nerve endings are found in nearly every tissue of the body and it is transmitted through two types of nerve fibers to the central nervous system (CNS). Myelinated fibers are relatively large in diameter, and relay neural signals at a rate of 12-30m/s (Ganong, 1983). Unmyelinated fibers are much smaller, and respond slowly, with a conduction rate of 0.5-2m/s. In addition, it has been suggested that a chemical agent is liberated during the process of stimulation of nerve endings to initiate pain (El-Bohy, 1988).

2.5 Muscle Electromyography

Electromyography (EMG) of the spinal muscles has been studied since the early 50s. Surface EMG is commonly used by orthopaedic scientists, therapists and clinicians to obtain information about the amplitude and temporal characteristics of the activation of different muscles. Because it is an indirect measure of contractile muscle activity, EMG has been frequently incorporated in biomechanical models to provide a predictive capability.

Magusson et al (1996) studied the myoelectric (EMG) response of the erector spinae to expected and unexpected sudden unloading. A simple modelling method was employed to estimate the load on the lumbar spine. The model is considered to be inadequate as only EMG signals from the erector spinae were used, while the anterior trunk muscles would be the major muscles to respond to such a sudden change of load. A pilot study by the authors demonstrated that the anterior trunk muscles such as rectus abdominis, internal and external oblique muscles would contract and generate a relatively high EMG level while the posterior erector spinae would relax in response to the sudden unloading condition. Although the onset and termination of action at each muscle to such a sudden unloading condition can be documented, the actual amount of load on the lumbar spine has not been accurately estimated. The actual affects and hence significance of the sudden unloading condition is still unknown.

In fact, the EMG amplitude may have been influenced by the fact that the force and EMG response were calibrated during a maximal isometric contraction, while the lifting tasks involved some dynamic activity for the muscles investigated (Basmajian and de Luca, 1985). The relationship between surface EMG amplitude and muscle moment has also been shown to be influenced by the posture or angle of the joints upon which the muscle acts (Rosenburg and Seidel, 1989). For the lumbar spine this means that the EMG-amplitude-to-lumbar-moment ratio may be at variance between squat and stoop postures. Hansson et al. (1984) found that the compressive load on the lumbar spine during isometric testing in squat and stoop postures were close to a level at which structural failure of the spine could be expected.

2.5.1 Electrode type, placement and distance

Surface EMG recordings provide a safe, easy and non-invasive method that allows objective quantification of muscle activity. The quality of the signal depends on a number of factors including electrode type and placement. However, only a limited amount of information is available on this topic.

Zedka et al., (1997) investigated the influence of electrode type, interelectrode distance and electrode orientation on EMG signals from the paraspinal muscles. Bipolar electrodes were placed over the erector spinae at different distance in series (cranio-caudal) and parallel (medio-lateral) directions. Subjects were asked to preformed different levels of isometric contraction and the RMS EMG signals were analysed. The average amplitude and total power of the EMG signal was found to increase while the mean frequency was found to decrease with increasing interelectrode distance in the parallel direction. Although a trend towards higher average amplitude, total power and mean frequency was found for electrodes placed in series compared to those in a parallel direction, the difference was not significant. Therefore, they concluded that consistent information of muscle activity could be obtained independently from interelectrode distance or orientation for both the Miniature Biopotential Skin Electrodes and 14445C Hewlett-Packard electrodes. They also found that the electrodes could be prevented from sliding during muscle contraction by placing them in "in parallel" orientation.

In contrast to Zedka's findings, Fridlund and Cacioppo (1986) suggested that electrodes should be placed parallel (cranio-caudal) to the muscle fibres whenever possible to maximize sensitively and selectivity. Perpendicular placements tend to lead to greater common mode rejection and less selectivity. They also highlighted the following elements that can improve the fidelity of surface EMG recording (Cram and Kasman (1998):

1. Select the appropriate proximity of a proposed site to the underlying muscle mass, keeping the minimum amount of tissue between electrodes and the muscle fibres themselves.
2. Avoid straddling the motor end plate region. If this is done, the amplitudes observed are typically lower due to differential amplification. Placing electrodes a little off the centre of the muscle will accomplish this goal.
3. Choose sites that are easy to locate (sites that have good anatomical landmarks to facilitate reliable placement of electrodes during subsequent recording sessions).
4. Choose sites that do not unduly obstruct vision or movement. Avoid areas that present problems from skin folds or bony obstruction.

5. Minimize cross-talk from proximal deep or superficial muscle by selecting the best electrode size and interelectrode spacing.

2.5.2 Data processing

Filtering and sampling

According to the standards for reporting EMG data detailed in the Journal of Electromyography and Kinesiology (1996), surface EMG should not be filtered above 10 Hz as a low cut-off, and below 350 Hz as a high cut-off since the power density spectra of the EMG contains most of its power in the frequency range of 5-500 Hz at the extremes. It is important to note that low pass filtration is performed to remove the higher frequency components of the signal producing a smoother trace to assist the identification of the EMG onset (Soderberg and Cook, 1984). However excessive smoothing of the data results in a loss of information and inaccurate identification of EMG onset (Halbertsma and DeBoer, 1981; Soderberg and Cook, 1984; Gabel and Brandt, 1994). In contrast, insufficient data smoothing results in delayed identification of EMG onset due to the high frequency changes in amplitude, resulting in a reduced mean for the specified number of samples (Hodges and Bui, 1996).

It is advisable that the minimal acceptable sampling rate is at least twice the highest frequency cut-off of the bandpass filter, and preferably higher to improve accuracy and resolution. It is also advisable that, prior to bandpass filtering, a sampling rate of 2500 Hz or above should be used for recording the raw EMG (Journal of Electromyography and Kinesiology, 1996). Various cut-off frequencies and sampling rates were used in different studies for analysing different muscle activities.

In 1996, Hodges and Bui recorded the electromyographic activity of trunk and limb muscles using both surface and fine-wire electrodes during a series of rapid upper limb movements. The EMG signals were bandpass filtered at 10-1000 Hz (Basmajian and Deluca, 1985; Perry, 1992) and sampled at a rate of 2000Hz with 12-bit analogue to digital conversion. They claimed that the low pass cut-off frequency that most accurately and consistently determined the EMG onset was 50Hz. Also, several previous studies have incorporated filtration similar to this value (Nashner et al., 1983; DiFabio, 1987)

Stokes et al. (2000) recorded signals from six bilateral trunk muscles using both surface and indwelling wire electrodes. Subjects were asked to stand in an apparatus with the pelvis immobilized and a harness around the thorax providing a preload and a force perturbation by a horizontal cable. EMG signals were bandpass filtered by a 10- to 100 Hz Chebychev Type II filter with no lag and rectified to reduce any electrocardiograph (EKG) or motion artifact and high frequency noise contamination of the signals.

Onset and termination of muscle activity identification

Onset and termination of EMG activity are two of the most common neuromuscular function parameters used in the evaluation of posture and movement. However, there is no standard method to determine these parameters, and little agreement exists regarding methods for identification of the onset and termination of EMG activity.

In the past, most studies either determined the onset and termination visually by moving a cursor across a display of the EMG signal (Armstrong, 1984; Barret, Shibasaki and Neshige, 1985; Woollacott et al., 1988; Latash et al., 1995) or semi-manually to decide on the threshold level by trial and error with the aid of computer program used to compute onset and termination of the bursts (Lidierth, 1986; Neeman et al., 1990). In general, the criteria used for these visually determined decisions were not reported.

Hodges and Bui (1996) stated that several studies reported the criteria for identification of onset of EMG activity as the earliest detectable rise in EMG activity above the steady state (Allum and Pfaltz, 1985; Crenna et al., 1987; Woollacott et al., 1988; Inglis et al., 1994) or the point where the signal first deviates more than 1 or 1.5 standard deviations from the level recorded during the steady state (Nashner et al., 1983; Nashner and Forssberg, 1986) without mention of how this is determined. Also, although visual onset determination is subjective, several authors reported the inaccuracy to be as low as ± 5-10ms, but again without mention of how this was assessed (Hallett et al., 1975; Horak et al., 1984). This inaccuracy is up to half of the experimental difference reported previously (Traub et al., 1980; Cordo and Nashner, 1982). In several studies, traces were assessed by two examiners and excluded if the identified onsets differed by greater than 5ms, providing some assurance of reliability (Brown and Frank, 1987; Crenna et al., 1987). However, although intra-tester reliability coefficients of up to 0.78-0.82 have been identified when comparing the visually derived EMG onsets identified by 3 experienced examiners, the percentage of trials in which the same millisecond was chosen as the EMG onset was between 30 to 36% when comparing examiners and between 47 to 56% for each examiner between days (DiFabio, 1987). Clearly, much variation in EMG onset determination occurs when this is done visually and is dependent on the experience and skill of the examiner. In an attempt to increase the objectivity (DiFabio, 1987) of the evaluation of EMG onset and to reduce observer bias (Studenski et al., 1991) an increasing number of studies rely on computer analysis. The following table shows the summary of onset determination criteria:

Table 2.1 Summary of onset determination criteria (Hodges and Bui, 1996)

Author	Burst	Baseline	Signal processing	Method
Neafsey et al. (1978)	Two consecutive 50ms bins>SD from baseline activity	Initial 500ms of 2s before stimulus	Averaged in 50ms epochs	Computer -based
Greenisen et al. (1979)	RMS>RMS threshold for >2 windows of 50ms (threshold criteria not stated)	Not stated	1000Hz low pass filter	Computer -based
Nashner et al. (1983)	First deviation>1.5 SD from baseline	100ms prior to stimulus	40Hz low pass filter	Not stated
DiFabio (1987)	>3 SD beyond baseline for 25ms	50ms prior to stimulus	50Hz low pass filter	Computer -based
Lee et al. (1987)	>2 SD beyond baseline for 40ms (both mean and median of sample)	500ms quiet standing prior to stimulus	10-100 Hz band pass filter	Computer -based
Chanaud & Macpherson (1991)	>2.5 SD beyond baseline	100ms quiet stance	Low pass filter	Not stated
Studenski et al. (1991)	3 x baseline for 8 of 16ms	100ms quiet standing	Not stated	Not stated
Happee (1992)	Segment of n samples>previous segment at 5% significance level and following sample>previous at 5% significance level and current sample and following samples>30% maximal signal (n=20% of movement duration)	Not stated	500Hz low pass filter	Not stated
Bullock-Saxton et al. (1993)	15% maximal contraction for cycle of gait	Not stated	Not stated	Not stated
Bullock-Saxton (1994)	5% of peak magnitude of burst	500ms prior to burst	Low pass filter, subtraction of baseline mean	Computer-based
Steele (1994)	>1 SD beyond baseline for 80ms (not below for>16ms)	240ms prior to signal to move	6Hz low pass filter	Computer-based
Karst & Willet (1995)	>1 SD beyond baseline, visually verified on computer screen and modified if inaccurate	First 15ms of sampling window	Ensemble average of four trials	Computer-based
Thompson & Mckinley (1995)	>95% confidence interval for >10ms	Not stated	10-500Hz band pass filter	Not stated

Although each of the computer algorithms uses different criteria to determine the EMG onset, no studies have evaluated the relative accuracy of each or identified if specific trace characteristics require different onset determination methods.

Hodges and Bui (1996) compared a variety of criteria used for the computer determination of EMG onset against the EMG onsets determined visually by an experienced examiner to identify the parameters, which provided the most accurate identification of EMG onset. Twenty-seven methods were compared which varied in terms of EMG processing (low pass filtering at 10, 50 and 500Hz), threshold value (1,2 and 3 SD beyond mean of baseline activity) and the number of samples for which the mean must exceed the defined threshold (20, 50 and 100ms). Three hundred randomly selected trials of a postural task were evaluated using each technique. The visual determination of EMG onset was found to be highly repeatable between days. Linear regression equations were calculated for the values selected by each computer method, which indicated that the onset values selected by the majority of the parameter combinations deviated significantly from the visually derived onset values. There were three combinations of parameters that resulted in regression lines with a y-intercept that did not significantly deviate from the visual data; 50ms/1 SD/50Hz, 25ms/3 SD/50Hz and 10ms/1 SD/500Hz, and these were all recommended for future use.

Stokes et al. (2000) studied whether increased preactivation of muscles was associated with decreased likelihood of muscular activation in response to a transient force perturbation. The onset of the force perturbation was first identified from the load cell recording by detecting the time at which there was a significant increase in the force-time slope. A 25- to 150-millisecond time window after the force perturbation was examined using two different methods, namely the Shewhart method and mean the EMG difference method to detect any short and medium latency muscle responses to the perturbation. For the Shewhart method, the baseline mean was commuted using the median mean EMG signal magnitude in five sequential 100-millisecond windows over the 500 milliseconds preceding the force perturbation. The standard deviation of the 100- millisecond window corresponding to the median mean was used to compute the three standard deviation threshold. For the mean EMG difference method, the difference between the mean EMG signal in a 25- to 150-millisecond window after the perturbation and a 275- to 150- millisecond window before the perturbation was computed for each EMG signal. A response was defined as an increase in the mean EMG signal after perturbation. This detection method was not influenced by the difference in the standard deviation of the baseline EMG signal between experimental conditions that could potentially produce detection bias with the Shewhart method. However, latencies could not be calculated by this method. Low preload efforts were found to be associated with higher muscle response frequencies using the Shewhart method. The average number of trials producing a detected response was observed to be greater at low preload than at high preload for bilateral internal and external obliques, bilateral rectus abdominis, left longissimus and left iliocostalis when averaged for all test conditions. While using the mean EMG difference method, there were significantly more detected responses at low preload effort for the same dorsal muscles, but for only one of the abdominal muscles. In fact, the detection of reliable response was difficult because detected responses were comparatively small to the fluctuation of the EMG signal from preactivated muscles and therefore, the proportions of responses detected by these two methods are not directly comparable. Regardless of the concern that the Shewhart detection algorithm might be influenced by the level of preactivation of the muscle and would therefore be biased by the preload effort, the consistency of response frequencies and preload effects between the two detection methods showed that this was not a problem.

Abbink et al. (1998) presented a method of automated detection of onset and termination of chewing (rhythmic) muscle activity in EMG. A threshold level in the EMG is computed, such that amplitudes in the EMG signal exceeding this level indicate muscle activity. The threshold level is determined using a statistical criterion based on the amplitude distribution of the entire EMG signal. With each burst first a search interval is needed in which the burst is located approximately in the middle. The software uses a set of pointers, one to each cycle, as references to establish the search intervals. In chewing when the position of the mandible is known, the onset of jaw closing is used as a reference. For jaw-elevator muscles that are involved in closing the jaw, a search interval is used that starts 100ms before jaw closing and that ends at the next onset of jaw closing. The cycle pointers are computed by determining where the signal crosses that threshold in upward direction. Once the search interval of a burst has been established, the program determines the onset of the burst by looking for a transition from amplitudes that are generally smaller than the basic EMG threshold level, to an amplitude that generally exceeds the basic EMG threshold level. The search starts in the burst, travelling in the direction of the beginning of the interval, and the program computes at each point i the so called transition index Trans(i) = $n< (i) \div n> (i)$. In this expression $n< (i)$ is the number of amplitudes smaller than the basic threshold level in the window of n amplitudes directly preceding amplitude i, and $n> (i)$ is the number of amplitudes smaller than basic threshold level in the n amplitudes directly following amplitude i. In the middle of the burst in both intervals most amplitudes exceed the basic threshold level, Trans(i) is about n. Shifting towards the beginning of the burst, the left interval will begin to move out of the burst and Trans(i) decreases. The onset of the burst is at the maximum transition index. To determine the termination of the burst a similar search is carried out in the opposite direction, looking for the minimum transition index. It was also stated that although this method is illustrated with EMG signals recorded from chewing muscles, it can be used for any EMG signal containing cyclic bursts of activity and thus may be applied in studies on rhythmic movements such as walking and breathing.

Cross-talk

EMG recording techniques require treating the muscle as a volume conductor; the pickup site does not discriminate between signals originating from the underlying muscle, an adjacent synergist, an opposing antagonist, or ambient electrical noise. All such sources will be recorded in direct relation to the amplitude of the signal received at the electrode site and thus generate cross-talk (Kamen and Caldwell, 1996). Therefore, cross-talk is a biological artifact; signals generated from muscles other than the muscles of interest and which can contaminate the EMG through volume conduction and thus distort its interpretation, relative to both amplitude and temporal characteristics.

Cram and Kasman (1998) stated that careful placement of closely spaced electrodes is the only hope for limiting the cross-talk artifact. In considering placement of electrodes for a muscle that has very close proximity to neighbouring muscles, it is sometimes possible to place recording electrodes on a distal portion of the muscle to avoid or minimize cross-talk. For example, electrode placement on the forearm is not directly over the belly of the flexor digitorum muscle, but detects enough of the action potentials from the distal portion of its muscle fibers to reflect finger movement.

Vink et al (1987) quantified cross-talk using 12 pairs of bipolar surface electrodes over the erector spinae group during isometric contractions of 10 to 100% maximum voluntary contraction at interval of 10%. Using the cross-correlation coefficient function, they demonstrated that the absolute maximum in the correlation coefficient was less than 0.30, or about 10 % of common signal when electrode pairs were placed more than 30mm apart. Therefore, they concluded that, even at the small distance of 30mm between electrode pairs, myoelectric signals are specific and optimise selective recording of localized muscle activity in the erector spinae.

According to Kamen and Caldwell (1996), there are a number of different procedures available to measure and identify the existence of EMG cross-talk. The simplest one is to perform functional resistance tests that isolate specific muscle groups and examine the activity in nonactive muscles (Winter et al., 1995). Stimulation of the nerve innervating a synergist or antagonist muscle at various intensities will often reveal the extent of volume conduction from adjacent muscles (Turker and Miles, 1990). Experiments requiring moderate or high-intensity stimulation to record M waves can use double-differential EMG to aid in discriminating between distant volume-conducted signals and signals from the relevant source muscle (Turker, 1993). Pairs of EMG signals in which the existence of cross-talk is suspected can be cross-correlated to examine the interrelationship between the two muscles (Winter et al., 1995). Also, signals that are highly intercorrelated either simultaneously or with a time lag attributable to volume conduction delays are frequently assumed to represent an appreciable degree of cross-talk.

Koh and Grabiner (1992) estimated the amount of cross-talk in surface EMG of human hamstring muscles using a protocol in which the quadriceps femoris was electrically stimulated via the femoral nerve. EMG was recorded from the vastus lateralis and the medial and lateral hamstring muscle groups. The amplitude of the EMG response of the vastus lateralis to electrical stimulation was adjusted to match that of its maximum voluntary effort (MVE) under isometric conditions. Subsequent power density spectrum analysis showed that the median frequencies of the signals generated by electrical stimulation and MVE were not significantly different. In conventional bipolar recordings, cross-talk in lateral hamstring EMG averaged 17.1% MVE and in medial hamstring EMG 11.3% MVE. The double differential technique significantly reduced cross talk to 7.6% MVE for the lateral hamstrings, and to 4.2% MVE for the medial hamstrings. The double differential technique appears to be more selective than the bipolar technique when recording EMG from muscles with highly active neighbours and thus should be used in such situations. They also suggested placing each pair of electrodes at least 3cm apart to minimized the volume-conducted signal and use smaller electrodes placed at closer distances to minimize the overlap between adjacent pairs of electrodes.

However, cross-talk interference from other neighbouring muscle sources is inevitable for some muscles during surface EMG recordings. Benhamou et al. (1995) suggested that splenius capitis surface recordings are virtually impossible to obtain without interference from other muscles, particularly the sternocleidomastoid. The use of intramuscular electrodes is advocated in these instances in which EMG cross-talk from surface electrodes is otherwise unavoidable.

2.6 Balance Preservation

Postural regulation plays an important role in preserving balance in a standing posture by keeping the projection of the body's centre of gravity within the base of support. Different mechanisms will be initiated to maintain the equilibrium of the body whenever it is undermined as suggested by different investigators.

(Hay and Redon, 1999) investigated the differential effect of a self-initiated or unpredictable and externally imposed unloading in children's and adult's postural equilibrium. He suggested that postural activities are controlled via both feedforward and feedback processes. However, feedback postural control is the only process available for maintaining balance when subjects are dealing with unpredictable externally generated postural disturbances. Therefore, the involvement of feedback and/or feedforward control mode was tested by comparing the displacement of the centre of pressure occurring as the result of unloading, in reactive or predictive condition. He found that the unloading resulted in a backward movement of the centre of pressure, which was smaller under self-initiated disturbance than imposed disturbances for all age groups. This difference varied depending mostly on the age-related change in the relative amplitude of the self-initiated disturbance, which decreased between 3 to 5, and 6 to 8 years old and increased again in the two older groups, 9 to 10 years old and adults. It was concluded that feedforward control becomes more efficient as children age, but that it's relative contribution to postural control does not show a monotonic pattern of development.

Load knowledge has been proved to be essential in lifting small objects with a precision grip, since the vertical lifting force pattern was found to be scaled to the object's weight (Forssberg et al. 1992). In bimanual lifting tasks involving the whole body, a similar scaling of the vertical force pattern can be assumed, for which adequate load knowledge would be required. Moreover, correct knowledge about the load mass to be lifted seemed important to make adequate preparations to counteract the threat to balance that is imposed upon the lifter by picking up a load in front of the body (Commissaris and Toussaint 1997).

Commissaris and Toussaint (1997) and Toussaint et al. (1998) studied the effect of the presence or absence of load knowledge on the low back loading and the control of balance in lifting tasks. The control of balance was studied by the position of the centre of gravity relative to the base of support, the horizontal and vertical momentum of the centre of gravity and the angular momentum of the whole body. They found significant changes in the horizontal and angular momentum prior to picking up a known load mass. The preparations were technique specific and picking up a load using stoop lift seemed to allow better balance preservation than using a squat lift (Commissaris and Toussaint, 1997). It was also found that the central nervous system would assemble a counteractive strategy scaling the magnitude of the anticipatory postural adjustment with the assessed level of predictability of the postural disturbance, and selectively tunes the anticipatory postural adjustments to store information gained from previous lifts (Toussaint et al., 1998).

Rietydk et al. (1999) studied the motor mechanisms used in postural control by determining the joint moments during balance recovery from medio-lateral perturbations. The observed joint moments served to move the centre of pressure in the appropriate direction and to control the lateral collapse of the trunk. The individual joints involved in controlling the centre of pressure contributed differing amounts to the total recovery response: the hip and spinal moments provided approximately 85% of the recovery, while the ankles contributed a small, but significant amount of about 15%. The differing contributions are based on the anatomical constraints and the functional requirements of the balance task. The onset of the joint moment was synchronous with the joint angle change, and occurred too early (56-116ms) to be as a result of active muscle contraction. Therefore, they concluded that the first line of defence was provided by muscle stiffness, not reflex-activated muscle activity.

2.7 Loading about lumbosacral joint

According to Lavender et al. (1999), the earliest attempts to quantify the spinal loads relied upon static sagittal plane analyses (Chaffin and Park, 1973), to determine the compressive force acting on the spine using a comparatively simple biomechanical model by computing the static moment introduced by the body and the load lifted. However, relative to the static models, the predicted external bending moments in the sagittal plane obtained with the dynamic models were found to be from 19 to 87% greater than those obtained using static analyses

(McGill and Norman 1986, Lavender et al. 1992). For the past few decades, the understanding of the external loads acting on the spine during dynamic lifting activities has been advanced through the use of inverse dynamic models, employed by many investigators (Frievalds and Cacioppo et al. 1986; McGill and Norman 1985; Schipplein, 1990; Leskinen et al. 1983; De Looze et al. 1992).

The knowledge of reaction forces and moments in relation to lifting postures are very helpful in understanding the mechanisms of low back disorder. The moment about the lumbosacral joint can be seen as a global measure of the load on the lower back since it determine the minimum trunk extensor force required from the back muscles, which mostly determines the compressive force on spinal motion segments (De Looze et al. 2000). The sagittal trunk flexion or extension, lateral trunk flexion or extension and axial trunk rotation are initiated, resisted, or controlled through muscle recruitment, and possibly through the resistance to elongation by the passive tissues. When the motion is not near the end range, it is principally those muscles that create the internal moments, equal in magnitude and opposite in direction to the external moments, necessary to stabilize and move the body. This makes the external moment a very good indicator of the minimum muscle force required to carry out the lift. Assuming that the muscles are the primary contributors to the mechanical loading of the spine, the moment serves as an good, but conservative indicator for the determination of spine loading. The moment was described as a conservative predictor for the reason that under some circumstances the moment will under predict the true spinal loads due to the kinetic redundancy within the system. This is especially true when there is a substantial co-contraction of antagonistic muscles (Lavender et al. 1999).

In 1999, Lavender et al. quantified the reaction forces and three directional external moments acting at the ankle, knee, and hip and L5/S1 joints during a sagitally symmetric and two asymmetric lifting tasks by using a dynamic linksegment biomechanical model. Higher forward bending moments in the sagittal plane of the spine were found from all tasks with heavier loads and at faster lifting speeds. Spine lateral bending and twisting moment were also higher during the mid-sagittal plane lifts with greater reach distance and faster lifting speed, respectively. The twisting moments on the spine were found to be the greatest when subjects lifted from in front and placed the load to the side, but were dependent upon the lifting speed and the load magnitude. Furthermore, the lateral bending moments increased during this same task with the heavier load. However, the spine lateral bending moments were greatest when lifting from one side to the other.

De Looze et al. (2000) investigated the trunk muscular reactions and low back torque in situations where the subject did not know the actual mass but only the mass would be within a certain range. They found that for a 6.5kg weight, the back muscle activation in grasping the box and the peak L5/S1 torque in actual lifting were 16% and 10% higher in the "unknown" compared with the "known" weight condition. For a 16.5kg weight, the back muscle activation during grasping were 10% lower and 10% higher during lifting in the "unknown" compared with the "known" weight condition. It was found that knowledge of the load had no effect on the activation of the abdominal muscles and they concluded that in the so-called "unknown" conditions, the risks of low back injury were increased in comparison with the conditions where the actual weight was known in advance. Their findings were supported by Butler et al.'s study concerning L5/S1 torques in 1993. Using masses from 0 to 30 kg, they found significantly higher peak torques at 0 kg in the unknown condition than in the known condition, but at 30 kg, no differences in torque was found. However, in contrast to Butler et al. and De Looze et al.'s study, Patterson et al. (1987) found higher peak torques in the unknown conditions regardless of the lifting weight. De looze et al. suggested that this finding was possibly due to their subjects aiming at lifting a mass that was even higher than the highest mass used, and they may not have known the upper limit of the range of masses.

The effect of unexpected loads on the back is probably a source of low back problems since excessive forces will be exerted upon the trunk (Marras et al., 1987) and there were higher shear forces when the load was unexpected (Magnusson et al., 1996). In fact, the sudden release of load can also have an adverse effect on the back as the externally applied load is suddenly released, the exerted muscle force created to maintain equilibrium would generate an unexpected acceleration and unbalance the body. Immediately, the body will respond by generating very large muscle forces, possibly larger than needed to regain the balance and this postural reaction required to regain balance could be hazardous to the low-back musculoskeletal system. However, only a few studies have been conducted. In order to gain a better understanding of the effect of sudden release of load on the low back, a study of the loads acting on the spine under different lifting conditions with different parameters such as lifting technique, lifting symmetry and lifting weight is required.

CHAPTER 3 MATERIALS AND METHODS

3.1 Subjects

Ten normal and healthy male volunteers age from 22 to 35 were recruited in this study. None of the subjects had any history of back injuries or significant back pain in the last two years. Anthropometric data including age, body height, body-weight and knee height (vertical distance from subject's femoral epicondyle to the floor) of each subject were documented.

3.2 Equipment

In order to study the effect of sudden release of load on the lumbar spine under different lifting postures and lifting conditions, a custom designed experimental set up was used to simulate a "guillotine" mechanism, producing a sudden release of load. A force plate (Kistler 9281C, Kistler Instrumente AG Winterthur, Switzerland) was used to measure kinetic data, a motion analysis system was used to measure kinematic data and two electromyography (EMG) systems were used to measure muscle activities, as detailed below.

3.2.1 Custom designed experimental set up

The custom designed experimental set up consisted of an AC power supply, accelerometer (Bruel and Kjaer type 4236, Denmark), electromagnet, optical sensor, sensor blocker, and a number of 10N dead weights. A lightweight fake load, 8N with dimension 0.31m x 0.22m x 0.23m, simulating a box contained packs of A4 photocopy paper was connected to a dead weight via the electromagnet to form a "guillotine" or quick release mechanism (Figure 3.1). By combining a switch, which was manually controlled by the experimenter, the power to the electromagnet would be turned off when the sensor blocker (Figure 3.2) passed by the optical sensor and that would release the electromagnet from the metal plate which connected to the pulley system to form a sudden unload condition. The accelerometer was installed inside the fake load to monitor the instantaneous acceleration and so the onset of the sudden unload could be recorded. The fake load could also be positioned on the left side of the force plate for symmetric and asymmetric lifting trials (Figure 3.2 and 3.3).

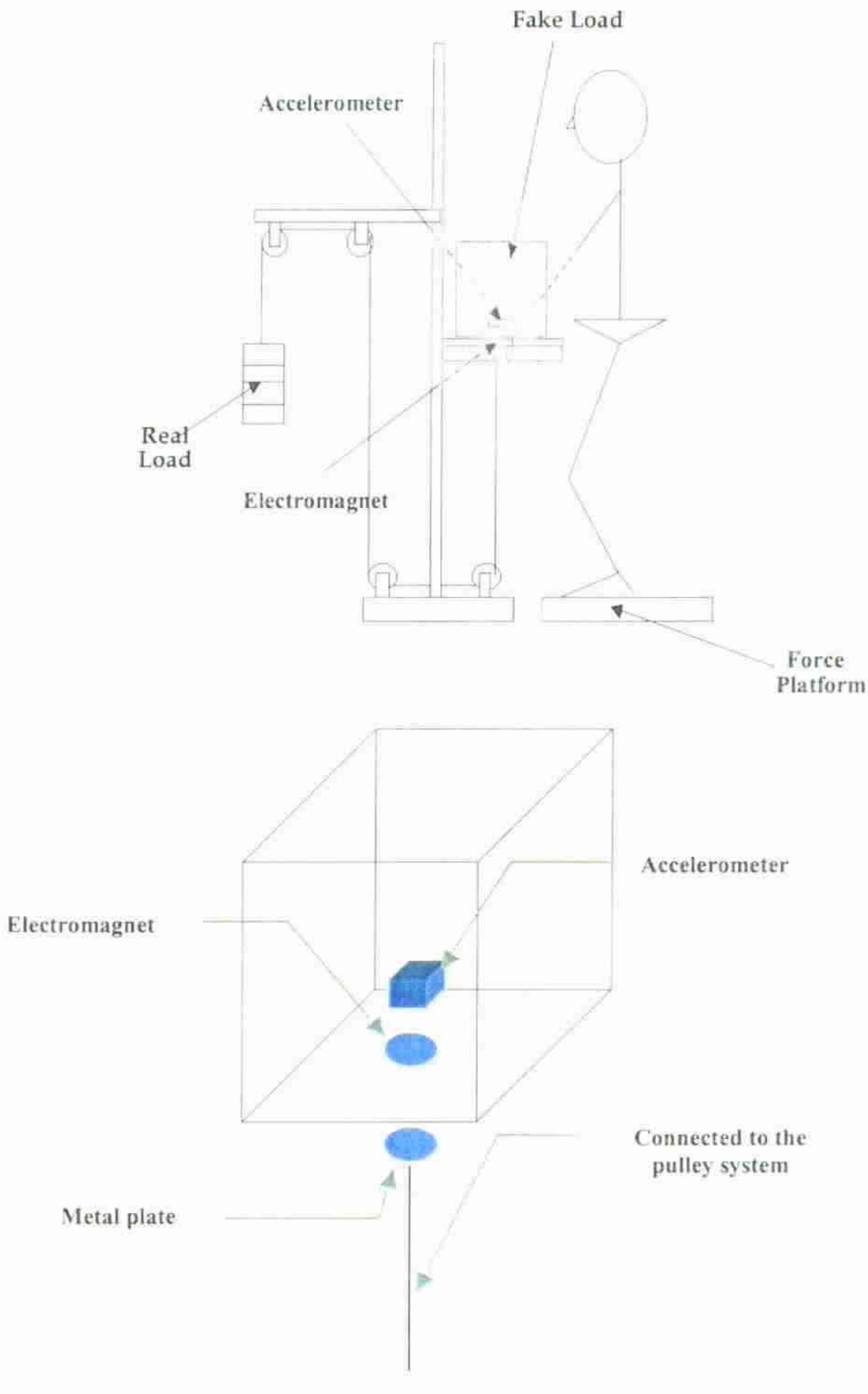


Figure 3.1 Schematic illustrations of the experimental set up with enlarged internal diagram showing how the fake load was connected to the pulley system via an electromagnet and a steel plate to form a "guillotine" mechanism.

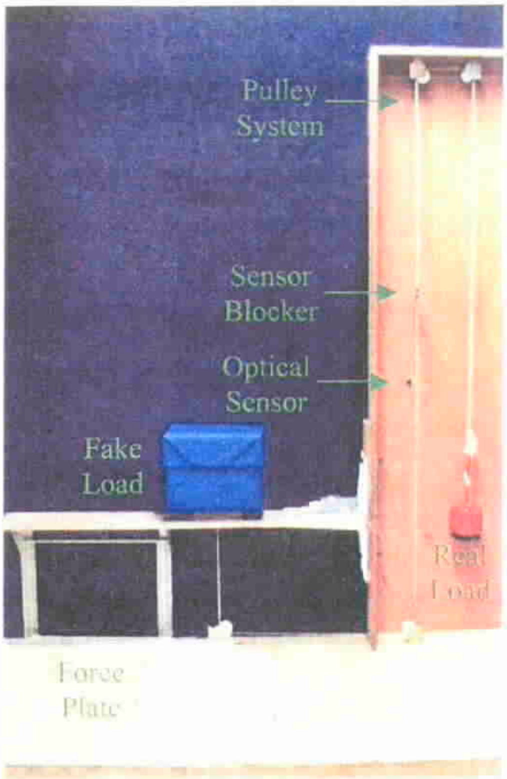


Figure 3.2 Experimental set up for symmetric lifting trials

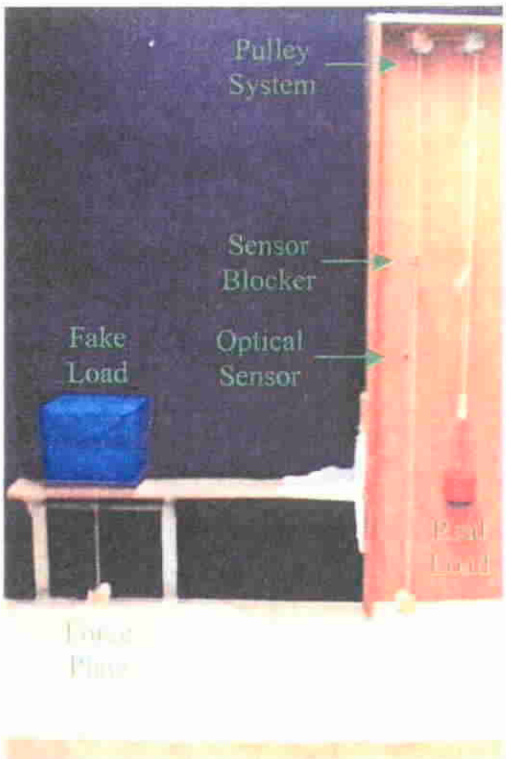


Figure 3.3 Experimental set up for asymmetric lifting trials

3.2.2 Motion analysis system

A three-dimensional motion analysis system, (Vicon 370, Oxford Metrics Ltd., UK) was used to monitor the three-dimensional coordinates of the lightweight spherical retro-reflective markers attached to the subject. Two sizes of markers (25mm and 10mm diameter) were used for the study since the system could not distinguish the large makers and captured them as one when they were placed close together; small markers that could be differentiated were used in these cases. The system tracked the trajectories of the markers in the field of view of four cathodes charged diode (CCD) cameras. Since the larger the viewing angles subtended at the cameras, the higher the accuracy of the captured image of the markers, different combinations of the camera locations were tried, and the optimum combination found was to install the four cameras such that all viewing angles subtended were larger than 30° (Table 3.1). The data of the CCD cameras were sampled at 120Hz. All the image data were collected and the 3D coordinates of each marker were anatomatically reconstructed by the motion analysis system software. Data were then saved for further analysis.

Table 3.1 Viewing angles formed by the four CCD cameras

Camera	1 & 2	1 & 3	1 & 4	2 & 3	2 & 4	3 & 4
Spatial angle between cameras	35°	55°	65°	33°	42°	35°

Experimental Calibration

Prior to each experiment, the motion analysis system was calibrated. Calibration consisted of two parts, static calibration and dynamic calibration. Once all the cameras had been linearized and positioned, the motion analysis system was calibrated to determine all camera positions and orientations relative to an origin and set of axes. The static calibration was conducted with a L-frame that fitted precisely over the corner of the force plate (Figure 3.4). The origin of the global Cartesian coordinate system of the motion analysis system was defined by the angle of the L-frame. Corner 341 was located at the corner of the force plate with the X-axis aligned with the medio-lateral direction, the Y-axis aligned with the antero-posterior direction and the Z-axis aligned with the vertical direction. The dynamic calibration was conducted by sweeping a 500mm wand around the whole working volume, which was defined to be (0.6m x 0.8m x 1.8m). The root mean square error was found to be 0.21% (1.1mm).

Analogue channels

The motion analysis system has the ability to sample up to 64 analogue channels. All analogue signals are synchronously captured with the video frames. In the laboratory, a force plate was connected to the motion analysis system through eight of the available analogue channels for measuring ground reaction force, and the accelerometer was connected through one of the available analogue channels for monitoring instantaneous acceleration. The electromyography signals were sampled through 18 of the available analogue channels for measuring muscle activities.

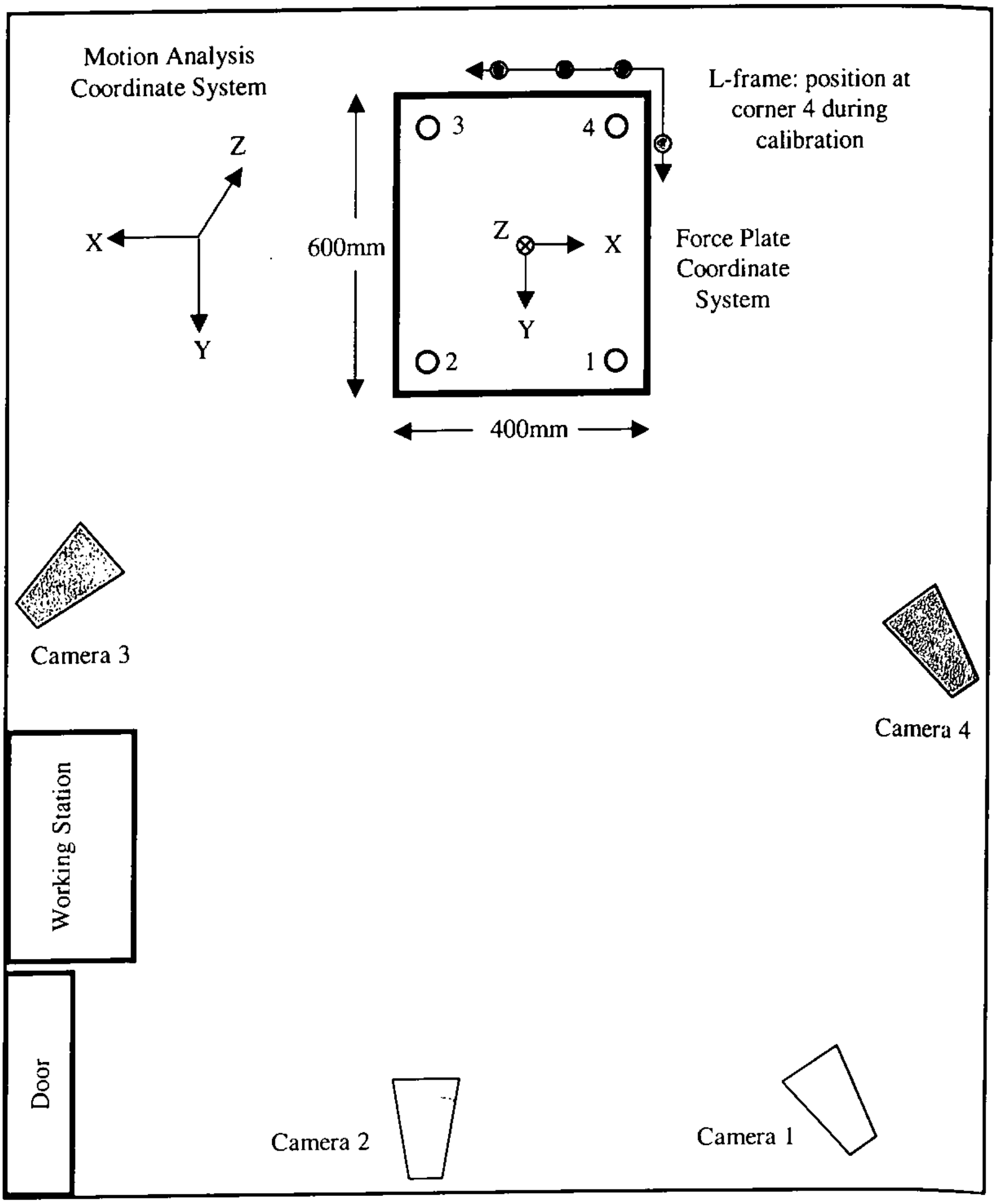


Figure 3.4 Floor plan of the laboratory

3.2.3 Force platform

During the experiment, the subject stood on a piezoelectric force plate (Kistler 9281C, Kistler Instrumente AG Winterthur, Switzerland). The force plate was utilized to monitor the three components of instantaneous ground reaction force, location of centre of pressure and vertical twisting moment. For the control unit of the force plate (Kistler, 5233A), figure 3.5, 250N was chosen for the range of horizontal shear force while 2.5kN was chosen for axial compressive force. The force plate data were sampled at 3000Hz.

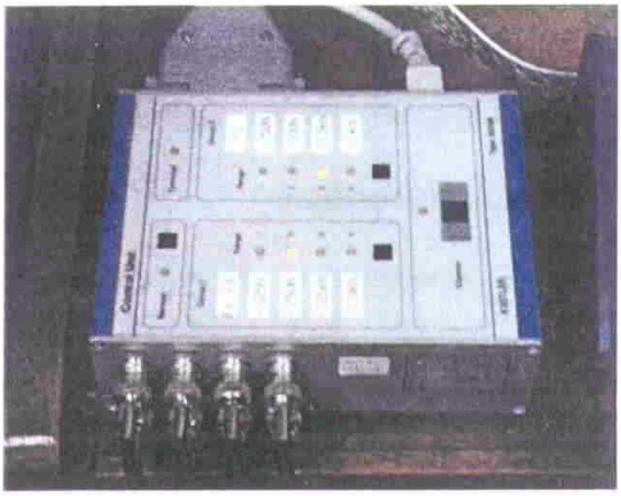


Figure 3.5 Control unit of the force plate

3.2.4 EMG systems

A telemetric Electromyography (EMG) system, (Telemyo, Noraxon, U.S.A., Inc) (Figure 3.6) and an online EMG system, RECEMG (Rehabilitation Engineering Centre, The Hong Kong Polytechnic University, HK) (Figure 3.7) were used for the measurement of muscle activities during the experiments. Both systems have a notch filter of 50Hz and a preamplification of 2000.

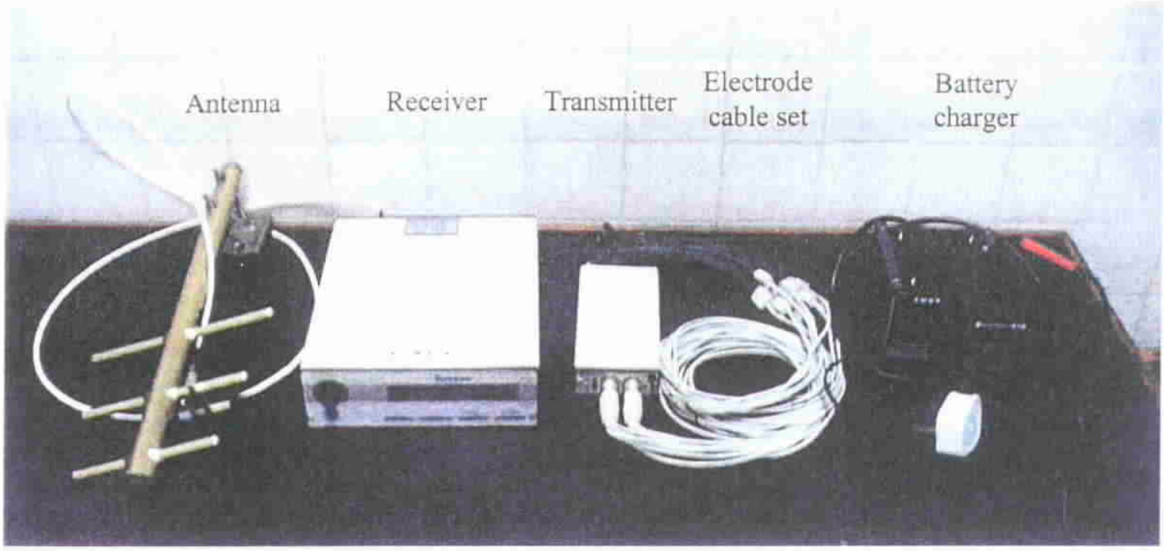


Figure 3.6 Telemetric EMG system (Telemyo, Noraxon, U.S.A. Inc)

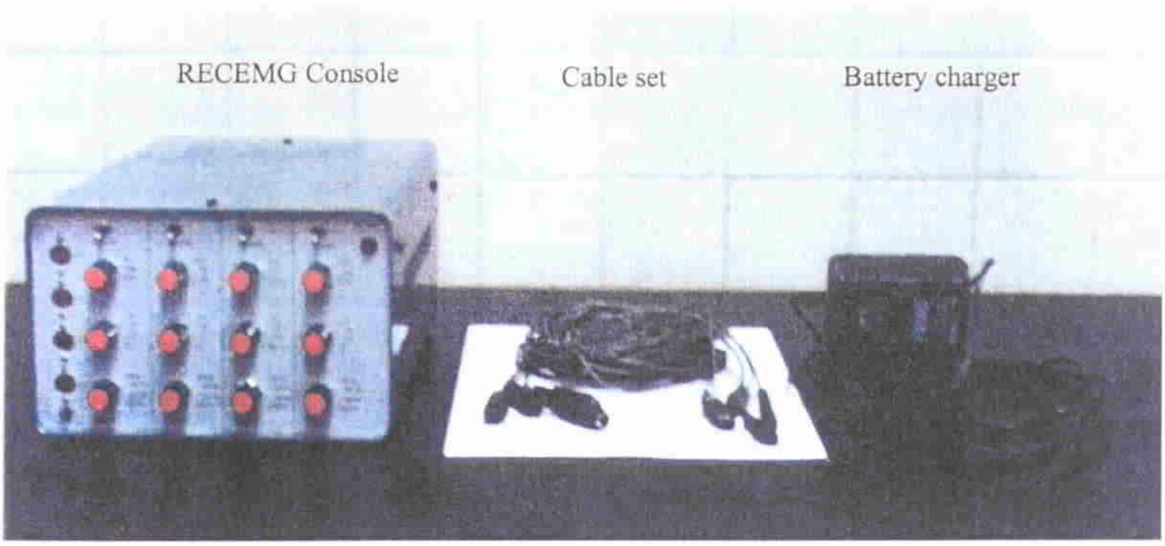


Figure 3.7 Online EMG system (RECEMG, PolyU, HK)

3.3 Experimental Protocol

For each subject, three lifting postures, each with a total of four different lifting weights were investigated in a total of 12 tests (Figure 3.8):

1. Symmetric squat lifting style with four different lifting weights
2. Symmetric stoop lifting style with four different lifting weights
3. Asymmetric (30° to the left from the mid-sagittal plane) stoop lifting style with four different lifting weights

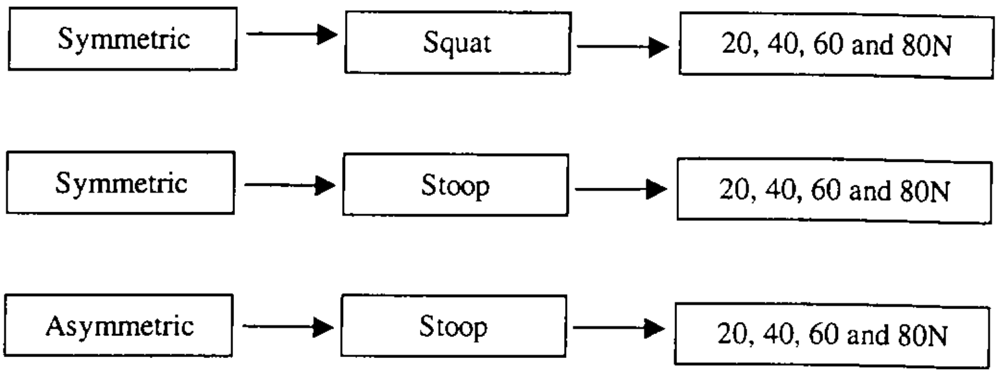


Figure 3.8 Three postures, each with a range of lifting weights

Subjects were instructed to perform lifting tasks with different lifting weights, symmetries and lifting postures, according to a randomly assigned sequence. The load initially at knee height and was set to be released at either the 3rd, 4th, 5th of the five lifting cycles at the lower one-third of the lifting distance or no release at all. An extra trial was conducted whenever no release was chosen for that trial.

The starting level of the fake load was always at the subject's knee height and the lifting distance was defined as the total vertical distance from the subject's knee height to the height at final lifting position. The final lifting position was defined as the subject's standing elbow height. The load release height was set at the lower one third of the total lifting distance.

3.4 Electromyography Measurement

3.4.1 Signal collection

Nine pairs of muscles were investigated. Bilateral latissimus dorsi (LD), lumbar erector spinae (ES), external oblique (EO), internal oblique (IO) and rectus abdominis (RA) were monitored by the telemetric EMG system (Telemyo, Noraxon U.S.A. Inc) while bilateral biceps femoris (long head) (BF), tibialis anterior (TA), rectus femoris (RF) and gastrocnemius (lateral head) (G) were monitored by the online EMG system (RECEMG, REC PolyU, HK). All muscles and a prominent bony landmark as a ground were located via palpation and the use of bony landmarks. Surface electrodes were placed for EMG data collection as shown in table 3.2, figures 3.9 and 3.10. This electrode placement has been shown to maximize signal-to-noise ratio and minimize levels of cross-talk (Cholewicki & McGill 1996, Cholewicki et al. 1997). All the muscles were located and marked, the area was shaved and abraded to reduce the skin resistance and scrubbed with alcohol to make the electrodes fully attach to the skin. These surface electrodes were filled with electrical conducting gel and secured to the skin at these sites by adhesive soft pad disks. The electrodes were also reinforced by micropore. The resistance of the skin was measured with digital voltmeter and the area was abraded and scrubbed again if the resistance was found to be more than 10kΩ.

Table 3.2 Location of electrode placements * *This electrode placement has been shown to maximize signal-to-noise ratio with minimum levels of cross-talk (Cholewicki & McGill 1996, Cholewicki et al. 1997)*

Name of Muscle	Location of Electrode Placement
*Latissimus Dorsi	Approximately three finger-breadths distal to and along posterior axillary fold
*Erector Spinae	Approximately 3 cm lateral to L3 spinous process
*Rectus Abdominis	Approximately 3 cm lateral to the umbilicus
*External Oblique	Approximately 15 cm lateral to the umbilicus
*Internal Oblique	Superior to the inguinal ligament and just below external oblique electrodes
Biceps Femoris	The midpoint of a line between the fibula head and the ischial tuberosity
Rectus Femoris	On the anterior aspect of the thigh, midway between the superior border of the patella and the anterior superior iliac spine
Tibialis Anterior	Approximately four finger-breadths below the tibial tuberosity and one finger-breadth lateral to the tibial crest
Gastrocnemius	Approximately one hand-breadth below the popliteal crease on the lateral mass of the calf
Ground	Fibula head

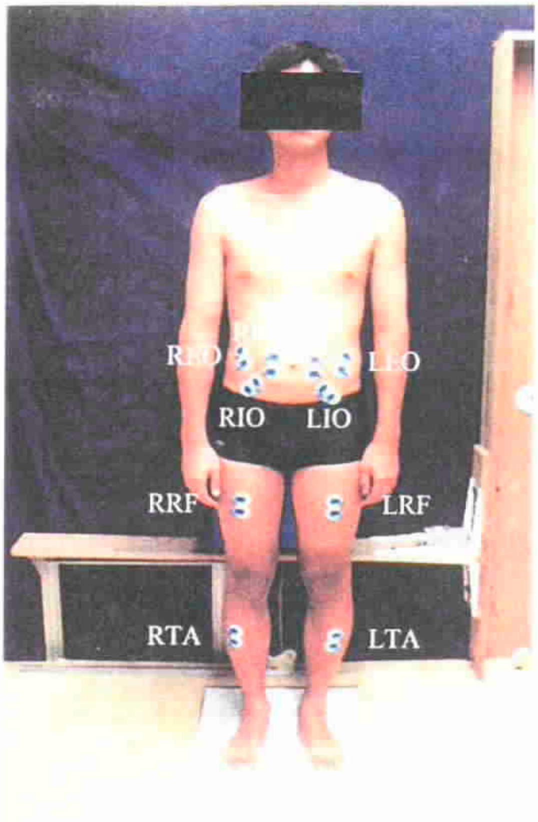


Figure 3.9 Anterior view of the location of electrode placements

Figure 3.10 Posterior view of the location of electrode placements

In order to make sure that EMG signals were from the designated muscles, the signals from each pair of electrodes were functionally tested by asking the subject to perform specific movements. (Table 3.3)

Table 3.3 Tasks to functionally test if each pair of electrodes was correctly placed *(Jeffrey R. Cram, Introduction to surface EMG, 1998)*

Name of Muscle	Task
Latissimus Dorsi	Medially rotate the arm
Erector Spinae	Forward flexion and return to midline of the torso
Rectus Abdominis	Tighten the abdomen
External Oblique	Lateral bending
Internal Oblique	Rotation of the torso
Biceps Femoris	Flex the knee
Rectus Femoris	Squat slightly
Tibialis Anterior	Dorsiflex the foot
Gastrocnemius	Plantar flex the foot

3.4.2 Signal processing

Eighteen channels of EMG were recorded using bipolar, Ag-AgCl, surface disposable electrodes. The electrodes were placed with a centre-to-centre spacing of approximately 3cm over the nine pairs of muscles. Raw EMG signals were 10Hz to 500Hz band-pass filtered and preamplified by 2000. All signals were then transmitted via shielded cables to the main EMG system and were converted by a 12 bit A/D card at a sampling rate of 3000Hz.

The signals were further smoothed by 50Hz lowpass filtering to eliminate artifacts that would alter the EMG signal. This filtered signal was full-wave rectified and processed by a moving average technique. The width of the moving average window was 25ms and each data point was the average of the rectified digital signal that occurred in the 25ms window. The window then moved forward by 1ms and the next EMG point was the average of this 25ms window (Hodges & Bui 1996)

3.4.3 Onset and termination of muscle activity identification

The onset and termination of muscle activity was identified by comparing the processed EMG signal with the computed threshold. With the stimulation of sudden release of load, a contraction muscle response was identified if the processed EMG signal exceeded a threshold of three standard deviations above a baseline mean. However, this would be identified as a relaxation muscle response if the processed EMG signal was below the threshold. The baseline mean of each muscle was computed using the median mean EMG signal magnitude in five sequential 100-ms windows, which were picked from the resting period of the EMG recording of each muscle. All of the 500-millisecond windows were picked from the period before the lifting task started. The resting period was sometimes less than 500 ms and in that case, the baseline mean would be computed using the median mean EMG signal magnitude of the picked window, and noted in the results. The standard deviation of the 100ms window corresponding to the median mean was used to compute the three standard deviation threshold. The potential effect of outliers resulting from EMG contamination by ECG or other artifacts can be minimized by using the median (Stokes et al. 2000). Both signal filtering and signal smoothing were done by a signal processing software (Matlab, The MathWorks Inc., Natick, MA); the program can also determine the threshold automatically with an input of the 500-ms window.

Since the muscle activities after the sudden release of load was of interest, both the full wave rectified EMG signal, the smoothed EMG signal, accelerometer and the threshold line were displayed together using a Matlab signal processing tool, SPTOOL, which allows import, analysis and manipulation of signals.

3.4.4 Data analysis

In order to analyse the EMG signals systematically, the following terms were defined:

- **Threshold line:** A line at the level of three standard deviations above a baseline mean.
- **Onset of sudden unload**: The onset time of sudden release of load was registered by the accelerometer and defined as at the time when the accelerometer showed a spike.
- **Onset of contraction response**: The muscle was defined as responded to the stimulation by contraction when the muscle activity (smoothed EMG signal) exceeded the threshold line after sudden release of load (Figure 3.11A & 3.11B).
- **Termination of contraction response**: A contraction muscle response was defined as terminating when the muscle activity dropped below the threshold line (Figure 3.11A & 3.11B).
- **Onset of relaxation response**: The muscle was defined as responding to the stimulation by relaxation when the muscle activity (smoothed EMG signal) was below the threshold line after sudden release of load (Figure 3.12A & 3.12B).
- **Termination of relaxation response:** A relaxation muscle response was defined as terminating when the muscle activity exceeded the threshold line (Figure 3.12A & 3.12B).
- **No response**: The muscle was defined as not responding to the stimulation when the muscle activity was not affected by the stimulation (Figure 3.13A & 3.13B).
- **Latency (t_L):** When a response was detected, its latency was measured as the time from the onset of the release of load to onset of the EMG response.
- **Duration of response (t_R)**: The duration of response was defined as the time period from the onset to the termination of the EMG response.

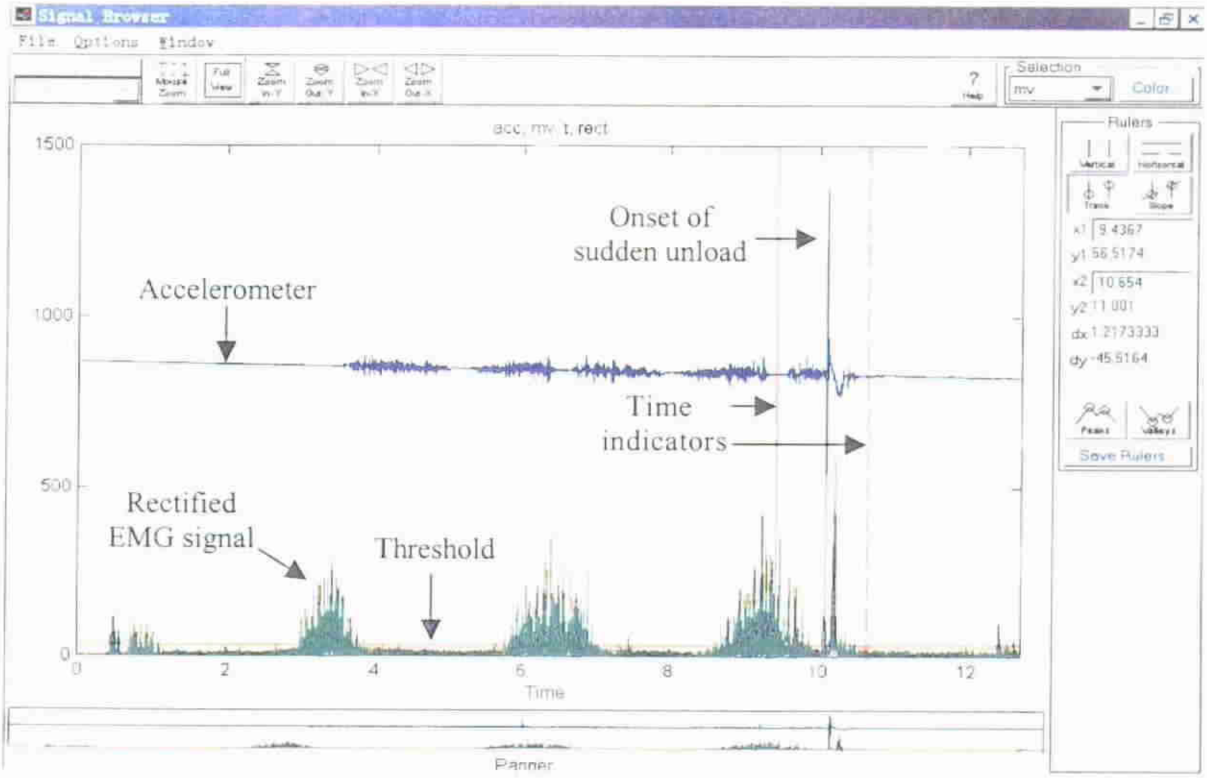


Figure 3.11A An illustration of a **contraction response** for onset and termination of muscle activity identification

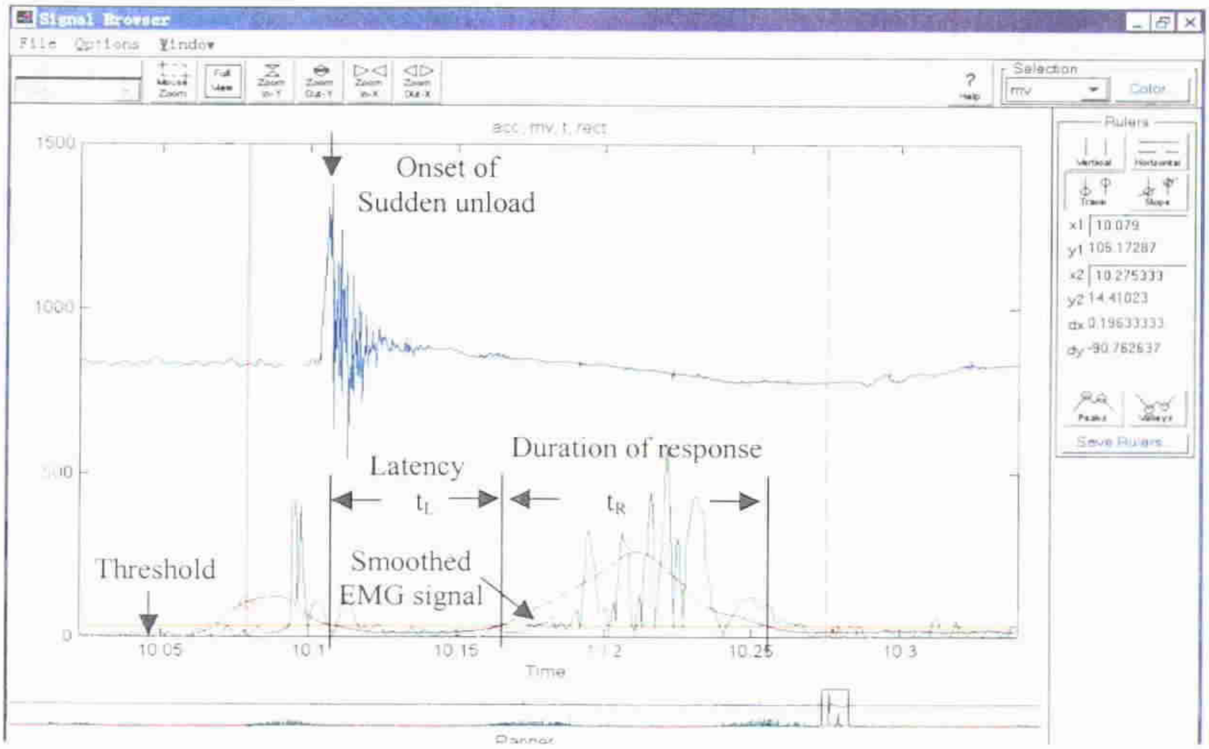


Figure 3.11B Enlarged EMG signal of a **contraction response** for onset and termination of muscle activity identification

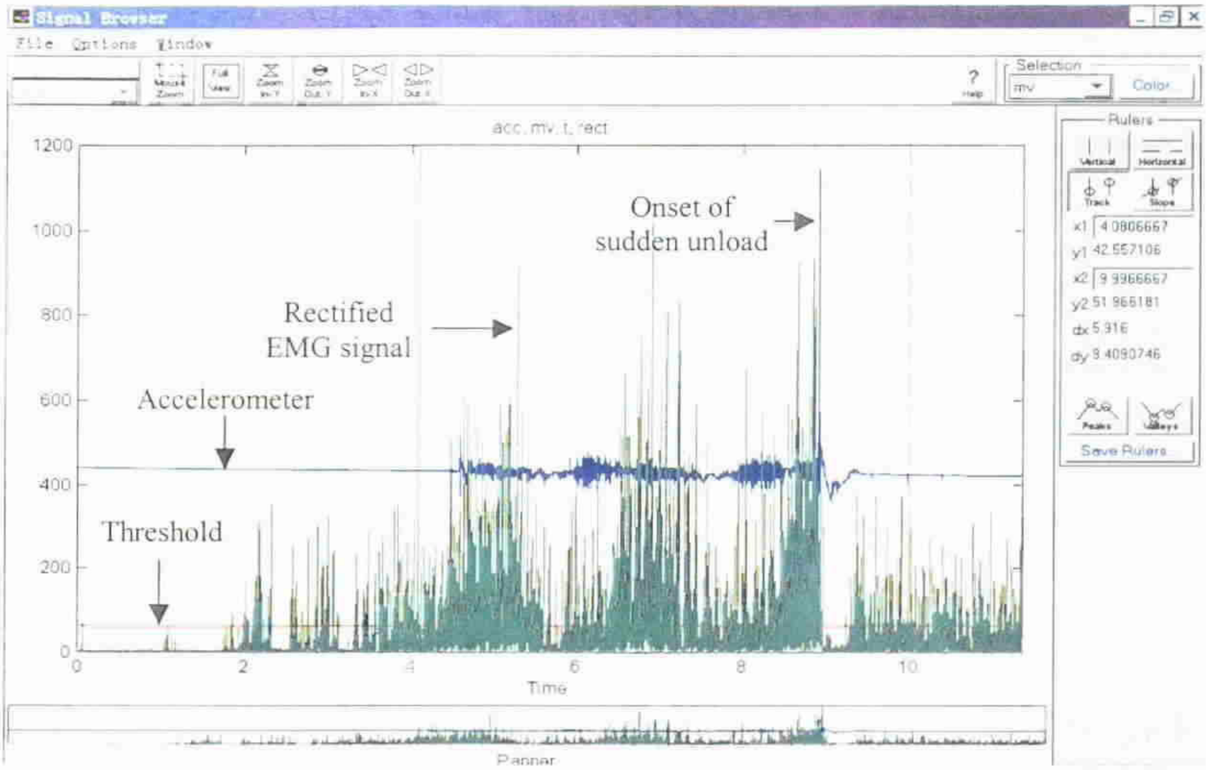


Figure 3.12A An illustration of **relaxation response** for onset and termination of muscle activity identification

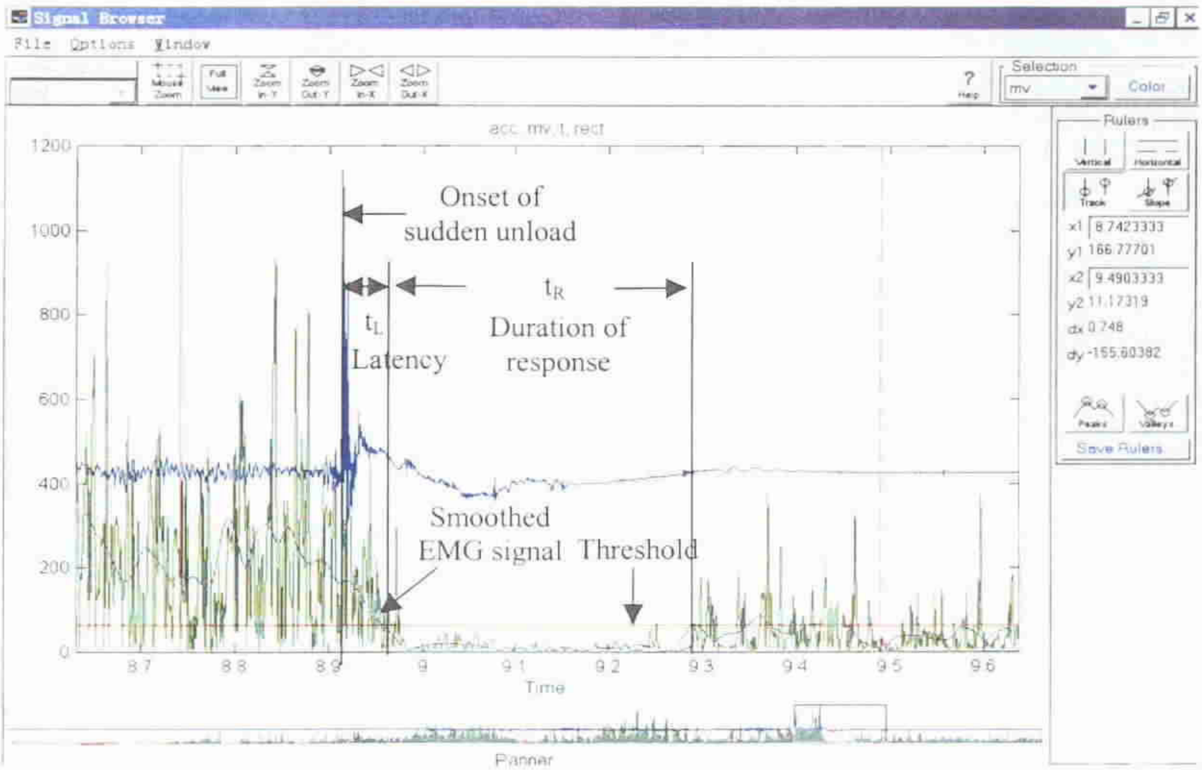


Figure 3.12B Enlarged EMG signal of a **relaxation response** for onset and termination of muscle activity identification

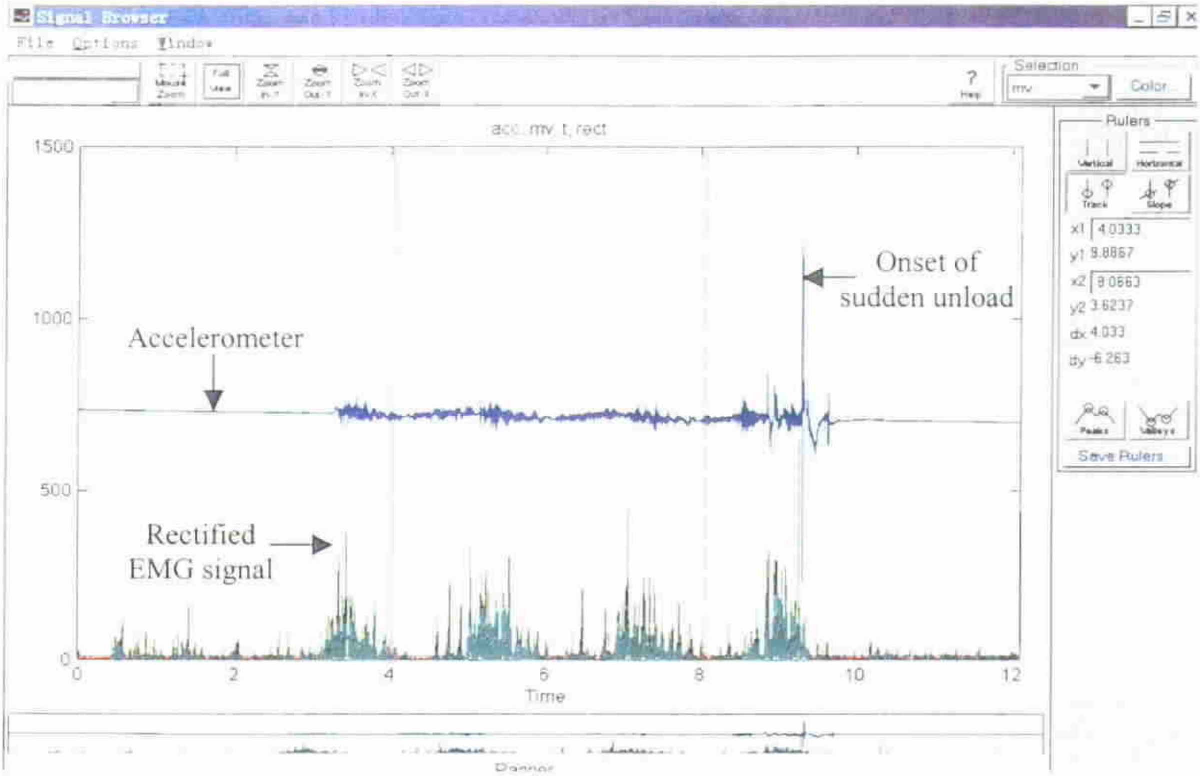


Figure 3.13A An illustration of an EMG signal which has **no response** to the stimulation and no termination of muscle activity need to be identified

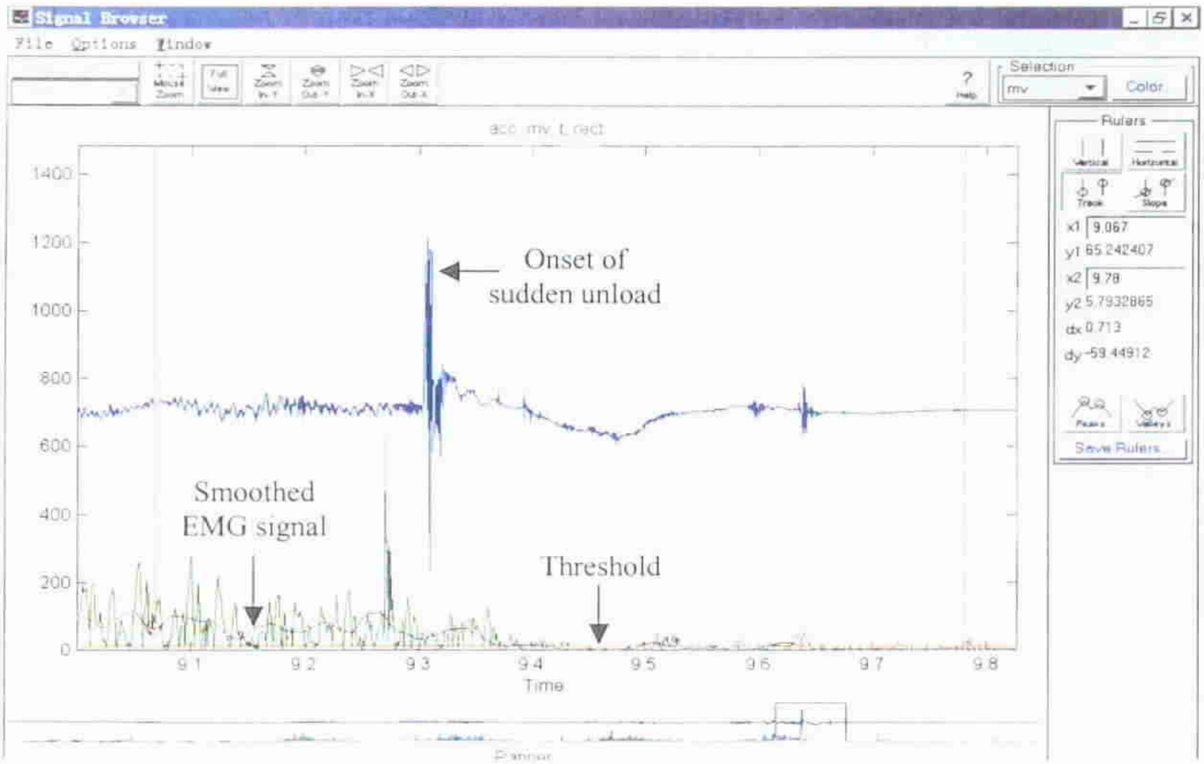


Figure 3.13B Enlarged EMG signal which showed **no response** to the stimulation

3.5 Centre Of Pressure Measurement

The ground reaction force measured by the force plate was sampled using eight analogue channels for different force components. The centre of pressure (COP) was calculated by using equations provided by the manufacturer (Kistler Instrumente AG Winterthur, Switzerland) (Appendix 2).

Before the dynamic lifting tests started, each subject was asked to stand on the force plate with feet approximately shoulder width apart. Subjects were then asked to lift the fake load with different lifting postures (squat lift and stoop lift) at different positions (closer to or farther from the fake load) until they found the most comfortable position for each lifting posture. The coordinates of the position of the subject's big toes and heels were recorded for both squat lift and stoop lift (Figure 3.14).

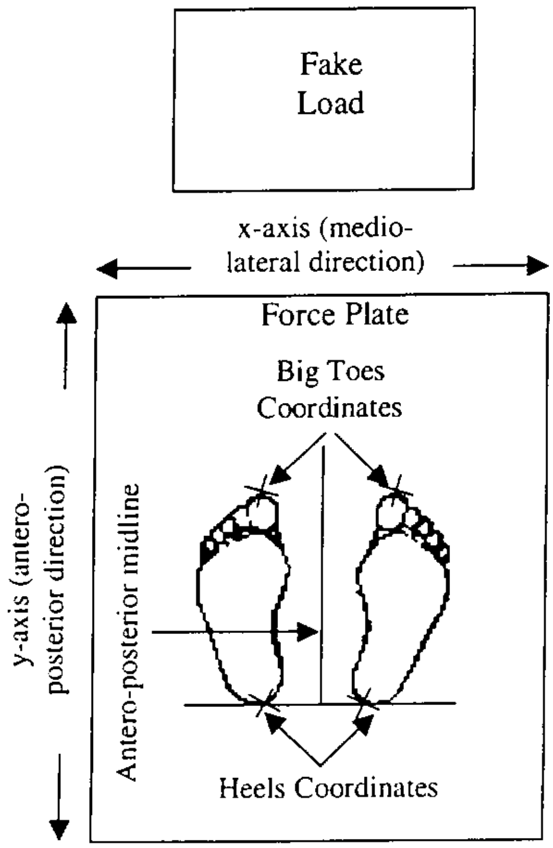


Figure 3.14 Coordinates of foot positions

During dynamic lifting tasks, the centre of pressure changed continuously both in antero-posterior and medio-lateral directions. The posterior boundary and the antero-posterior midline of the base of support were calculated by the coordinates of the heels. The antero-posterior coordinates of the centre of pressure were recorded as the perpendicular distance from the line joining the heels while the medio-lateral coordinates of the centre of pressure were recorded as the perpendicular distance from the antero-posterior midline. The coordinates of the big toes were marked for reproduction of the foot positions by the subject.

3.5.1 Data analysis

For the dynamic situation, there are body weight, ground reaction force and inertia effects due to the motion. The ground reaction force will be equal but opposite to the vector sum of the body weight and the inertial effects. The movement of the centre of pressure (COP), should be within the base of support if the subject is able to maintain his equilibrium. Imbalance was judged to occur when the heels lost contact with the ground or when a compensatory step was made to prevent falling. If the subject does not move his feet, the more the COP moves towards the boundary of the base of support, i.e. the line joining the heels, the more likely the subject is going to fall. Therefore, if the distance between the COP and the posterior boundary of the base of support after sudden release of load was observed to be inversely proportional to the lifting weight, then the maximum weight that can be lifted without loss of balance under sudden release of load situation can be predicted by extrapolation.

In order to analyse the COP displacements systematically, the following terms were defined (Figure 3.15):

- **Onset of sudden unload:** The onset time of sudden release of load was registered by the accelerometer.

- **Antero-posterior COP response time (Tyr):** Duration of time measured from the onset of sudden unload to the point where antero-posterior COP displacement value starts to drop rapidly (COP shifts sharply towards the heels).
- **Antero-posterior COP peak time (Typ):** Duration of time measured from the onset of sudden unload to the point where the antero-posterior COP displacement value reaches a minimum after the sharp drop and starts to rise (COP shifts back towards the toes).
- **Medio-lateral COP response time (Txr):** Duration of time measured from the onset of sudden unload to the point where medio-lateral COP value starts to rise sharply (COP shifts sharply to the right side).
- **Medio-lateral COP peak time (Typ):** Duration of time measured from the onset of sudden unload to the point where medio-lateral COP displacement value reaches a maximum after the sharp rise and starts to drop (COP shift back to the left side).
- **Posterior limit of COP displacement** (L): The antero-posterior COP displacement value at Typ.

A typical example of the antero-posterior (COPy) and medio-lateral (COPx) COP displacement during the sudden release cycle under symmetric squat lift is shown (Figure 3.15). The figure shows the recorded parameters and the complete lifting cycle immediately prior to sudden release for comparison. A typical example of the antero-posterior COP excursion relative to the line joining the heels after sudden release of load of symmetric squat lift with different lifting weights is also shown (Figure 3.16).

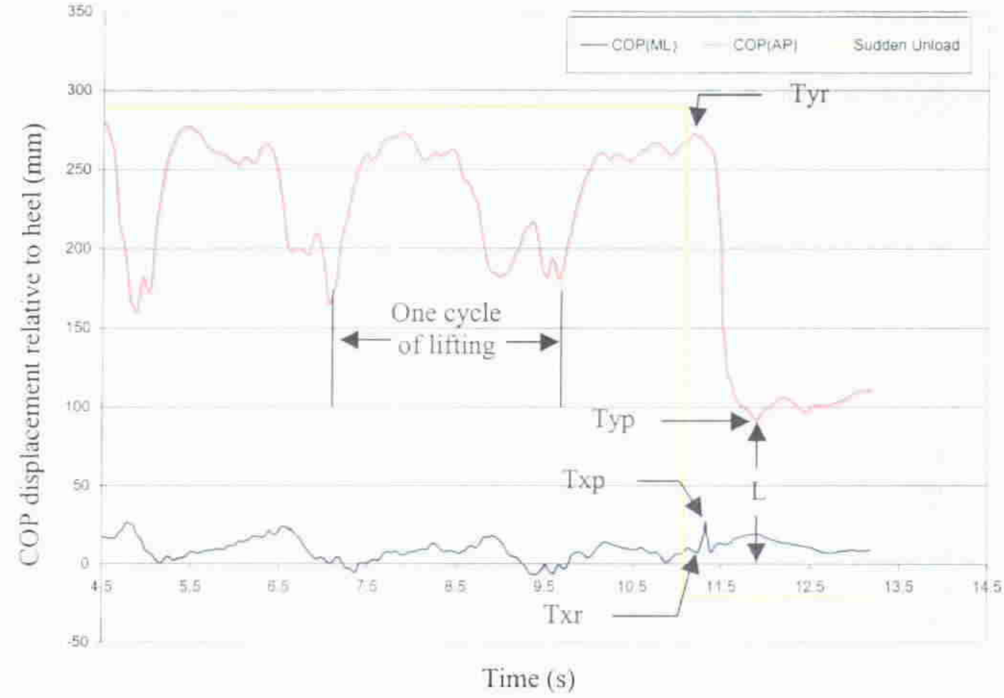


Figure 3.15 A typical example of COP excursion (ML=medio-lateral; AP=antero-posterior)

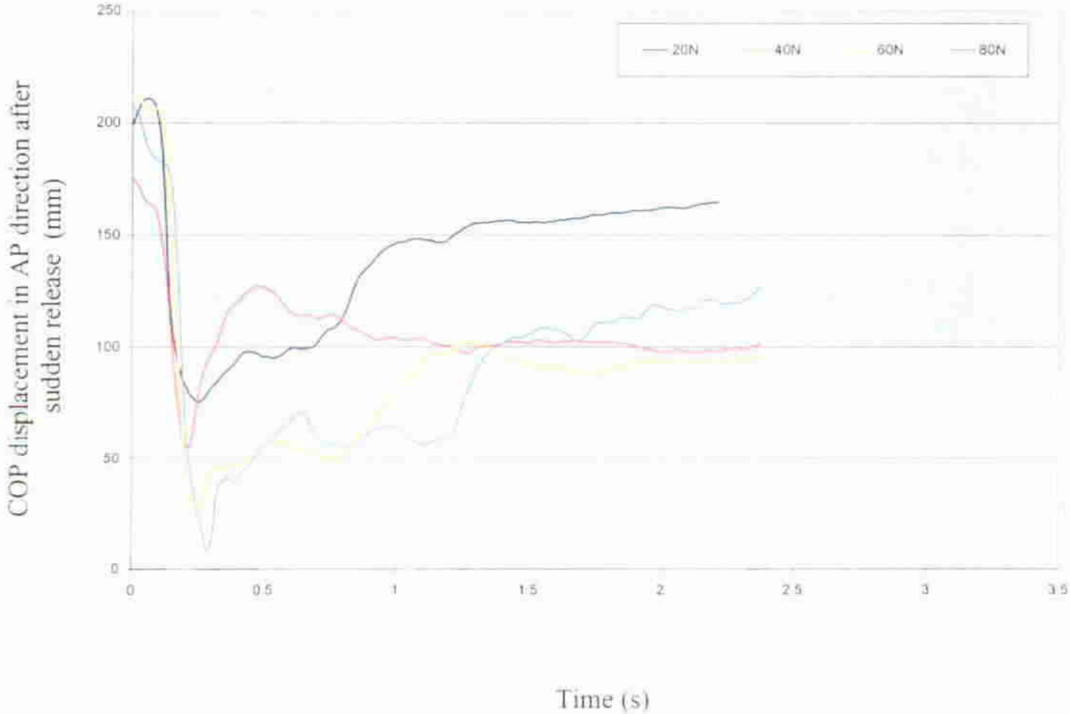


Figure 3.16 A typical example of the antero-posterior COP excursion for symmetric squat lift with different lifting weights after sudden release

3.6 Load At The L5/S1 Spinal Disc Centre

This study employed a biomechanical model consisting of seven body-segments, namely the left and right feet, shanks, thighs and the pelvis to determine the net moment at the L5/S1 spinal disc centre. Each body segment was defined (Table 3.4) and considered as a kinematically rigid body. They were considered to be connected at the ankle, knee and hip joint centres.

Table 3.4 Definition of body segments

Body Segment	Definition
L. and R. Foot	The bottom of the foot to the medial malleolus
L. and R. Shank	The medial malleolus to the lateral tibial condyle
L. and R. Thigh	The lateral tibial condyle to the greater trochanter
Pelvis	The greater trochanter to the mid point of posterior superior iliac spines

In order to calculate the net force and net moment acting about the lumbosacral joint by using an inverse dynamics approach, anthropometric, kinematic, and kinetic data were required. Figure 3.17 showed the algorithms used in determining the net force and net moment acting about the lumbosacral joint.

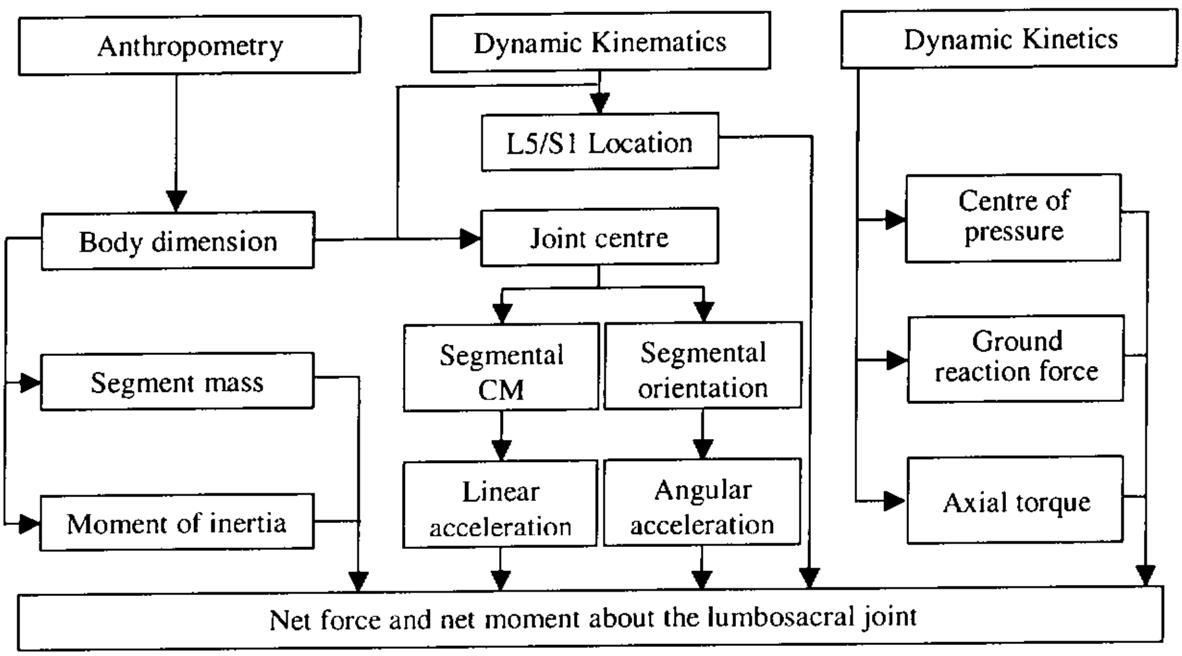


Figure 3.17 Flow diagram of the algorithms used in determining the net force and net moment about the lumbosacral joint

3.6.1 Anthropometric data

Anthropometric data consisted of body dimensions, segment masses and moments of inertia. Twenty-four linear regression equations developed by Vaughan et al. (1992) were adopted to calculate the masses and moments of inertia of the left and right feet, shanks and thighs. According to Vaughan et al. (1992), the accuracy of the mass and moment of inertia calculation was improved by using a dimensional consistency approach, which is based not only on body weight but also body segment dimensions. The estimation of the mass and moment of inertia of the pelvis was based on the approach of Zheng et al. (1998). All the equations for calculating of mass and moment of inertia of each segment are shown in Appendix 5 and all the definitions of the abbreviations of the equations can be found in Appendix 4.

A total of twenty-two body segment dimensions were defined (Table 3.5), which included body weight, anterior superior iliac spine (ASIS) breadth, foot breadth, shank and thigh circumference, knee and malleolus diameter, malleolus height, length of each segment and pelvis width. Body weight was measured by weight balance, knee and malleolus diameter were measured by caliper, shank and thigh circumferences were recorded using a tape, and all other parameters were measured using the motion analysis system.

Table 3.5 Definition of body segment dimensions

Name	Symbol	Definition
Body weight	A1	Subject's total body weight with swimsuit on only
ASIS breadth	A2	The horizontal distance between the ASIS
L. thigh length R. thigh length	A3 A4	The vertical distance between the greater trochanter and the lateral femoral epicondyle
L. thigh circumference R. thigh circumference	A5 A6	The circumference of the thigh at a level midway between the greater trochanter and the lateral femoral epicondyle
L. shank length R. shank length	A7 A8	The vertical distance between the lateral femoral epicondyle and the lateral malleolus
L. shank circumference R. shank circumference	A9 A10	The maximum circumference of the shank
L. knee diameter R. knee diameter	A11 A12	The maximum breadth of the knee across the femoral epicondyles
L. foot length R. foot length	A13 A14	The distance from the posterior margin of the heel to the tip of the longest toe
L. malleolus height R. malleolus height	A15 A16	The vertical distance from the bottom of the foot to the lateral malleolus
L. malleolus diameter R. malleolus diameter	A17 A18	The maximum distance between the medial and lateral malleoli
L. foot breadth R. foot breadth	A19 A20	The breadth across the distal ends of metatarsals I and V
Pelvis length	A21	The vertical distance between the pubis and the iliac crest
Pelvis width	A22	The horizontal distance between the abdomen and the back at the iliac crest level

In order to measure body segment dimensions using the motion analysis system, an anthropometric marker set consisting of twenty-seven retro-reflective markers was established (Table 3.6). Figure 3.18 and figure 3.19 show the anterior and posterior view of the anthropometric maker placements on a subject during the experiment.

Table 3.6 Anatomical location of anthropometric marker placements

Marker No.	Position	Marker No.	Position
M1	L. Lateral Mallelous	M18	L. Greater Trochanter
M2	R. Lateral Mallelous	M19	R. Greater Trochanter
M3	L. Medial Mallelous	M20	Pubic
M4	R. Medial Mallelous	M21	L. ASIS
M7	L. Lateral Femoral Epicondyle	M22	R. ASIS
M8	R. Lateral Femoral Epicondyle	M23	L. Iliac Crest
M9	L. Distal End of Metetarsals V	M24	R. Iliac Crest
M10	R. Distal End of Metetarsals V	M25	Mid Point of Iliac Crest (Dorsal)
M11	L. Heel	M26	Mid Point of Iliac Crest (Frontal)
M12	R. Heel	M27	L. Distal End of Metetarsals I
M16	L. Medial Femoral Epicondyle	M28	R. Distal End of Metetarsals I
M17	R. Medial Femoral Epicondyle		

Figure 3.18 Anterior view of the anthropometric marker placements

Figure 3.19 Posterior view of the anthropometric marker placements

3.6.2 Kinematic data

Kinematic analyses are concerned with the description of the geometric and time dependent aspects of motion in terms of displacement, velocity and acceleration without dealing with the factors causing the motion. In this study, kinematic data consisted of the linear and angular accelerations of the seven defined body segments (left and right feet, shanks, thighs and pelvis).

Linear acceleration of body segments

In order to calculate the linear acceleration of the centre of mass of different body segments along different axes, the centre of mass of all seven body segments need to be found first. In turn, joint centres need to be located for the calculation of segmental centres of mass. The locations of all joint centres (left and right ankle joints, knee joints and hip joints) and the centre of masses (left and right feet, shanks and thighs) were found by adopting the uvw reference system and the equations developed by Vaughan et al. (1992).

The uvw reference system is a reference coordinate system which is formed by attaching a set of retro-reflective markers to the subject to define each body segment, together with equations and anthropometric data to calculate all the joint centres and all the centre of masses, except for the pelvis. The centre of mass of the pelvis was found by adopting an equation from Winter (1990). Finally, the linear acceleration of the centre of mass of all the body segments along x, y and z axis were found by the 5-point method. All the equation used to calculate the joint centres, centre of masses and linear accelerations can be found in Appendix 6. Table 3.7 shows the anatomical locations of the dynamic marker placements and figure 3.20 shows the dynamic marker placements of a subject during dynamic lifting trials for calculating joint centres and centres of mass.

Table 3.7 Anatomical location of dynamic marker placements

Marker No.	Position	Marker No.	Position
M1	L. Lateral Mallelous	M10	R. Distal End of Metetarsals V
M2	R. Lateral Mallelous	M11	L. Heel
M3	L. Medial Mallelous	M12	R. Heel
M4	R. Medial Mallelous	M13	Mid Point of PSIS
M7	L. Lateral Femoral Epicondyle	M14	L. PSIS (L. Dimple)
M8	R. Lateral Femoral Epicondyle	M15	R. PSIS (R. Dimple)
M9	L. Distal End of Metetarsals V		

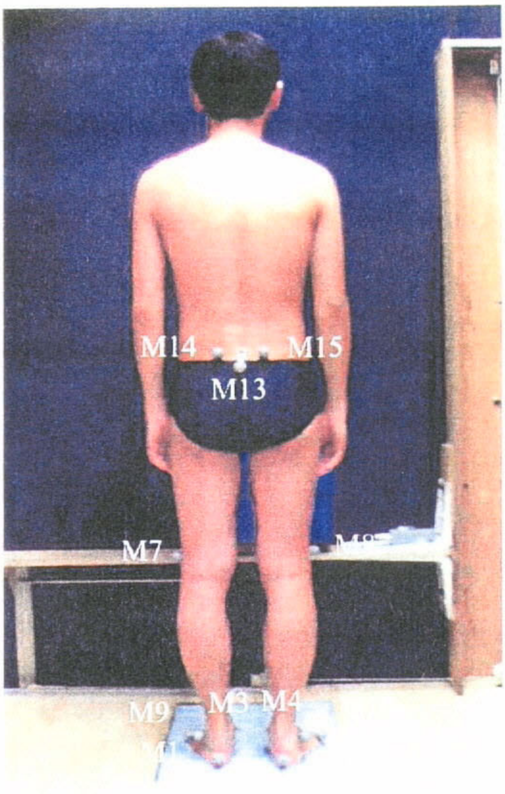


Figure 3.20 Location of dynamic marker placements

Angular acceleration of body segments

In order to calculate the angular acceleration of the centre of mass of different body segments about different axes, the orientation of all body segments needed to be found first. In turn, joint centres need to be located for determination of the orientation of body segments. Similarly, the locations of all joint centres (left and right ankle, knee and hip joints) and the orientation of all body segments (left and right feet, shanks, thighs and the pelvis) were found by adopting the uvw reference system and the equations developed by Vaughan et al. (1992).

According to Cappozzo (1983) and Vaughan et al. (1992), the local coordinate system embedded at the centre of mass of each body segment could be generated after computing the locations of the joint centres, and used to describe the spatial orientation of each body segment. Cardan angles also need to be determined for the calculation of the angular acceleration of all body segments. Cardan angles are three successive angles of rotation that transform a Cartesian coordinate system to another by an ordered sequence of rotations. In this study, Cardan angles were used to transform the local coordinate system to the coordinate system of the motion analysis system. The details can be found in Appendix 7. Finally, all angular accelerations of the centre of mass of all body segments about x, y and z axes were found by the 5-point method. All the equations used to calculate the spatial orientation and angular accelerations can also be found in Appendix 7, and the definitions of the abbreviations used in the equations can be found in Appendix 4.

Static trials

Due to the limited number of cameras of the motion analysis system, it was found that four of the dynamic markers that were placed on the anterior side of the subject's body could not be seen by three or more cameras during dynamic lifting trials. They were the left tibialis anterior (M5), right tibialis anterior (M6), left anterior superior iliac spine (M21) and right anterior superior iliac spine (M22). As these markers are required as input to the biomechanical model, two static trials were therefore performed for each subject in order to compute the relative positions of those dynamic markers which could not be seen during lifting trials.

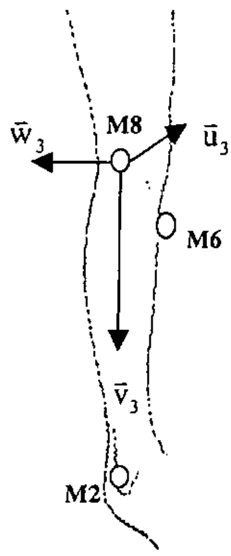
I. Static trial for the right shank

According to Vaughan et al., (1992), the right knee joint centre during dynamic lifting trials was found by using points M2 (right lateral mallelous), M6 (right tibialis anterior) and M8 (right lateral femoral epicondyle) to form a reference coordinate system together with equation 3.1 as follows:

$$\vec{v}_{3(t)} = \frac{\vec{P}_{M2(t)} - \vec{P}_{M8(t)}}{\left|\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right|}$$

$$\vec{w}_{3(t)} = \frac{\left(\vec{P}_{M6(t)} - \vec{P}_{M8(t)}\right)\times\left(\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right)}{\left|\left(\vec{P}_{M6(t)} - \vec{P}_{M8(t)}\right)\times\left(\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right)\right|}$$

$$\vec{u}_{3(t)} = \vec{v}_{3(t)} \times \vec{w}_{3(t)}$$



$$\vec{P}_{RK(t)} = \vec{P}_{M8(t)} + 0.423\,(A_{11})\,(\vec{u}_{3(t)}) - 0.198\,(A_{11})\,(\vec{v}_{3(t)}) + 0.406\,(A_{11})\,(\vec{w}_{3(t)})$$

Equation 3.1

where A_{11} is right knee diameter, $\vec{P}_{RK(t)}$ is the right knee joint centre and subscript 3 means body segment 3, right shank

A static trial was conducted since M6 could not be seen during dynamic lifting trials. Four markers, M2, M4 (right medial mallelous), M6 and M8 (Table 3.8 and figure 3.21) were attached to each subject during static trial. M2, M4 and M8 were used to form a local reference coordinate system and the spatial relationship ($\vec{P}_{C1}$) between M6 and the defined local reference coordinate system could be found by equation 3.2:

$$\left[\vec{P}_{C1}\right] = \left[\vec{R}_3\right]\left[\vec{P}_{M6} - \vec{P}_{M2}\right]$$

Equation 3.2

where

$$\left[\vec{R}_3\right] = \begin{bmatrix} \vec{u}_3 \\ \vec{v}_3 \\ \vec{w}_3 \end{bmatrix}$$

$$\vec{u}_3 = \frac{\vec{P}_{M8} - \vec{P}_{M2}}{\left|\vec{P}_{M8} - \vec{P}_{M2}\right|}$$

$$\vec{v}_3 = \frac{\left(\vec{P}_{M4} - \vec{P}_{M2}\right)\times\left(\vec{P}_{M8} - \vec{P}_{M2}\right)}{\left|\left(\vec{P}_{M4} - \vec{P}_{M2}\right)\times\left(\vec{P}_{M8} - \vec{P}_{M2}\right)\right|}$$

$$\vec{w}_3 = \vec{u}_3 \times \vec{v}_3$$

The relative position of M6 (right tibialis anterior) during dynamic lifting trials with respect to the local reference coordinate system defined could then be found using equation 3.3.

$$\left[\vec{R}_{3(t)}\right]^{-1}\left[\vec{P}_{C1}\right] = \left[\vec{P}_{M6(t)} - \vec{P}_{M2(t)}\right]$$
$$\vec{P}_{M6(t)} = \left[\vec{R}_{3(t)}\right]^{-1}\left[\vec{P}_{C1}\right] + \vec{P}_{M2(t)}$$

Equation 3.3

The spatial relationship ($\vec{P}_{C1}$) is fixed with the assumption that the defined body segment behaves as a rigid body. All the definitions of the abbreviations of the above equations can be found in Appendix 4.

Table 3.8 Anatomical location of static marker placements for shank

Marker No.	Position	Marker No.	Position
M1	L. Lateral Mallelous	M5	L. Tibial Tubercle
M2	R. Lateral Mallelous	M6	R. Tibial Tubercle
M3	L. Medial Mallelous	M7	L. Lateral Femoral Epicondyle
M4	R. Medial Mallelous	M8	R. Lateral Femoral Epicondyle

Figure 3.21 Location of static marker placements for shank

II. Static trial for the left shank

The relative position of M5 (left tibialis anterior) during dynamic lifting trials was found by using a similar calculation as that for M6 (right tibialis anterior) and can be found in Appendix 8.

III. Static trial for pelvis

A static trial was also conducted for the pelvis segment since M21 (left anterior superior iliac spine) and M22 (right anterior superior iliac spine) could not be seen during dynamic lifting tasks and were needed for the calculation of centre of mass and orientation of the pelvis. During the static trial, the five markers M13 (left posterior superior iliac spine), M14 (right posterior superior iliac spine), M15 (mid point of posterior superior iliac spine), M21, and M22 were attached to the subject (Table 3.9), and M13, M14 and M15 were used to form a local reference coordinate system $[\vec{R}_7]$ as follows:

$$[\vec{R}_7] = \begin{bmatrix} \vec{u}_7 \\ \vec{v}_7 \\ \vec{w}_7 \end{bmatrix}$$

Where subscript 7 means body segment 7, (the pelvis) and

$$\vec{w}_7 = \frac{\vec{P}_{M14} - \vec{P}_{M15}}{|\vec{P}_{M14} - \vec{P}_{M15}|}$$

$$\vec{u}_7 = \frac{(\vec{P}_{M14} - \vec{P}_{M15}) \times (\vec{P}_{M13} - \vec{P}_{M15})}{|(\vec{P}_{M14} - \vec{P}_{M15}) \times (\vec{P}_{M13} - \vec{P}_{M15})|}$$

$$\vec{v}_7 = \vec{w}_7 \times \vec{u}_7$$

Then, the spatial relationship, $\vec{P}_{C3}$, (between M21 and the defined local reference coordinate system) and $\vec{P}_{C4}$ (between M22 and the defined local reference coordinate system) could then be found using equations 3.4 and 3.5:

$$\left[\vec{P}_{C3}\right]=\left[\vec{R}_{7}\right]\left[\vec{P}_{M21}-\vec{P}_{M14}\right]$$

Equation 3.4

and

$$\left[\vec{P}_{C4}\right]=\left[\vec{R}_{7}\right]\left[\vec{P}_{M22}-\vec{P}_{M14}\right]$$

Equation 3.5

Finally, the relative positions of M21 and M22 during dynamic lifting trials with respect to the local reference coordinate system defined could be found using equations 3.6 and 3.7.

$$\left[\vec{R}_{7(t)}\right]^{-1}\left[\vec{P}_{C3}\right]=\left[\vec{P}_{M21(t)}-\vec{P}_{M14(t)}\right]$$
$$\vec{P}_{M21(t)}=\left[\vec{R}_{7(t)}\right]^{-1}\left[\vec{P}_{C3}\right]+\vec{P}_{M14(t)}$$

Equation 3.6

and

$$\left[\vec{R}_{7(t)}\right]^{-1}\left[\vec{P}_{C4}\right]=\left[\vec{P}_{M22(t)}-\vec{P}_{M14(t)}\right]$$
$$\vec{P}_{M22(t)}=\left[\vec{R}_{7(t)}\right]^{-1}\left[\vec{P}_{C4}\right]+\vec{P}_{M14(t)}$$

Equation 3.7

The spatial relationships ($\vec{P}_{C3}$ and $\vec{P}_{C4}$) are fixed with the assumption that the defined body segment behaves as a rigid body. All the definitions of the abbreviations used in the above equations can be found in appendix 4.

Table 3.9 Anatomical location of static marker placements for pelvis

Marker No.	Position	Marker No.	Position
M13	L. PSIS (L. Dimple)	M21	L. ASIS
M14	R. PSIS (R. Dimple)	M22	R. ASIS
M15	Mid point of PSIS		

IV Location of the lumbosacral joint

The location of L5/S1 was found by digitization of a skeleton (with body weight of 70kg and body height of 1.76m) since it couldn't be monitored directly during dynamic lifting trials. D1 and D2 (Figures 3.22 and 3.23) were the most prominent points of the medial and lateral side of the L5/S1 intervertebral disc, their relative spatial coordinates with respect to a defined origin were obtained by using a desktop 3-D digitising system (Immersion Corporation. U.S.A). The mid point of these two coordinates was defined to be the centroid of the L5/S1 intervertebral disc. Similar to the shank and pelvis, three makers, M13 (Mid point of PSIS), M14 (Left PSIS) and M15 (Right PSIS) were also attached to the skeleton and digitized to form a local reference coordinate system $[\vec{R}_{D7}]$ as follows:

$$[\vec{R}_{D7}] = \begin{bmatrix} \vec{u}_{D7} \\ \vec{v}_{D7} \\ \vec{w}_{D7} \end{bmatrix}$$

where

$$\vec{w}_{D7} = \frac{\vec{P}_{M14} - \vec{P}_{M15}}{|\vec{P}_{M14} - \vec{P}_{M15}|}$$
$$\vec{u}_{D7} = \frac{(\vec{P}_{M14} - \vec{P}_{M15}) \times (\vec{P}_{M13} - \vec{P}_{M15})}{|(\vec{P}_{M14} - \vec{P}_{M15}) \times (\vec{P}_{M13} - \vec{P}_{M15})|}$$
$$\vec{v}_{D7} = \vec{w}_{D7} \times \vec{u}_{D7}$$

The spatial relationship, $\vec{P}_{C5}$, between the disc centroid ($\vec{P}_{L5/S1}$) and the defined local reference coordinate system was then found by equation 3.8:

$$[\vec{P}_{C5}] = [\vec{R}_{D7}][\vec{P}_{L5/S1} - \vec{P}_{M14}]$$

Equation 3.8

Finally, the relative position of the lumbosacral joint during dynamic lifting trials with respect to the local reference coordinate system defined can be found using equation 3.9 and assuming that $[\vec{R}_{D7}] = [\vec{R}_7]$:

$$\left[\vec{R}_{7(t)}\right]^{-1}\left[\vec{P}_{C5}\right]=\left[\vec{P}_{L5/S1(t)}-\vec{P}_{M14(t)}\right]$$
$$\vec{P}_{L5/S1(t)}=\left[\vec{R}_{7(t)}\right]^{-1}\left[\vec{P}_{C5}\right]+\vec{P}_{M14(t)}$$

Equation 3.9

The spatial relationship ($\vec{P}_{C5}$) is fixed with the assumption that the defined body segment behaves as a rigid body. All the definitions of the abbreviations used in the above equations can be found in appendix 4.

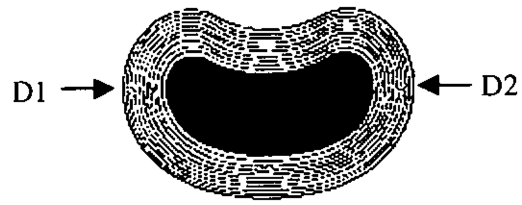


Figure 3.22 Top view of the location of the digitization points of the L5/S1 intervertebral disc (El-Bohy, A.A.R., 1988)

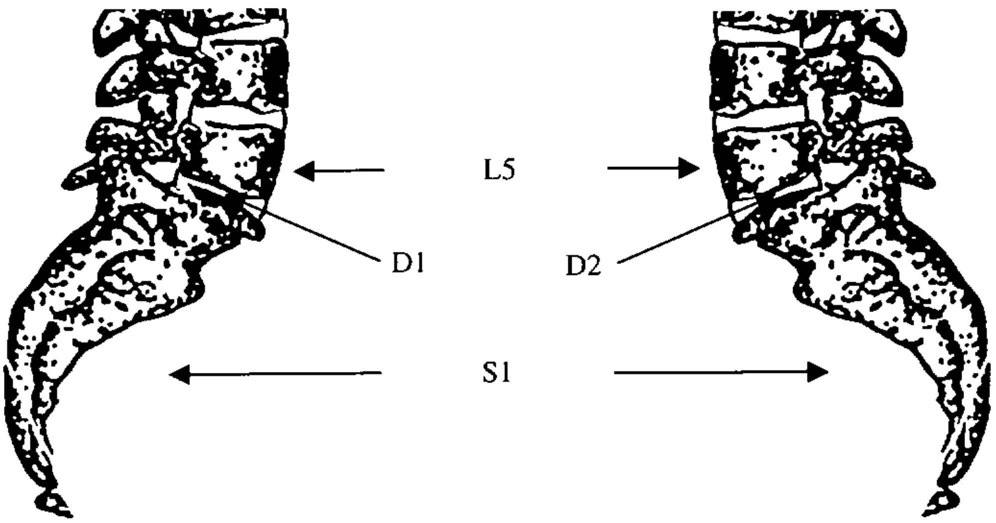


Figure 3.23 Medial and lateral view of the location of the digitization points of the L5/S1 intervertebral disc (El-Bohy, A.A.R., 1988)

3.6.3 Kinetic data

Kinetic analyses are based on kinematics and incorporate the effects of forces and moments that cause the motion. Kinetic data includes the ground reaction force, centre of pressure and axial torque. The ground reaction forces were obtained directly from the force plate and the centre of pressure and axial torque were calculated according to the following equations provided by the manufacturer:

Table 3.10 Equations for calculating centre of pressure and axial torque

Parameter	Calculation	Description
Fx	=fx12+fx34	Medio-lateral force
Fy	=fy14+fy23	Antero-posterior force
Fz	=fz1+fz2+fz3+fz4	Vertical force
Mx	=b*(fz1+fz2-fz3-fz4)	Plate moment about X-axis
My	=a*(-fz1+fz2+fz3-fz4)	Plate moment about Y-axis
Mx'	=b*(fz1+fz2-fz3-fz4)+Fy*az0	Plate moment about top plate surface
My'	=a*(-fz1+fz2+fz3-fz4)+Fx*az0	Plate moment about top plate surface
Mz	=b*(-fx12+fx34)+a*(fy14-fy23)	Plate moment about z-axis
Tz	Mz-Fy*ax+Fx*ay	Free moment, Vertical torque
ax	=(Fx*az0-My)/Fz	X-Coordinate of force application point (COP)
ay	=(Fy*az0+Mx)/Fz	Y-Coordinate of force application point (COP)

3.6.4 Data processing

The kinetic and kinematic data were filtered at an effective cut-off frequency of 40Hz and 5Hz respectively, using a low pass fourth-order Butterworth filter with zero-phase lag. The cut-off frequencies were determined using frequency spectrum analysis, a method to analyse the characteristics of a signal. The main frequency band of the signals was found by a mathematical process, Fast Fourier Transformation (FFT) that transforms signals from time-domain to frequency-domain. With the sampling frequency of 3000Hz for kinetic data and 120Hz for kinematic data, 1024 number of samples and Hamming window for both kinematic and kinetic data, the cut-off frequency for the kinetic data was found to be 50Hz while the cut-off frequency for the kinematic data was found to be 5Hz, which was also used by De Looze et al. (2000).

With input of anthropometric, kinematic and kinetic data, equations 3.10 & 3.11 were applied to the feet, lower legs, upper legs and pelvis at each time instant to obtain the torque at the L5/S1 joint using a computer program written in Visual Basic. Axis directions have been defined such that when the subject is standing in an upright, erect posture, the positive X-direction is cranial; the Y-direction is anterior and the Z-direction is left lateral.

Since the summation of all forces acting on the lumbosacral joint should be equal to zero, then the net force at the lumbosacral joint at a time t = sum of gravitational force of all segments + sum of inertia force of all segments – ground reaction force as equation 3.10

$$\Sigma\bar{F} = 0$$

$$\bar{F}_{L5/S1(t)} = \sum_{i=1}^{7} m_i\bar{g} + \sum_{i=1}^{7} m_i\bar{a}_{i(t)} - \bar{F}_{r(t)}$$

$$\begin{pmatrix} F_{L5/S1x(t)} \\ F_{L5/S1y(t)} \\ F_{L5/S1z(t)} \end{pmatrix} = \sum_{i=1}^{7} m_i \begin{pmatrix} 0 \\ 0 \\ g \end{pmatrix} + \sum_{i=1}^{7} m_i \begin{pmatrix} a_{ix(t)} \\ a_{iy(t)} \\ a_{iz(t)} \end{pmatrix} - \begin{pmatrix} F_{rx(t)} \\ F_{ry(t)} \\ F_{rz(t)} \end{pmatrix}$$

Equation 3.10

Where

$\bar{F}_{L5/S1(t)}$ = Resultant force at the L5/S1 at time t with upper part on lower part

$\sum_{i=1}^{7} m_i\bar{g}$ = Summation of gravitational force of all segments

$\sum_{i=1}^{7} m_i\bar{a}_{i(t)}$ = Summation of inertia force of all segments

$\bar{F}_{r(t)}$ = Ground reaction force

Similarly, the sum of all moments acting on the lumbosacral joint should also be equal to zero. The net moment at the lumbosacral joint at time t is equal to the sum of moments due to gravitational force with moment arm from the centre of mass of each segment to the lumbosacral joint + sum of moments due to linear inertia force with moment arm from the centre of mass of each segment to the lumbosacral joint + summation of moment due to angular inertia force of all segments – axial torque due to ground reaction force – moment due to ground reaction force with moment arm from the centre of the force plate to the lumbosacral joint, as given by equation 3.11:

$$\begin{aligned}
\sum \vec{M} &= 0 \\
\vec{M}_{L5/S1(t)} &= \sum_{i=1}^{7}\left[\left(\vec{P}_{i(t)} - \vec{P}_{L5/S1(t)}\right)\times m_i\vec{g}\right] + \sum_{i=1}^{7}\left[\left(\vec{P}_{i(t)} - \vec{P}_{L5/S1(t)}\right)\times m_i\vec{a}_{i(t)}\right] \\
&\quad + \sum_{i=1}^{7}\left(I_i\vec{\alpha}_{i(t)}\right) - \vec{T}_{(t)} - \left(\vec{P}_{r(t)} - \vec{P}_{L5/S1(t)}\right)\times \vec{F}_{r(t)}
\end{aligned}$$

$$\begin{aligned}
\begin{pmatrix} M_{L5/S1x(t)} \\ M_{L5/S1y(t)} \\ M_{L5/S1z(t)} \end{pmatrix} &= \sum_{i=1}^{7}\left\{\begin{pmatrix} x_{i(t)} - x_{L5/S1(t)} \\ y_{i(t)} - y_{L5/S1(t)} \\ z_{i(t)} - z_{L5/S1(t)} \end{pmatrix} \times m_i \begin{pmatrix} 0 \\ 0 \\ g \end{pmatrix}\right\} + \sum_{i=1}^{7}\left\{\begin{pmatrix} x_{i(t)} - x_{L5/S1(t)} \\ y_{i(t)} - y_{L5/S1(t)} \\ z_{i(t)} - z_{L5/S1(t)} \end{pmatrix} \times m_i \begin{pmatrix} a_{ix(t)} \\ a_{iy(t)} \\ a_{iz(t)} \end{pmatrix}\right\} \\
&\quad + \sum_{i=1}^{7}\begin{pmatrix} I_{ix}\,\alpha_{ix(t)} \\ I_{iy}\,\alpha_{iy(t)} \\ I_{iz}\,\alpha_{iz(t)} \end{pmatrix} - \begin{pmatrix} 0 \\ 0 \\ T_{z(t)} \end{pmatrix} - \begin{pmatrix} x_{r(t)} - x_{L5/S1(t)} \\ y_{r(t)} - y_{L5/S1(t)} \\ z_{r(t)} - z_{L5/S1(t)} \end{pmatrix} \times \begin{pmatrix} F_{rx(t)} \\ F_{ry(t)} \\ F_{rz(t)} \end{pmatrix}
\end{aligned}$$

Equation 3.11

where

$\vec{M}_{L5/S1(t)}$ = Resultant moment at L5/S1

$\sum_{i=1}^{7}\left[\left(\vec{P}_{i(t)} - \vec{P}_{L5/S1(t)}\right)\times m_i\vec{g}\right]$ = Summation of moments due to gravitational force

$\sum_{i=1}^{7}\left[\left(\vec{P}_{i(t)} - \vec{P}_{L5/S1(t)}\right)\times m_i\vec{a}_{i(t)}\right]$ = Summation of moments due to linear inertia force

$\sum_{i=1}^{7}\left(I_i\vec{\alpha}_{i(t)}\right)$ = Summation of moments due to angular inertia force of all segments

$\vec{T}_{(t)}$ = Axial torque due to ground reaction force

$\left(\vec{P}_{r(t)} - \vec{P}_{L5/S1(t)}\right)\times \vec{F}_{r(t)}$ = Moment due to ground reaction force

3.6.5 Data analysis

The lifting event and moment at L5/S1 can be divided into three phases. The first phase occurs as the subject bends or kneels down to grasp the box; the resultant L5/S1 moment is due to the subject's upper body. Secondly, as the subject lifts the box and returns to an upright position, the L5/S1 moment is a result of the body mass of the subject and the weight lifted, which rises and reaches a peak value as the pulling force on the box increases. Thirdly, as the subject encounters the sudden release of load and returns to a balanced position, the L5/S1 moment shows a steep drop and fluctuates for a period of time after sudden load release as the subject tries to preserve balance while still holding the fake load. The net moment calculated at L5/S1 during the sudden release cycle and the complete lifting cycle immediately prior to sudden release is shown (Figure 3.24).

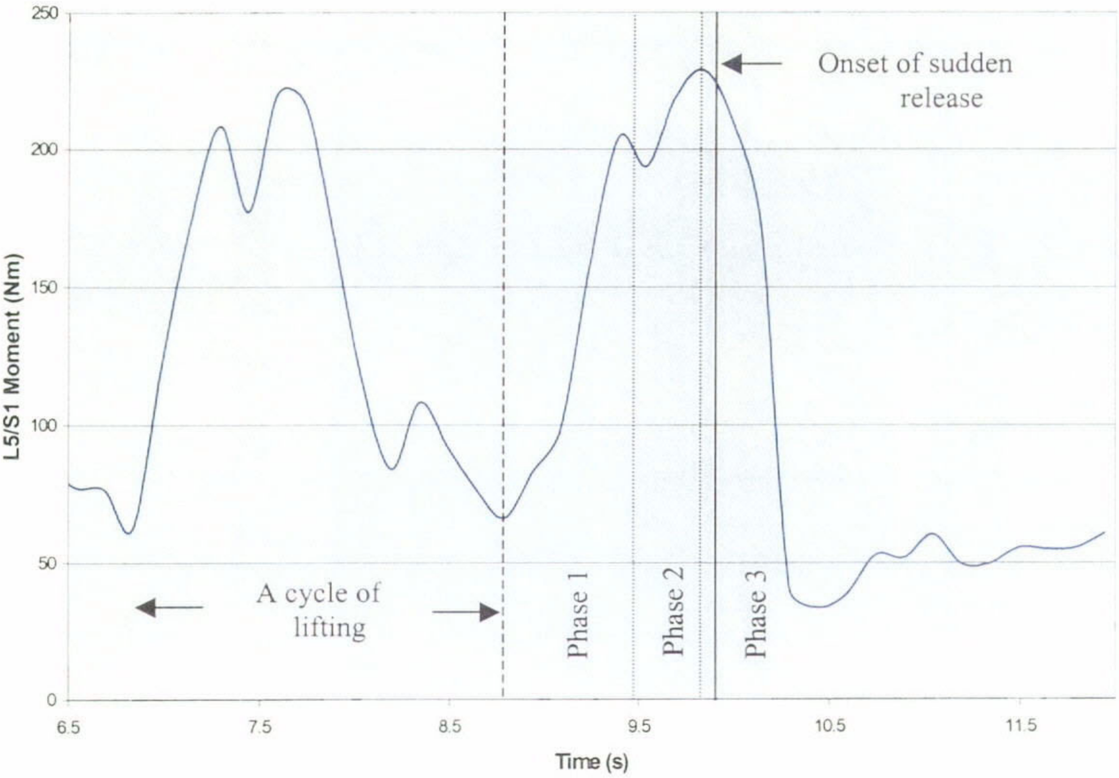


Figure 3.24 A typical profile of net moment at L5/S1

The L5/S1 torque was projected onto the anatomical axis systems yielding the component of interest, namely the trunk flexing-extension torque, lateral bending

torque and axial rotation torque at L5/S1. In order to investigate the resultant moment at L5/S1 under different sudden release of load conditions systematically, the following terms were defined
(Figure 3.25-27):

- **Peak axial rotation moment during lifting (M_{PAR})** : The peak axial rotation moment at L5/S1 during the sudden release cycle (Figure 3.25).
- **Peak lateral bending moment during lifting (M_{PLB})** : The peak lateral bending moment at L5/S1 during the sudden release cycle (Figure 3.26).
- **Peak flexion-extension moment during lifting (M_{PFE})** : The peak flexion-extension moment at L5/S1 during the sudden release cycle (Figure 3.27).
- **Onset of sudden unload (T_{OS})** : The onset time of sudden release of load registered by accelerometer.
- **Axial rotation moment at onset of sudden unload (M_{AR})** : The axial rotation moment at L5/S1 at onset of release of load during the sudden release cycle (Figure 3.25).
- **Trough of axial rotation moment (M_{TAR})** : The first trough of the axial rotation moment at L5/S1 after sudden release of load (Figure 3.25).
- **Time of the trough of axial rotation moment (T_{AR})** : The time of the first trough of the axial rotation moment at L5/S1 after sudden release of load (Figure 3.25).
- **Lateral bending moment at onset of sudden unload (M_{LB})** : The lateral bending moment at L5/S1 at onset of release of load during the sudden release cycle (Figure 3.26).
- **Trough of lateral bending moment (M_{TLB})** : The first trough of the lateral bending moment at L5/S1 after sudden release of load (Figure 3.26).
- **Time of the trough of lateral bending moment (T_{LB})** : The time of the first trough of the axial lateral bending moment at L5/S1 after sudden release of load (Figure 3.26).

- **Flexion-extension moment at onset sudden unload (M_{FE})** : The flexion-extension moment at L5/S1 during the sudden release cycle (Figure 3.27).
- **Trough of flexion-extension moment (M_{TFE})** : The first trough of the flexion-extension moment at L5/S1 after sudden release of load (Figure 3.27).
- **Time of the trough of flexion-extension moment (T_{AR})** : The time of the first trough of the flexion-extension moment at L5/S1 after sudden release of load (Figure 3.27).

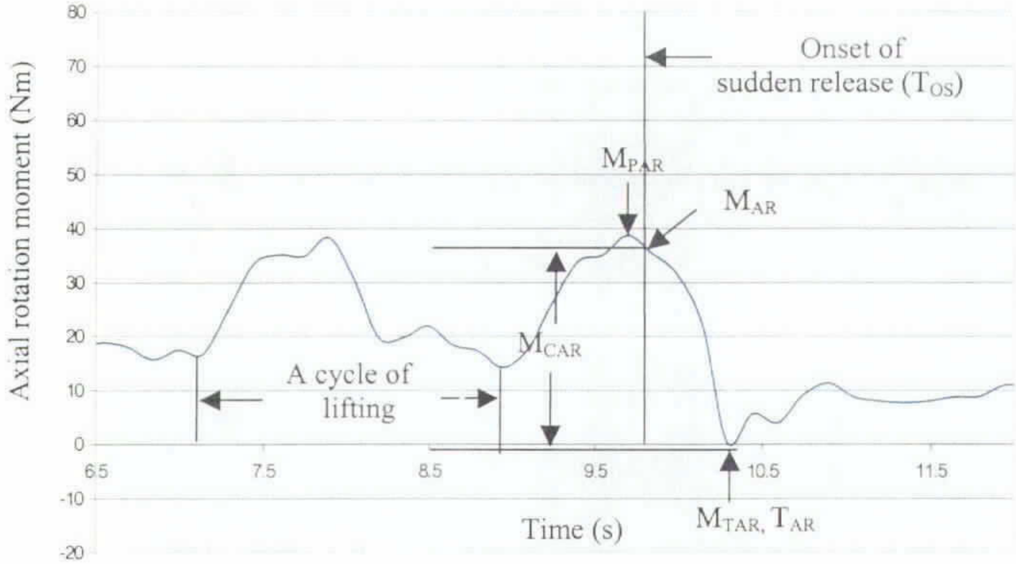


Figure 3.25 A typical profile of axial moment during lifting and encounter sudden release of load (with positive rotate to the left and negative rotate to the right)

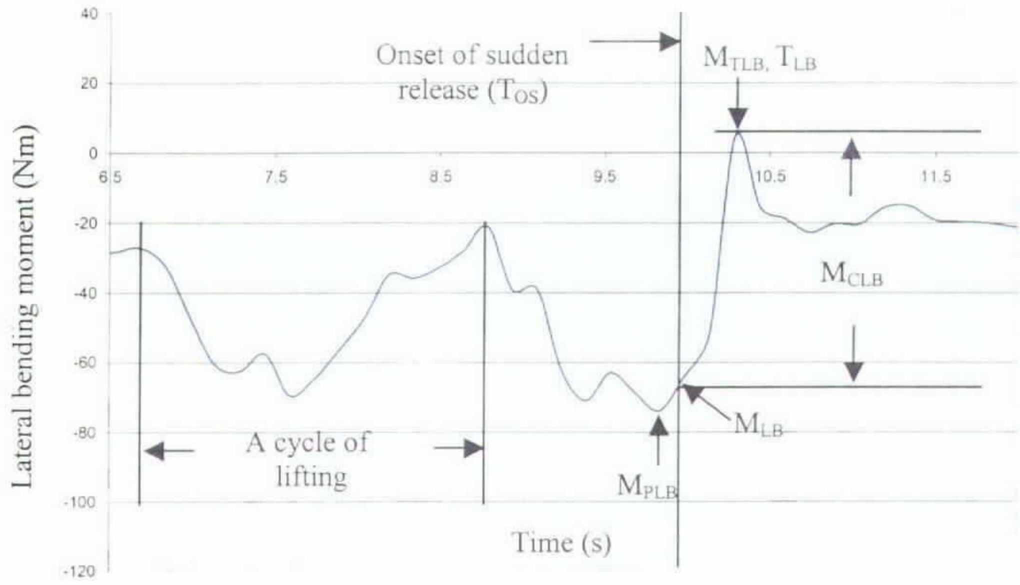


Figure 3.26 A typical profile of lateral bending moment during lifting and encounter sudden release of load (with positive bend to the right and negative bend to the left)

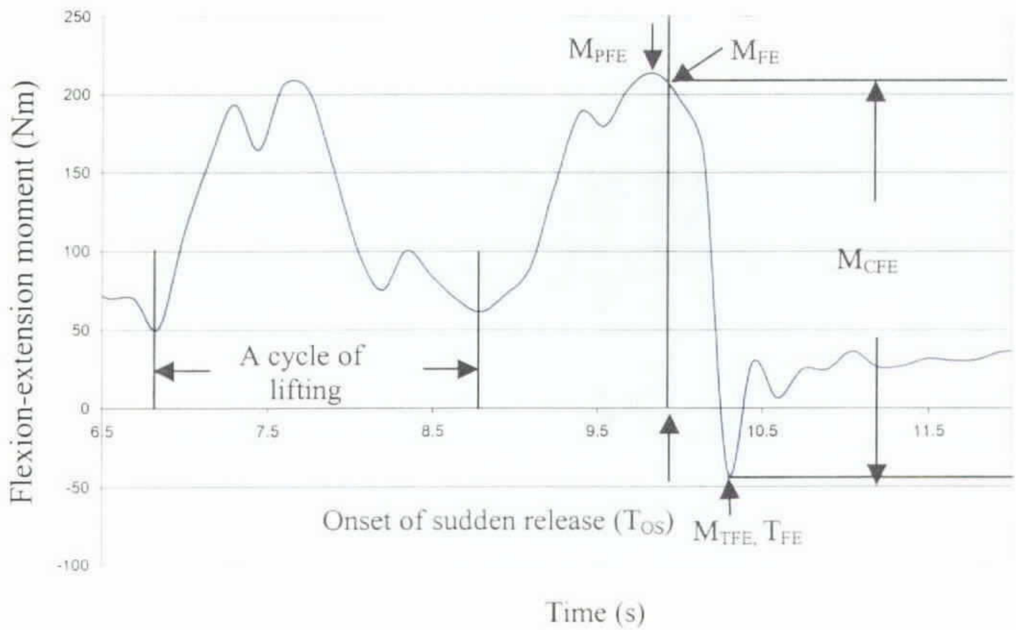


Figure 3.27 A typical profile of flexion-extension moment during lifting and encounter sudden release of load (with positive is flexion and negative is extension)

Figures 3.25-27 show that sudden release of load causes an "overshoot" in L5/S1 moment. When sudden release of load happened unexpectedly during a lifting task, the flexion moment ended suddenly, accelerating the trunk backwards, and unbalancing the body. Following the sudden release of load, moments reached their troughs in a short period of time. Thus, in order to examine the effect of rate of change moment after sudden release of load on balance preservation, the rate of change of moment from onset of sudden release of load to the trough was determined for each lifting condition as follows:

Absolute change of axial rotation moment after sudden release of load (M_{CAR}):

$= M_{TAR} - M_{AR}$

Absolute rate of change of axial rotation moment after sudden release of load (R_{AR}):

$= M_{CAR} / (T_{AR} - T_{OS})$

Absolute change of lateral bending moment after sudden release of load (M_{CLB}):

$= M_{TLB} - M_{LB}$

Absolute rate of change of lateral bending moment after sudden release of load (R_{LB}):

$= M_{CLB} / (T_{LB} - T_{OS})$

Absolute change of flexion-extension moment after sudden release of load (M_{CFE}):

$= M_{TFE} - M_{FE}$

Absolute rate of change of flexion-extension moment after sudden release of load (R_{FE}):

$= M_{CFE} / (T_{FE} - T_{OS})$

3.7 Experimental Procedures

Ten normal and healthy male volunteers were recruited in this study. None of the subjects had any history of back injuries or significant back pain for the last two years. Before the experimental testing, the purpose and the procedures of the experiment were explained to each subject clearly by the investigator. All subjects understood and agreed to the experimental protocol and gave written and informed consent.

3.7.1 Subject preparation

Each subject was asked to wear a swimsuit and be bare footed for the whole experiment. Anthropometric data including body weight, body height, knee height, lifting distance, shank and thigh circumferences, knee and malleolus diameters were collected after the subject warmed up and stretched-out briefly. Nine pairs of different muscles (bilateral latissimus dorsi, lumbar erector spinae, external oblique, internal oblique, rectus abdominis, biceps femoris, tibialis anterior, rectus femoris and gastrocnemius) and a prominent bony landmark as a ground were located via palpation and the use of bony landmarks. The areas were then scrubbed with alcohol and abraded to reduce the resistance of skin to less than 10k Ω, as measured with digital voltmeter. Each subject was asked to perform some specific movements in order to functionally test if all the EMG electrodes were correctly located. A total of 8 markers were attached to each subject for static trial capture and a total of 13 markers were attached to each subject for dynamic trial capture after removal of the static markers.

3.7.2 Dynamic lifting tasks

Following the measurement of anthropometric data and preparation for EMG data collection, each subject was then connected to the EMG systems and readied for performing the dynamic lifting tasks. The fake load was fixed at the subject's knee height for all trials and the lifting distance was defined as the total vertical distance from the subject's knee height to the height at final lifting position. Final lifting position was defined as the subject's elbow height. The load releasing height was set at one third of the total lifting height. Before the dynamic lifting tests started, both stoop and squat lifting postures were demonstrated by the investigator. The subject was then asked to stand on the force platform with feet separate at shoulder width and to try lifting at different standing positions (closer to or farther from the fake load) with stoop and squat lifting postures until they found the most comfortable position. The standing positions were then marked and the subject was required to stand at the same specific position when performing lifting tasks with specific lifting posture. Each subject was then instructed to perform lifting tasks with different lifting weights (20, 40, 60 and 80N), symmetry (symmetric and asymmetric with 30° to the left from the mid-sagittal plane) and lifting postures (squat lift and stoop lift) from his knee height according to a randomly assigned sequence. There were a total of 12 different lifting conditions for each subject and three trials for each lifting condition. The subject was asked to perform all lifting trials at same speed by following a metronome. The time for cutting off the power supply to the electromagnet was controlled by the investigator according to a randomly assigned sequence. The load was set to be released at either the 3^{rd}, 4^{th}, 5^{th} of the five lifting cycles or no release at all. An extra trial was conducted when no release was chosen for that trial. Each subject was allowed to rest for about 2 minutes between successive trials to prevent possible muscle fatigue. Figures 3.28, 3.29 and 3.30 show a subject performing lifting tasks with symmetric stoop and squat lifting posture and asymmetric stoop lifting posture respectively.

Figure 3.28 A subject performing symmetric squat lift

Figure 3.29 A subject performing symmetric stoop lift

Figure 3.30 A subject performing asymmetric stoop lift

CHAPTER 4 RESULTS

Three lifting postures with four lifting weights were investigated in this study resulting a total of 12 different lifting conditions for each subject. Subjects stood on a force plate and were instructed to perform a series of lifting tasks from knee height according to a randomly assigned sequence. The load was set to be released on either the 3^{rd}, 4^{th}, 5^{th} of the five lifting cycles at the lower one-third of the lifting distance or no release at all. An extra trial was conducted when no release was chosen for that trial. The centre of pressure, electromyographic activities of nine pairs of muscles and the loading on the L5/S1 joint in response to sudden release of load were studied. Data from ten normal and healthy male volunteers with mean age 26.5 ± 3.8, body height 172.3 ± 5.3cm, body weight 588.6 ± 85.6N and knee height 49 ± 1.3cm were analysed (Table 4.1).

Table 4.1 Subject anthropometry

Subject No.	Age (Years)	Body Height (cm)	Body Weight (N)	Knee Height (cm)
1	27	168	540	48
2	26	167	569	48
3	29	175	698	49
4	26	174	626	48
5	28	183	734	52
6	35	175	624	48
7	24	176	581	49
8	26	171	567	47
9	22	167	457	48
10	22	167	491	49
Mean	26.5	172.3	588.6	49
S.D.	3.8	5.3	85.6	1.3

All data were analysed using repeated measures ANOVA. If there was no interaction between within-subject factors, then contrast tests would be used to compare the differences between levels within the within-subject factor, which had significant effect. However, if there were significant interactions between within-subject factors, then the data would be analysed by splitting the data set into several subsets and analysed separately and contrast tests would also be used to compare the differences between levels within the within-subject factor, which had significant effect.

4.1 Centre Of Pressure Measurement

For the centre of pressure (COP) measurement, the following parameters were determined for each trial:

- The antero-posterior COP response time (Tyr)
- The medio-lateral COP response time (Txr)
- The antero-posterior COP peak time (Typ)
- The medio-lateral COP peak time (Txp)
- The posterior limit of COP displacement (L)

Definition of the above parameters can be found in section 3.5.1. It was found that all the lifting postures had similar COP displacement patterns, but different magnitudes for the defined parameters were found under different lifting conditions. All of the subjects were able to maintain equilibrium and no compensatory step was found during sudden release of load under all of the lifting conditions.

4.1.1 Response time

Sudden release of load causes an obvious COP deflection in both antero-posterior and medio-lateral directions. The response times in antero-posterior and medio-lateral directions were determined (Table 4.2 and Figures 4.1-4.2) and found to occur between 70 ms and 124 ms, and 84 ms and 251 ms respectively. Repeated measures ANOVA with two within-subject factors, lifting posture and lifting weight were performed. Factor one consisted of three levels denoting three different lifting postures namely symmetric squat, symmetric stoop and asymmetric stoop. Factor two consisted of four levels denoting four different lifting weights, which were 20, 40, 60 and 80N. It was shown that there was no interaction between within-subject factors for both antero-posterior (Tyr) and medio-lateral (Txr) COP response time with $p=0.134$ and $p=0.858$ respectively. The main effect of factor one (posture) had no statistically significant effect on antero-posterior COP response times ($p=0.36$) but had a significant effect on medio-lateral COP response time (p=0.038). Contrast tests showed that both symmetric squat lift and symmetric stoop lift are significantly different from asymmetric stoop lift (p=0.041 and p=0.03, respectively). However, symmetric squat lift is not significantly different from symmetric stoop lift (p=0.75). The pattern of medio-lateral COP response time is very similar for all lifting postures, but symmetric squat and stoop lift have closer values than the asymmetric stoop lift. The main effect of factor two, weight, had significant effect on both antero-posterior and medio-lateral COP response time (p=0.001 and p=0.008, respectively). Table 4.3 shows the result of contrast tests comparing the four levels within weight factor. It was shown that there were significant differences in the antero-posterior COP response time between the 20N and 60N, 20N and 80N, 40N and 60N, and, 40N and 80N loads. For medio-lateral COP response time, significant differences were found between 20N and 40N, 20N and 60N as well as 20N and 80N loads.

Table 4.2 Mean and standard deviation of COP response times in antero-posterior and medio-lateral directions with different lifting conditions after sudden release

Lifting conditions	Antero-posterior COP response time (Tyr)		Medio-lateral COP response time (Txr)	
	Mean (ms)	S.D. (ms)	Mean (ms)	S.D. (ms)
Sq20N	96.0	32.5	173.4	47.4
Sq40N	87.7	26.1	224.4	56.6
Sq60N	123.8	31.7	250.5	63.7
Sq80N	101.4	16.4	188.3	46.2
St20N	70.3	34.5	177.1	57.9
St40N	74.2	29.1	215.2	56.4
St60N	115.2	27.6	240.9	51.3
St80N	115.7	20.0	153.3	30.8
A20N	71.7	37.4	119.7	41.8
A40N	85.7	19.1	146.3	34.1
A60N	104.6	40.3	153.6	52.9
A80N	115.1	17.0	83.6	37.1

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

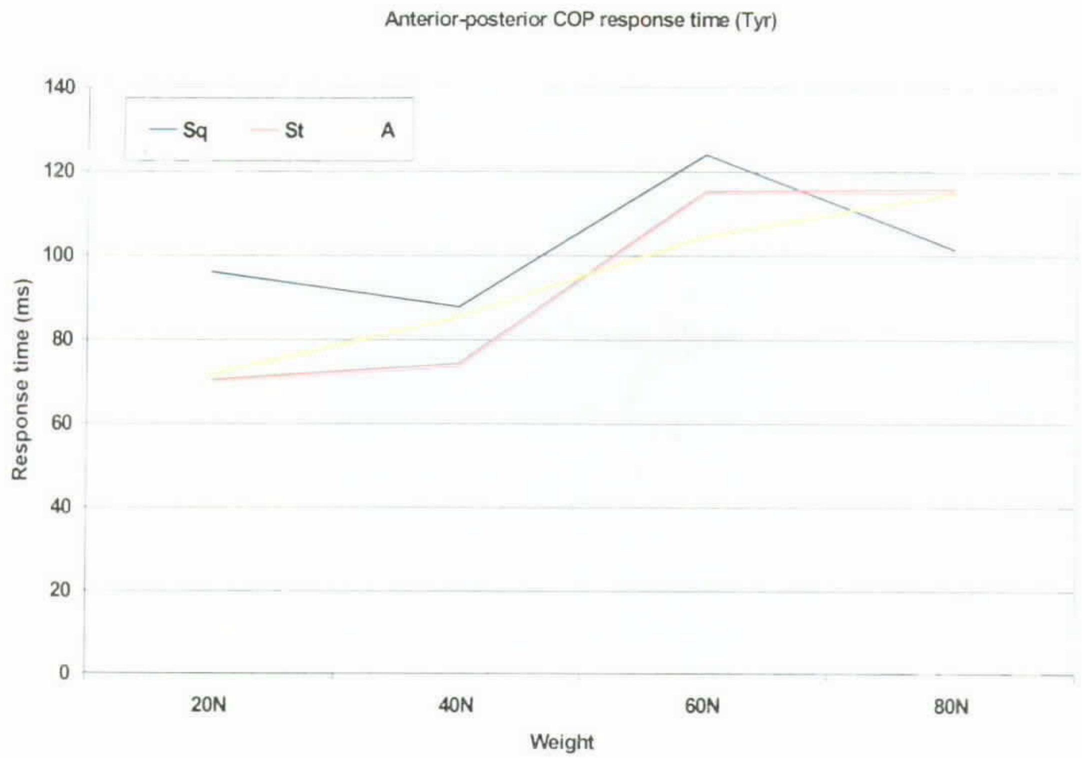


Figure 4.1 Antero-posterior COP response time (Tyr) for all lifting postures
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

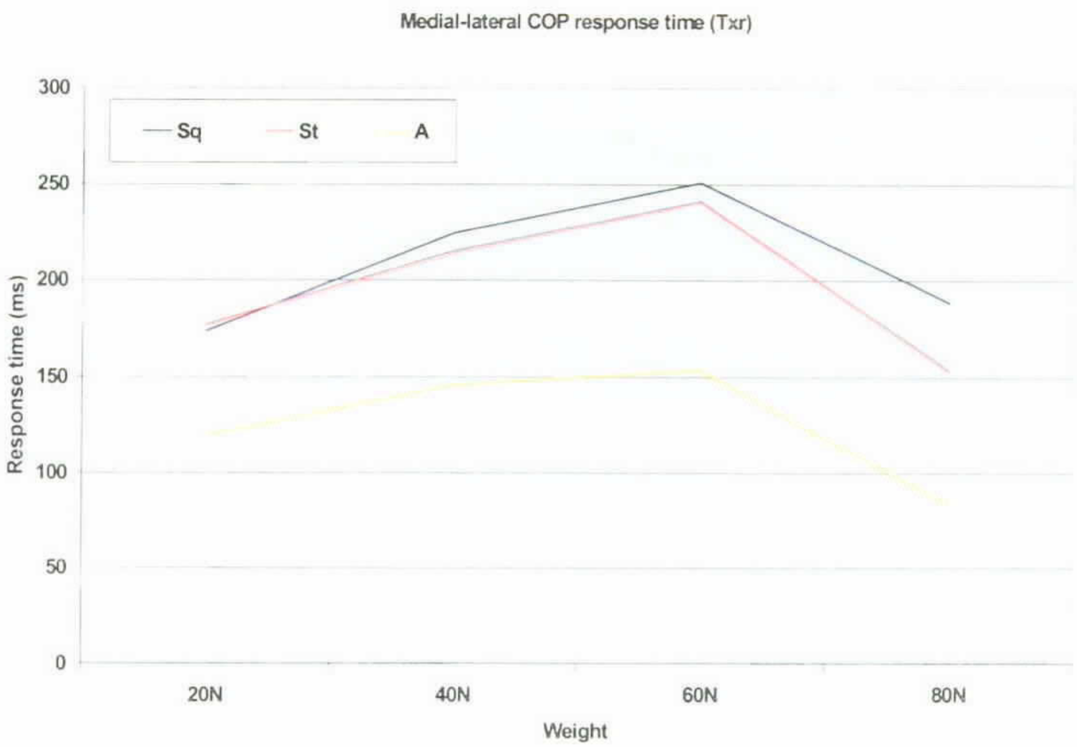


Figure 4.2 Medio-lateral COP response time (Txr) for all lifting postures
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

Since factor posture has no significant effect on the antero-posterior COP response time (Txr) with p=0.36, Txr was plotted against lifting weights as follow:

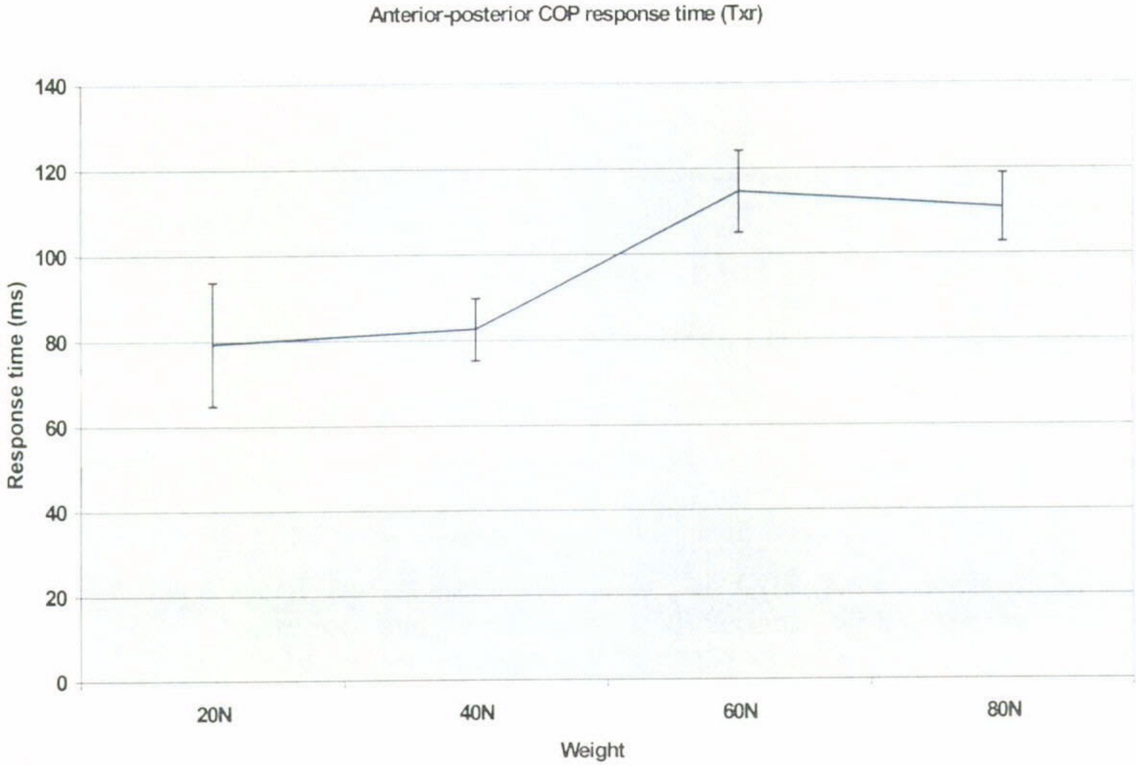


Figure 4.3 Antero-posterior COP response time (Txr)

The COP response time in both antero-posterior and medio-lateral directions increased as the lifting weight increased from 20N to 60N, but, decreased as the lifting weight increase from 60N to 80N.

Table 4.3 Contrast tests comparing the antero-posterior (Tyr) and medio-lateral (Txr) COP response times among the four levels of weight

Tyr	20N	40N	60N	80N	Txr	20N	40N	60N	80N
20N					20N				
40N	0.950				40N	0.042*			
60N	0.016*	0.006*			60N	0.037*	0.132		
80N	0.017*	0.007*	0.554		80N	0.001*	0.155	0.778	
* for p < 0.05									

4.1.2 Peak time

The peak COP deflections in antero-posterior and medio-lateral directions occurred between 240 ms and 381 ms, and 270 ms and 416 ms, respectively (Table

4.4 and Figures 4.4-4.5). Repeated measures ANOVA showed that there was an interaction (p=0.001) between the two within-subject factors (lifting posture and weight) for antero-posterior peak time (Typ) and therefore the data set was divided into four lifting weight groups for analysis. Posture was found to have no significant effect with p=0.117, p=0.26 and p=0.421 for lifting weights of 40N, 60N and 80N respectively. For lifting weight of 20N, posture was found to be significantly different with p=0.001 and contrast tests were used to compare the three levels of posture. Significant differences were found between symmetric squat lift and asymmetric stoop lift, symmetric stoop lift and asymmetric stoop lift as well as symmetric squat lift and symmetric stoop lift with p=0.022, p=0.032 and p=0.001, respectively.

Table 4.4 Mean and standard deviation of COP peak times in antero-posterior and medio-lateral directions with different lifting conditions after sudden release

Lifting conditions	Antero-posterior COP peak time (Typ)		Medio-lateral COP peak time (Txp)	
	Mean (ms)	S.D. (ms)	Mean (ms)	S.D. (ms)
Sq20N	380.7	103.7	411.5	130.9
Sq40N	284.3	80.5	270.0	65.3
Sq60N	287.3	62.3	295.5	60.9
Sq80N	294.7	49.3	316.4	93.4
St20N	257.5	51.1	358.6	87.9
St40N	239.6	47.5	354.7	184.2
St60N	294.0	69.8	357.3	156.0
St80N	289.4	62.6	277.8	129.0
A20N	301.9	79.4	395.2	88.0
A40N	279.6	62.3	398.1	121.0
A60N	333.7	111.4	346.1	61.9
A80N	311.5	106.8	317.7	80.6

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

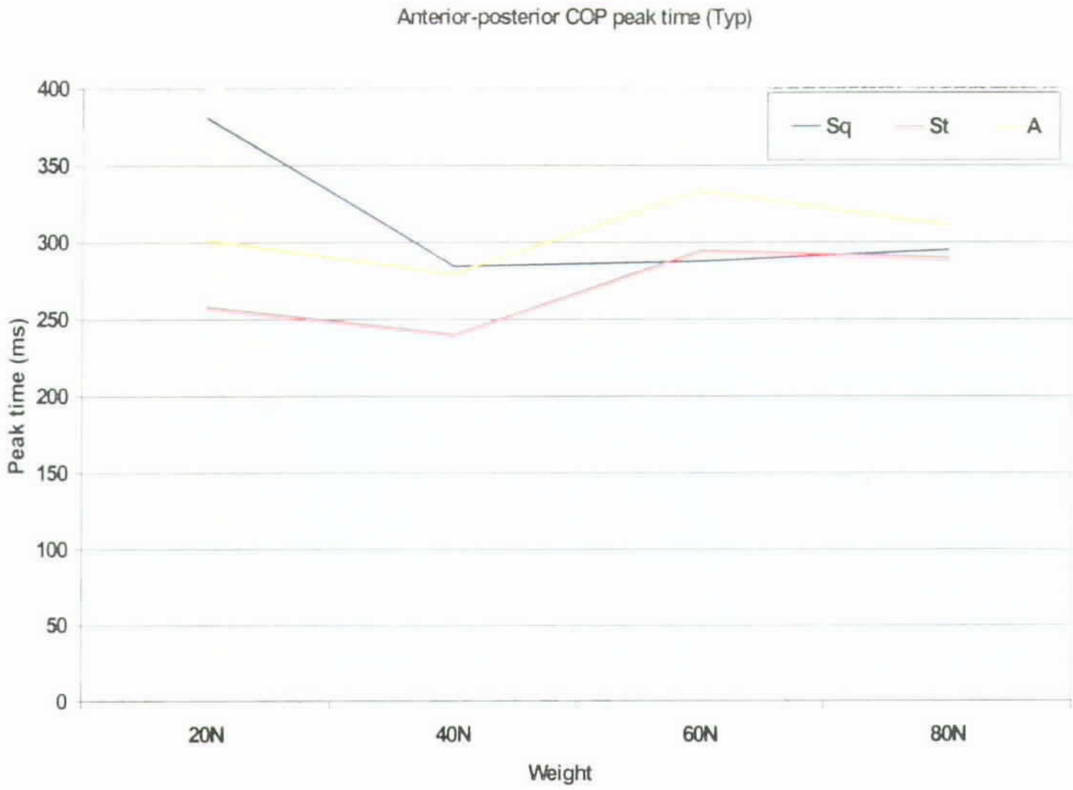


Figure 4.4 Antero-posterior COP peak time (Typ) for all lifting postures
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

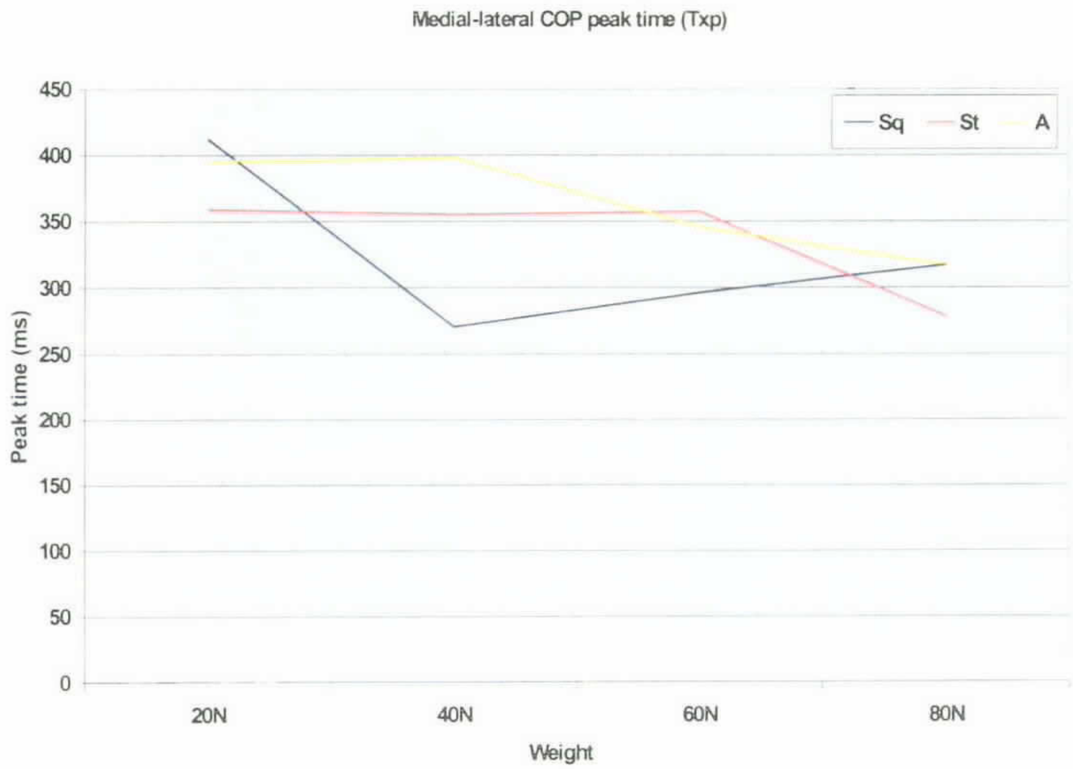


Figure 4.5 Medio-lateral COP peak time (Txp) for all lifting postures
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

For the medio-lateral COP peak times, repeated measures ANOVA showed there was no significant interaction between within-subject factors (posture and weight) with p=0.203 and the main effect of posture was not significant with p=0.345. However, weight was found to have significant effect on the medio-lateral COP peak times with p=0.026. Contrast tests were used to compare the four levels of weight. It was found that there were significant differences in the medio-lateral COP peak times between 20N and 60N, 20N and 80N, 40N and 60N and 40N and 80N (Table 4.5).

Table 4.5 Contrast tests comparing the medio-lateral COP peak times among the four levels of weight

Txp	20N	40N	60N	80N
20N				
40N	0.950			
60N	0.016*	0.006*		
80N	0.017*	0.007*	0.554	
* for p < 0.05				

Since the factor of posture has no significant effect on the medio-lateral COP peak time (Txp), Txp was plotted against lifting weights as follows:

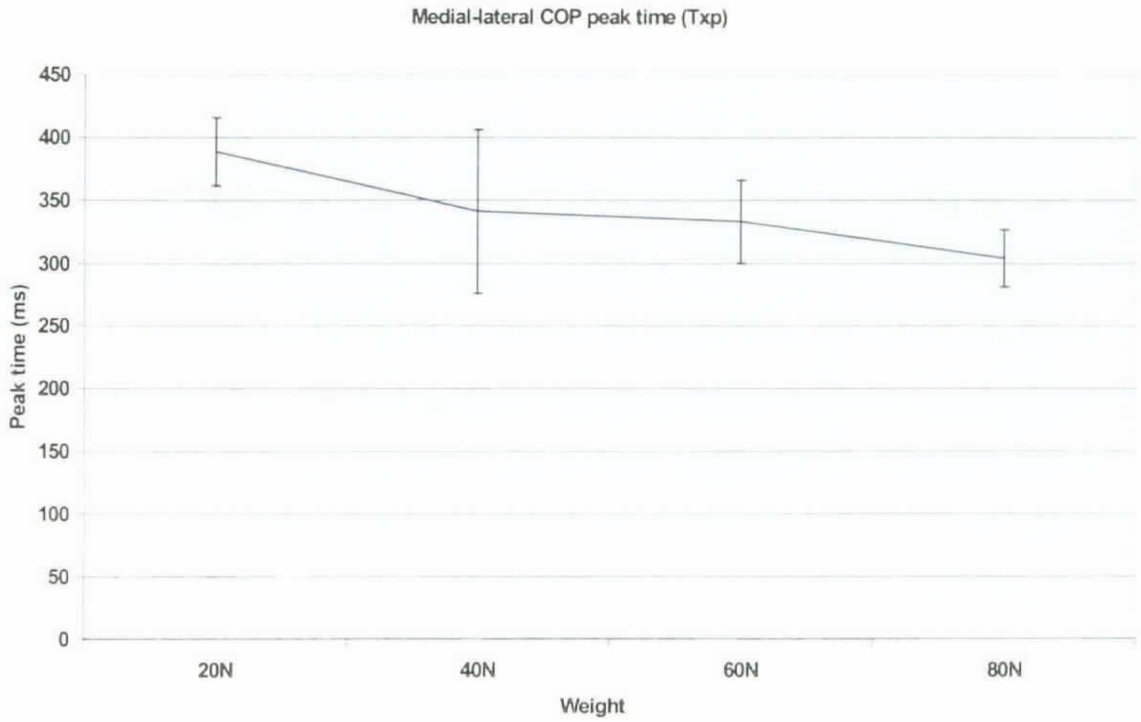


Figure 4.6 Medio-lateral COP peak time (Txp)

The figure shows that the medio-lateral COP peak time decreases as the lifting weight increases.

4.1.3 Antero-posterior COP displacement

The antero-posterior COP displacement in response to the sudden release of load seems to be affected by the lifting posture as well as the lifting weight. Adopting a symmetric stoop lifting posture seems to result in the least posterior COP displacement after sudden release of load, while adopting a symmetric squat lifting posture showed the greatest posterior COP displacement under same lifting conditions after sudden unload, furthermore, the heavier the lifting weight, the more posterior the COP displacement (Figure 3.16). The mean posterior limits of the COP displacement (L) for all of the subjects under different lifting conditions and the extrapolated limit of release of load (Lo) for all of the subjects using different lifting postures are shown in Table 4.6. By using linear regression with repeated measures, the mean extrapolated limit of release of load for symmetric squat lift, symmetric stoop lift and asymmetric stoop lift were 114N (S.D. 20N, R^2 = 0.919), 205N (S.D. 68N, R^2 = 0.627) and 173N (S.D. 42N, R^2 = 0.831) respectively. Repeated measures ANOVA showed there was an interaction between the within-subject factors (posture and weight) with $p=0.001$ and therefore the posterior limit of COP displacement (L) was analysed according to lifting weights. For lifting weights of 20N and 40N, there were no significant differences between the three lifting postures. However, for a lifting weight of 60N, the posterior limit of the COP displacement (L) of symmetric squat lift was significantly less (i.e., closer to the line joining the heels) than symmetric stoop lift and asymmetric stoop lift ($p=0.015$ and $p=0.019$, respectively). Similar result was found for a lifting weight of 80N with $p=0.04$ and $p=0.015$. For both 60N and 80N loads, no significant difference between the posterior limits of COP displacement under symmetric stoop lift and asymmetric stoop lift was found.

The posterior limit of COP displacement (L) was also analysed according to lifting postures. For all lifting postures, weight was found to have significant effect on the posterior limit of COP displacement with p=0.001. Contrast tests were used to compare the four levels of weight for all lifting postures (Tables 4.7-4.9). For symmetric squat lift, the posterior limit of COP displacement (L) for 80N load was significantly less than 60N, 40N and 20N. Similarly, the posterior limit of COP displacement, L, for 60N was significantly shorter than 40N and 20N while the value L for 40N was significantly shorter than 20N. For symmetric stoop lift, L for 80N load was significantly less than 60N and 40N, and for 60N load was significantly less than 40N. For asymmetric stoop lift, only the posterior limit L for 40N and 60N load was not significantly different.

Table 4.6 The mean posterior limits of the COP displacement (L) for all of the subjects under different lifting conditions and the extrapolated limit of release of load (Lo) for all of the subjects using different lifting postures

Subject No.	Posterior limit of COP displacement of symmetric squat lift after sudden unload, L (mm)				*Lsqo (N)	Posterior limit of COP displacement of symmetric stoop lift after sudden unload, L (mm)				*Lsto (N)	Posterior limit of COP displacement of symmetric stoop lift after sudden unload, L (mm)				*Lao (N)
	Sq20N	Sq40N	Sq60N	Sq80N		St20N	St40N	St60N	St80N		A20N	A40N	A60N	A80N	
1	71.3	25.5	20.8	15.5	88	65.1	47.0	60.9	33.2	176	87.6	66.4	67.0	47.6	162
2	88.2	63.1	55.9	40.2	131	90.5	60.5	76.8	53.8	200	81.8	84.1	76.2	39.8	155
3	114.7	79.0	63.2	56.3	132	77.2	64.6	46.9	66.9	312	109.8	116.2	117.1	103.4	178
4	85.3	63.5	33.9	24.4	98	75.5	32.0	58.6	53.9	337	68.3	55.1	78.2	69.8	165
5	87.6	63.6	34.1	24.0	97	62.8	57.9	57.8	39.7	206	76.5	64.9	47.6	41.2	143
6	76.8	51.7	46.8	21.5	107	52.4	51.8	37.8	27.6	146	98.1	99.3	87.2	94.6	171
7	111.3	92.2	71.8	62.4	151	71.9	58.7	49.9	43.3	168	80.3	61.5	50.7	18.2	103
8	78.7	81.4	86.0	83.6	113	93.3	92.5	98.0	93.1	204	107.3	88.1	98.1	90.6	178
9	123.6	95.1	66.0	32.2	102	102.4	94.7	51.3	49.6	123	76.5	68.8	59.4	58.4	257
10	76.6	50.1	37.2	34.0	120	75.8	81.0	52.6	52.7	184	73.4	56.4	65.8	47	223
Mean	91.4	66.5	51.6	39.4	114	76.7	64.1	59.0	51.4	205	85.9	86.1	74.7	61.1	173
S.D	18.3	21.2	20.4	21.6	20	15.1	19.9	17.0	18.5	72	14.4	20.1	21.5	27.7	56

Note: * Lsqo is the extrapolated limit of release of load for symmetric squat lift, Lsto is the extrapolated limit of release of load for symmetric stoop lift and Lao is the extrapolated limit of release of load for asymmetric stoop lift.

Table 4.7 Contrast tests comparing the posterior limits of COP displacement for symmetric squat lift among the four levels of weight

Lsq	20N	40N	60N	80N
20N				
40N	0.001*			
60N	0.001*	0.004*		
80N	0.001*	0.001*	0.004*	
* for $p < 0.05$				

Table 4.8 Contrast tests comparing the posterior limits of COP displacement for symmetric stoop lift among the four levels of weight

Lst	20N	40N	60N	80N
20N				
40N	0.024*			
60N	0.006*	0.482		
80N	0.001*	0.059	0.105	
* for $p < 0.05$				

Table 4.9 Contrast tests comparing the posterior limits of COP displacement for asymmetric stoop lift among the four levels of weight

La	20N	40N	60N	80N
20N				
40N	0.012*			
60N	0.026*	0.745		
80N	0.004*	0.031*	0.011*	
* for $p < 0.05$				

4.2 Electromyography Measurement

For the ten subjects using three lifting postures (symmetric squat, symmetric stoop and asymmetric stoop), four lifting weights (20, 40, 60 and 80N) and recording nine pairs of muscles (bilateral latissimus dorsi (LD), lumbar erector spinae (ES), external oblique (EO), internal oblique (IO), rectus abdominis (RA), biceps femoris (long head) (BF), tibialis anterior (TA), rectus femoris (RF) and gastrocnemius (lateral head) (G)), with three repetitions of each test condition, there were a total of 360 possible trials resulting 6480 muscle recordings. About 3.8% of the muscle recordings were lost due to electrode adhesion problems or instrumentation problems. During data analysis, latency (onset of response time after sudden release of load) and duration of response time were determined from each muscle's EMG record. Duration of co-contraction between anterior and posterior muscles was also studied. Definition of all the determined parameters can be found in section 3.4.4.

Among the nine pairs of muscles studied, the latissimus dorsi, external oblique, internal oblique, rectus abdominis, rectus femoris and tibialis anterior responded by contraction after sudden release of load in all trials, while the posterior muscles (lumbar erector spinae, biceps femoris and gastrocnemius) all responded by relaxation. Both internal oblique and latissimus dorsi muscles showed no response to the stimulation at low weight in symmetric stoop lift. The rectus abdominis and internal oblique muscles also showed no response to the stimulation at low weight in asymmetric stoop lift.

4.2.1 Muscle latency

The mean and standard deviation of the latency of each muscle under different load release conditions were determined (Tables 4.10-4.12 and Figures 4.7-4.9). It was found that the tibialis anterior and rectus femoris were the first two muscles to respond to the stimulation and were followed by the external oblique, internal oblique, rectus abdominis or erector spinae in almost all trials. Following the above muscles was the biceps femoris and then the latissimus dorsi, while the last muscle to respond was the gastronemius. For all lifting postures, a trend was observed where the latency of muscles which responded by contraction decreased with increasing lifting weight, and the latency of muscles which responded by relaxation increased with increasing lifting weight. For symmetric squat and stoop lift, the latency of the left side is very close to the right side, however, for asymmetric stoop lift, the latency of the right latissimus dorsi, erector spinae, biceps femoris and gastronemius are longer than the left side.

Table 4.10 Mean (standard deviation) response times of muscles after sudden release for symmetric squat lift

Type of response	Latency (S.D) in ms				
	Muscle	20N	40N	60N	80N
Contraction	SqLTA	52.3(16.3)	50.0(15.9)	59.4(12.2)	56.3(21.2)
Contraction	SqRTA	54.9(23.2)	53.6(23.4)	54.2(23.7)	52.4(16.1)
Contraction	SqLEO	72(19.8)	68.2(15.8)	56.8(21.7)	73.8(21.1)
Contraction	SqREO	69.0(21.9)	67.2(17.3)	67.2(26.3)	60.8(21.0)
Contraction	SqLIO	70.2(22.6)	59.6(38.3)	66.2(22.7)	62.6(24.5)
Contraction	SqRIO	70.1(27.2)	65.1(22.8)	60.2(17.4)	53.1(18.8)
Contraction	SqLLD	200(85.4)	181.0(68.4)	168.0(87.1)	172.6(81.8)
Contraction	SqRLD	255.1(114.1)	207.7(96.4)	173.2(90.7)	167.7(84.6)
Contraction	SqLRA	102.3(35.7)	78.2(27.9)	80.6(21.3)	72.0(16.9)
Contraction	SqRRA	96.7(24.5)	86.3(28.1)	82.2(27.5)	74.5(24.8)
Contraction	SqLRF	60.0(13.1)	54.7(11.6)	56.7(11.7)	61.6(13.5)
Contraction	SqRRF	63.5(13.0)	56.2(17.7)	58.5(18.4)	57.6(20.9)
Relaxation	SqLES	122.8(26.5)	125.4(32.7)	137.0(31.3)	148.8(30.3)
Relaxation	SqRES	119.6(28.3)	132.6(31.9)	140.5(37.2)	135.2(29.3)
Relaxation	SqLBF	156.4(66.3)	155.3(64.9)	186.5(74.5)	188.1(89.9)
Relaxation	SqRBF	157.3(70.6)	150.0(51.3)	170.5(70.4)	173.2(70.1)
Relaxation	SqLG	258.9(114.7)	248.8(125.9)	285.5(120.4)	293.2(137.1)
Relaxation	SqRG	250.8(107.6)	239.0(108.5)	273.6(105.8)	288.2(110.3)

For explanation of abbreviations refer to page xvii

Table 4.11 Mean (standard deviation) response times of muscles after sudden release for symmetric stoop lift

Type of response	Latency (S.D) in ms				
	Muscle	20N	40N	60N	80N
Contraction	StLTA	66.3(22.0)	72.2(20.9)	66.1(27.6)	69.6(25.7)
Contraction	StRTA	76.6(20.4)	75.2(20.8)	68.5(19.1)	68.6(23.2)
Contraction	StLEO	93.0(20.6)	90.5(20.8)	84.1(22.9)	76.8(36.5)
Contraction	StREO	95.1(27.4)	94.4(21.4)	83.5(27.2)	83.6(25.6)
Contraction	StLIO	NR	96.2(28.8)	95.0(31.4)	85.6(24.9)
Contraction	StRIO	NR	93.5(31.5)	85.5(31.8)	88.3(33.1)
Contraction	StLLD	NR	163.1(89.8)	155.1(78.1)	139.5(52.2)
Contraction	StRLD	NR	170.0(86.3)	163.5(73.3)	138.6(73.1)
Contraction	StLRA	110.3(29.4)	104.0(28.6)	90.4(20.3)	103.8(23.6)
Contraction	StRRA	115.6(26.9)	111.0(23.8)	102.5(29.3)	100.6(28.4)
Contraction	StLRF	83.5(20.7)	79.2(13.1)	77.5(21.6)	76.3(16.5)
Contraction	StRRF	82.0(13.4)	77.7(13.6)	72.7(15.4)	71.6(15.3)
Relaxation	StLES	63.0(15.4)	90.1(23.8)	97.2(25.8)	109.5(34.3)
Relaxation	StRES	55.1(14.7)	86.0(18.8)	110.8(41.9)	115.0(40.8)
Relaxation	StLBF	137.7(67.5)	150.7(62.6)	155.7(49.1)	179.5(65.1)
Relaxation	StRBF	135.3(49.3)	136.4(67.9)	162.6(61.8)	182.6(59.4)
Relaxation	StLG	180.0(84.7)	185.0(93.4)	223.1(85.4)	307.8(120.7)
Relaxation	StRG	185.9(95.1)	188.2(58.2)	212.4(97.2)	295.4(109.7)

For explanation of abbreviations refer to page xvii; NR=No Response

Table 4.12 Mean (standard deviation) response times of muscles after sudden release for asymmetric stoop lift

Type of response	Latency (S.D) in ms				
	Muscle	20N	40N	60N	80N
Contraction	ATA	62.8(22.5)	70.8(17.8)	64.7(11.0)	67.2(18.6)
Contraction	ARTA	60.1(15.8)	69.7(18.0)	60.1(15.4)	59.6(13.8)
Contraction	ALEO	90.0(23.3)	61.7(15.1)	68.0(20.1)	69.8(13.6)
Contraction	AREO	78.4(15.8)	60.3(8.6)	64.0(14.9)	60.0(8.9)
Contraction	ALIO	NR	106.5(34.7)	100.0(35.7)	110.1(36.7)
Contraction	ARIO	NR	95.3(32.8)	82.3(27.7)	72.5(18.5)
Contraction	ALLD	191.2(117.7)	170.1(84.9)	168.2(72.6)	157.8(88.8)
Contraction	ARLD	246.5(98.5)	252.3(120.9)	192.3(108.0)	84.2(23.2)
Contraction	ALRA	NR	NR	66.8(23.1)	76.3(27.6)
Contraction	ARRA	NR	NR	93.7(23.3)	79.8(36.9)
Contraction	ALRF	82.3(14.1)	66.4(16.8)	77.3(12.6)	68.2(13.0)
Contraction	ARRF	80.5(16.4)	71.1(15.6)	73.2(12.5)	72.5(4.8)
Relaxation	ALES	65.3(14.0)	68.0(14.2)	72.0(13.0)	87.7(32.2)
Relaxation	ARES	70.9(15.2)	74.0(11.6)	97.5(26.6)	111.8(41.7)
Relaxation	ALBF	107.2(22.7)	112.2(52.2)	109.3(34.0)	117.6(53.8)
Relaxation	ARBF	152.3(59.7)	144.3(68.2)	137.2(60.7)	142.6(41.2)
Relaxation	ALG	131.2(65.6)	125.3(57.0)	126.4(67.1)	145.3(43.2)
Relaxation	ARG	185.4(69.6)	195.0(85.7)	248.0(94.2)	305.3(113.8)

For explanation of abbreviations refer to page xvii; NR=No Response

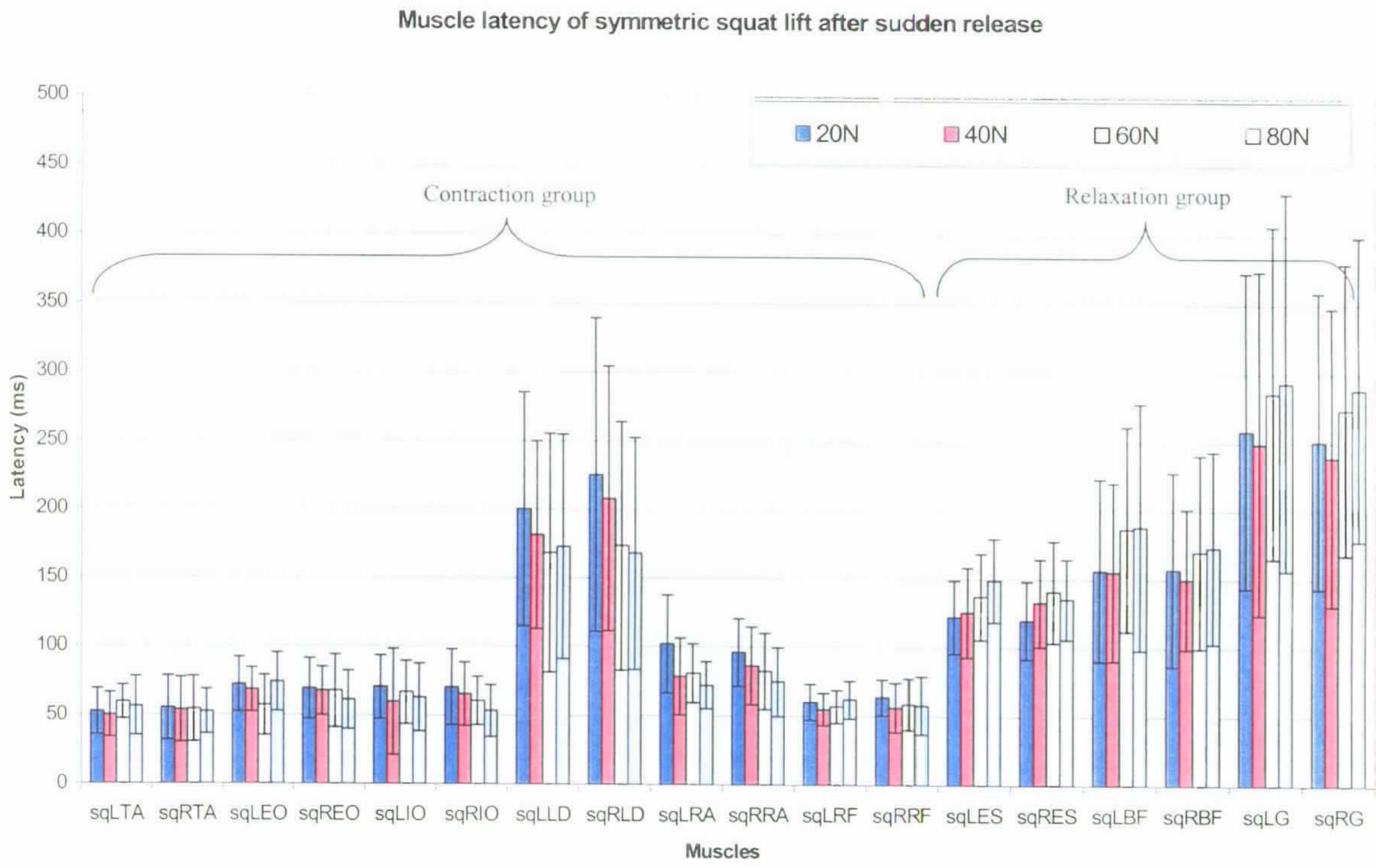


Figure 4.7 Mean and S.D. of muscle latency with different lifting weights after sudden release for symmetric squat lift
Note: For explanation of abbreviations refer to page xvii

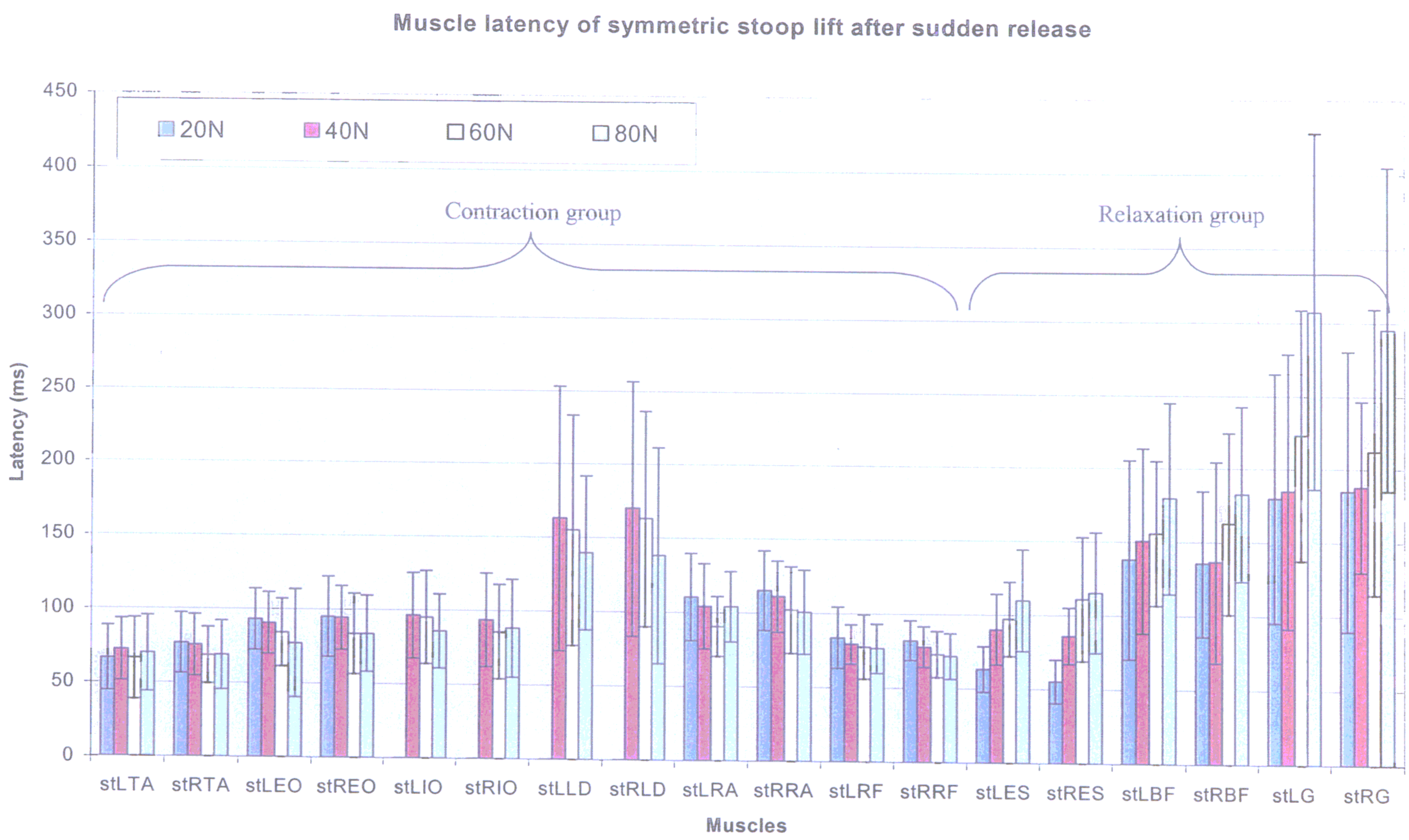


Figure 4.8 Mean and S.D. of muscle latency with different lifting weights after sudden release for symmetric stoop lift

Note 1: For explanation of abbreviations refer to page xvii

Note 2: Both bilateral IO and LD had no response to the stimulation at the lifting weight of 20N

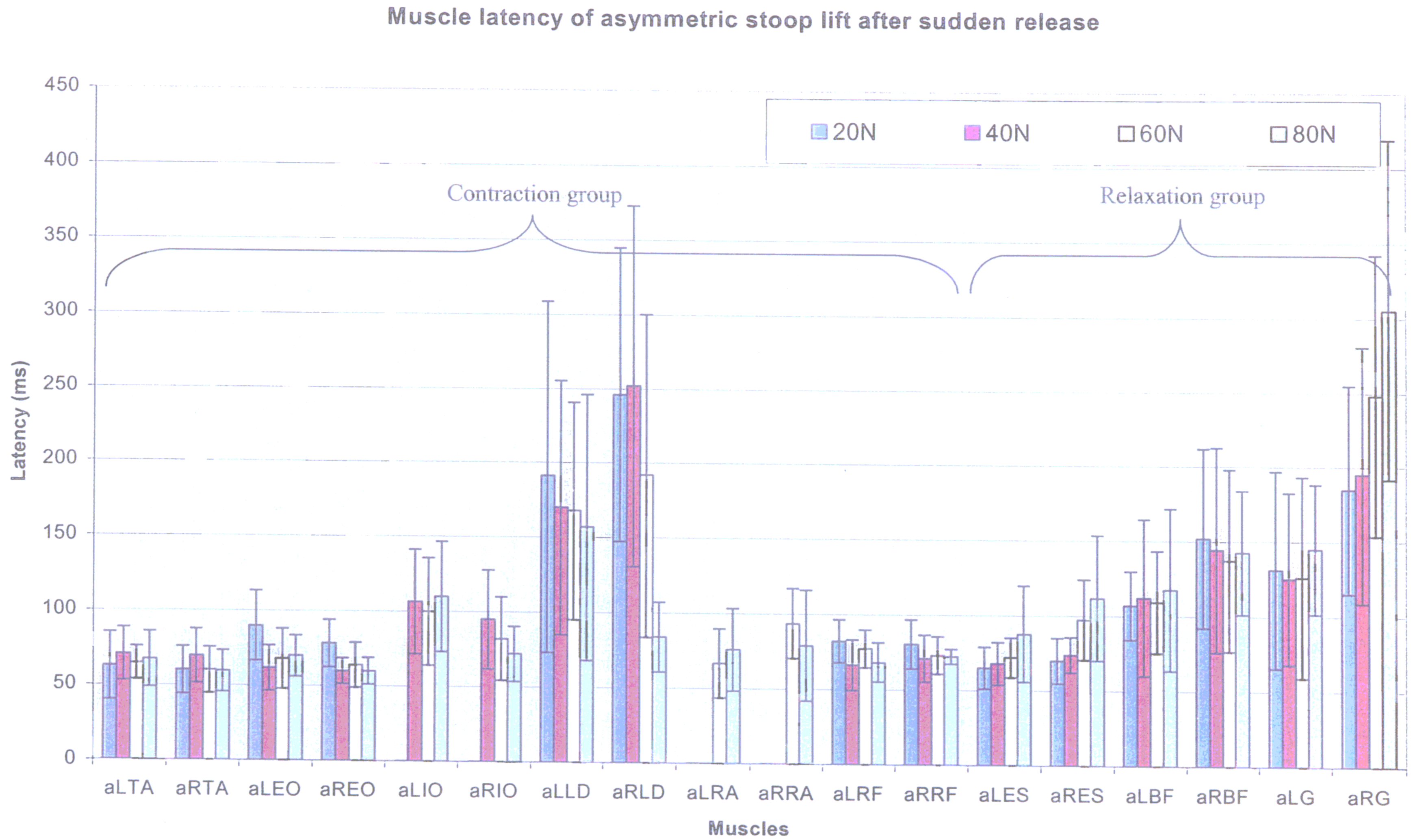


Figure 4.9 Mean and S.D. of muscle latency with different lifting weights after sudden release for asymmetric stoop lift
Note 1: For explanation of abbreviations refer to page xvii
Note 2: Bilateral IO and RA had no response to the stimulation at the lifting weight of 20N and 20N and 40N respectively

Repeated measures ANOVA were conducted for the muscle latency. According to the pattern of response, the nine bilateral muscles were divided into contraction group and relaxation group for data analysis. The contraction group consisted of the tibialis anterior (TA), external oblique (EO), internal oblique (IO), rectus abdominis (RA), rectus femoris (RF) and latissimus dorsi (LD) while the relaxation group consisted of erector spinae (ES), biceps femoris (BF) and gastronemius (G). The analyses included four within-subject factors:

Factor 1: Muscles, with six levels for the contraction group (TA, EO, IO, LD, RA and RF) and three levels for the relaxation group (ES, BF and G)

Factor 2: Lifting postures, with three levels (symmetric squat, symmetric stoop and asymmetric stoop)

Factor 3: Lifting weights, with four levels (20, 40, 60, and 80N)

Factor 4: Sides, with two levels (left and right side)

It was shown that there were significant interactions between the within-subject factors with $p<0.05$ for both the contraction group and relaxation group muscles and therefore the data was analysed according to the muscles.

For TA, EO, RF and BF, the three left within-subject factors (posture, weight and side) were not statistically significant with $p>0.05$. Side has no significant effect on the latency with $p=0.205$, 0.075, 0.745 and 0.093 for TA, EO, RF and BF, respectively. However, the main effect of posture was found to be significant with $p=0.01$, 0.001, 0.001 and 0.003 for TA, EO, RF and BF respectively. Contrast tests were performed to compare the difference between the three levels (Table 4.13).

Table 4.13 Contrast tests comparing the latency among the three postures for tibialis anterior, external oblique, rectus femoris and biceps femoris

Muscle	Contrast tests comparing the latency among the three levels of posture		
	Symmetric squat and asymmetric stoop	Symmetric stoop and asymmetric stoop	Symmetric squat and symmetric stoop
TA	0.063	0.080	0.023*
EO	0.463	0.003*	0.001*
RF	0.001*	0.089	0.001*
BF	0.008*	0.003*	0.317
* for p<0.05			

The main effect of weight was found to be significant for EO and RF with p=0.022 and p=0.019 but not significant for TA and BF with p=0.834 and p=0.163 respectively. Contrast tests were performed to compare the difference between the four levels for EO and RF (Table 4.14).

Table 4.14 Contrast tests comparing the latency among the four weights for external oblique and rectus femoris

EO	20N	40N	60N	80N	RF	20N	40N	60N	80N
20N					20N				
40N	0.009*				40N	0.005*			
60N	0.051	0.493			60N	0.077	0.357		
80N	0.044*	0.494	0.944		80N	0.064	0.863	0.564	
* for p < 0.05									

Since there was no significant interaction between the within-subject factors and posture is the only factor that has significant effect on the latency for TA and BF, the pooled mean of latency for TA and BF were plotted as follows (Figure 4.10):

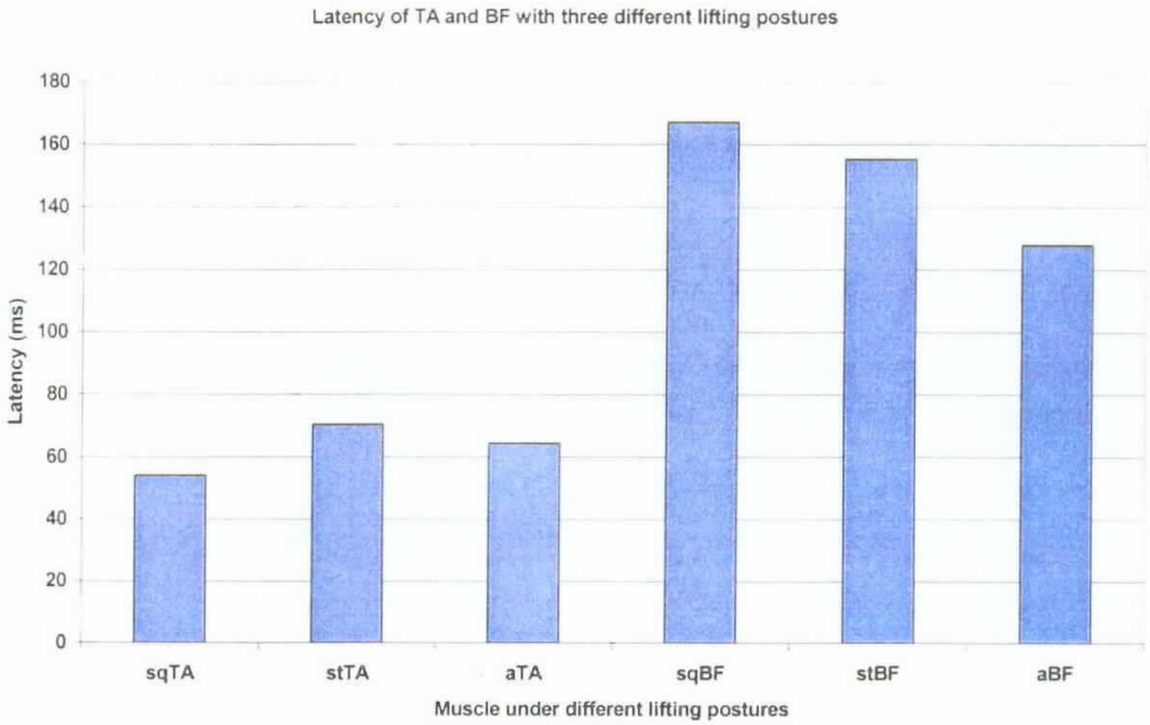


Figure 4.10 Pooled mean of latency of TA and BF with different lifting postures
Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For TA, the latency of symmetric stoop lift was significantly longer than symmetric squat lift (p=0.023). The latency of asymmetric stoop was found to be longer than symmetric squat and shorter than symmetric stoop, however, these differences were not significant (p=0.063 and p=0.08, respectively). For BF, the latency of asymmetric stoop was significantly shorter than symmetric squat and symmetric stoop (p=0.008 and p=0.003, respectively). The latency of symmetric stoop was shorter than symmetric squat, however, this difference was not significant (p=0.317).

There was also no significant interaction between the within-subject factors for EO and RF. Posture and weight factors had a significant effect on the latency. The pooled mean of latency for EO and RF were plotted as follows (Figure 4.11):

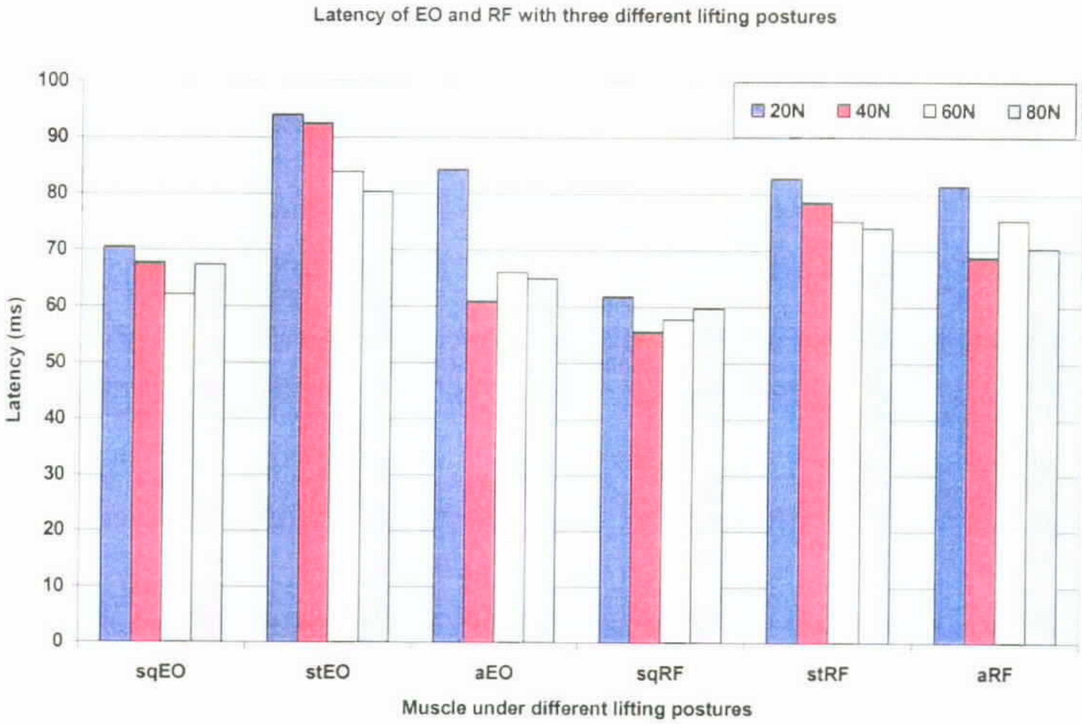


Figure 4.11 Pooled mean of latency of EO and RF with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For EO, the latency of symmetric stoop lifting was significantly longer than symmetric squat and asymmetric stoop lifting ($p=0.001$ and $p=0.003$, respectively). The latency difference between symmetric squat and asymmetric stoop lifting was not significant ($p=0.463$). The latency for 20N load was significantly longer than 40N and 80N ($p=0.009$ and $p=0.044$, respectively) while all other differences between weights were not significant. For RF, the latency of symmetric squat lifting was significantly shorter than symmetric stoop and asymmetric stoop lifting (both $p=0.001$). The latency difference between symmetric stoop lifting and asymmetric stoop was not significant ($p=0.089$). The latency for 20N load was significantly longer than 40N and all other differences between weights were not significant.

For IO, LD, RA, ES and G, the interactions between the three left within-subject factors (posture, weight and side) were statistically significant with and therefore analysed according to lifting weight. Data were analysed according to side (left and right) if significant interaction was still found between the within-subject factors (posture and side) (Table 4.15).

Table 4.15 Muscle latency comparison using repeated measures ANOVA with two within-subject factors (posture and side) and contrast tests comparing the latency among the three levels of posture

Repeated measures ANOVA				Contrast tests comparing the latency among the three levels of posture		
Muscles	Weight	Within-subject factors	P value	sq-a	st-a	sq-st
LD	20N	Posture	0.001*	0.890	0.001*	0.001*
		Side	0.230	Effect was not significant		
	40N	Posture	0.427	Effect was not significant		
		Side	0.116	Effect was not significant		
	60N	Posture	0.151	Effect was not significant		
		Side	0.575	Effect was not significant		
	80N	Posture	0.122	Effect was not significant		
		Side	0.080	Effect was not significant		
RA	20N	Posture	0.001*	0.001*	0.001*	0.329
		Side	0.641	Effect was not significant		
	40N	Posture	0.001*	0.001*	0.001*	0.049*
		Side	0.324	Effect was not significant		
	60N	Posture	0.143	Effect was not significant		
		Side	0.025*	L was significantly different from R		
	80N	Posture	0.037*	0.719	0.101	0.003*
		Side	0.902	Effect was not significant		
ES	20N	Posture	0.001*	0.001*	0.051	0.001*
		Side	0.708	Effect was not significant		
	40N	Posture	0.001*	0.001*	0.017*	0.002*
		Side	0.418	Effect was not significant		
	60N	Posture	0.001*	0.001*	0.065	0.031*
		Side	0.046*	L was significantly different from R		
	80N	Posture	0.001*	0.001*	0.266	0.028*
		Side	0.623	Effect was not significant		
IO	20N	Posture	0.001*	0.001*	0.001*	0.001*
		Side	0.993	Effect was not significant		
	40N	Posture	0.006*	0.029*	0.529	0.005*
		Side	0.732	Effect was not significant		
	60N	Posture	0.009*	0.012*	0.934	0.018*
		Side	0.148	Effect was not significant		
	80N	Left	0.007*	0.015*	0.038*	0.105
		Right	0.011*	0.068	0.130	0.015*
G	20N	Posture	0.011*	0.015*	0.471	0.009*
		Side	0.573	Effect was not significant		
	40N	Posture	0.083	Effect was not significant		
		Side	0.326	Effect was not significant		
	60N	Posture	0.009*	0.006*	0.305	0.046*
		Side	0.221	Effect was not significant		
	80N	Left	0.006*	0.009*	0.001*	0.825
		Right	0.918	Effect was not significant		
*for p<0.05						

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For LD, the main effect of posture was not significant for all weights except 20N (p=0.001) and the main effect of side was not significant for all weights. The pooled mean of latency of LD was plotted as follows (Figure 4.12):

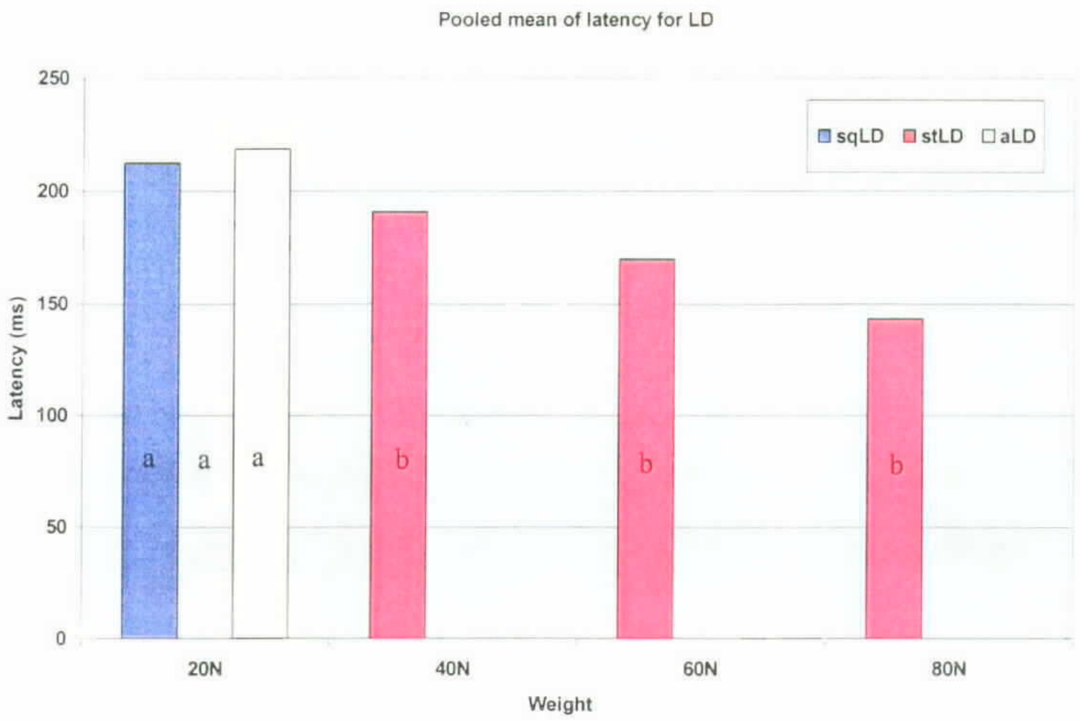


Figure 4.12 Pooled mean of latency for LD with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift
a=Pooled mean of latency for LD with only side had no significant effect and LD has no response to the sudden release of load at the weight of 20N when symmetric stoop lift was adopted
b=Pooled mean of latency for LD with both posture and side had no significant effect

At the lifting weight of 20N, LD showed no response to the sudden release of load when symmetric stoop lift was adopted. The latency differences between symmetric stoop lifting and symmetric squat and asymmetric stoop lifting were significant (both p=0.001). The latency of symmetric squat lifting shorter than asymmetric stoop lifting, however, the difference was not significant (p=0.89). A trend of decreasing with increasing lifting weight was also observed.

For RA, the main effect of posture was significant for all lifting weights except 60N (p=0.143), and the main effect of side was not significant for all lifting weights except 60N (p=0.025). The pooled mean of latency of RA is shown (Figure 4.13):

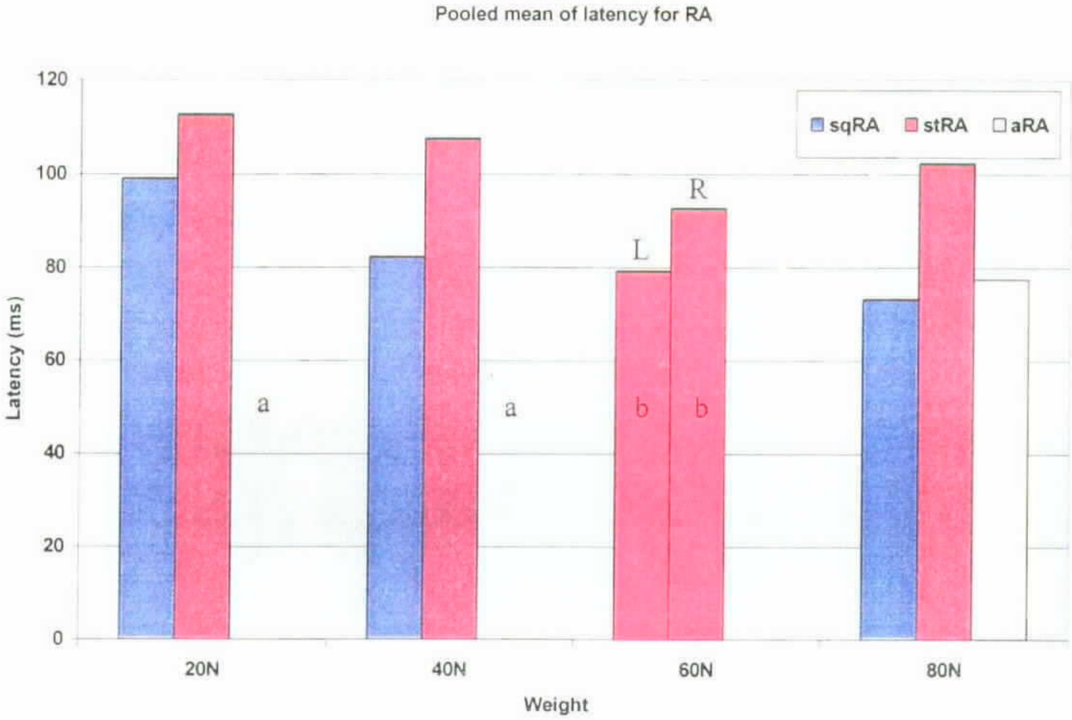


Figure 4.13 Pooled mean of latency for RA with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

a=RA has no response to the sudden release of load at the weight of 20N and 40N when asymmetric stoop lift was adopted

b=Pooled mean of latency for RA at lifting weight of 60N with posture had no significant effect while side had significant effect

At the lifting weight of 60N, posture had no significant effect on latency (p=0.143) while the latency of left RA was significantly shorter than the right RA (p=0.025). At the lifting weights of 20N and 40N, RA had no response to the sudden release of load when asymmetric stoop lift was adopted. For both 20N and 40N, symmetric squat and symmetric stoop lift were significantly different from asymmetric stoop lift (both p=0.001). The latency of symmetric squat lifting was shorter than symmetric stoop lifting both 20N and 40N loads. However, this difference was only significant for 40N load (p=0.049). At the lifting weight of 80N, the latency of symmetric stoop lift was significantly longer than symmetric squat lift (p=0.003) but not significantly longer than asymmetric stoop lift (p=0.101). The latency of asymmetric stoop lift was also longer than symmetric squat lift, but again not significantly so (p=0.719). By comparing the symmetric squat and stoop lifting for 20N, 40N and 80N, a trend of decreasing latency with increase in lifting weight was also seen.

For ES, the main effect of posture was significant for all lifting weights while the main effect of side was not significant for all lifting weights except for 60N (p=0.046). The pooled mean of latency of ES is shown in figure 4.14:

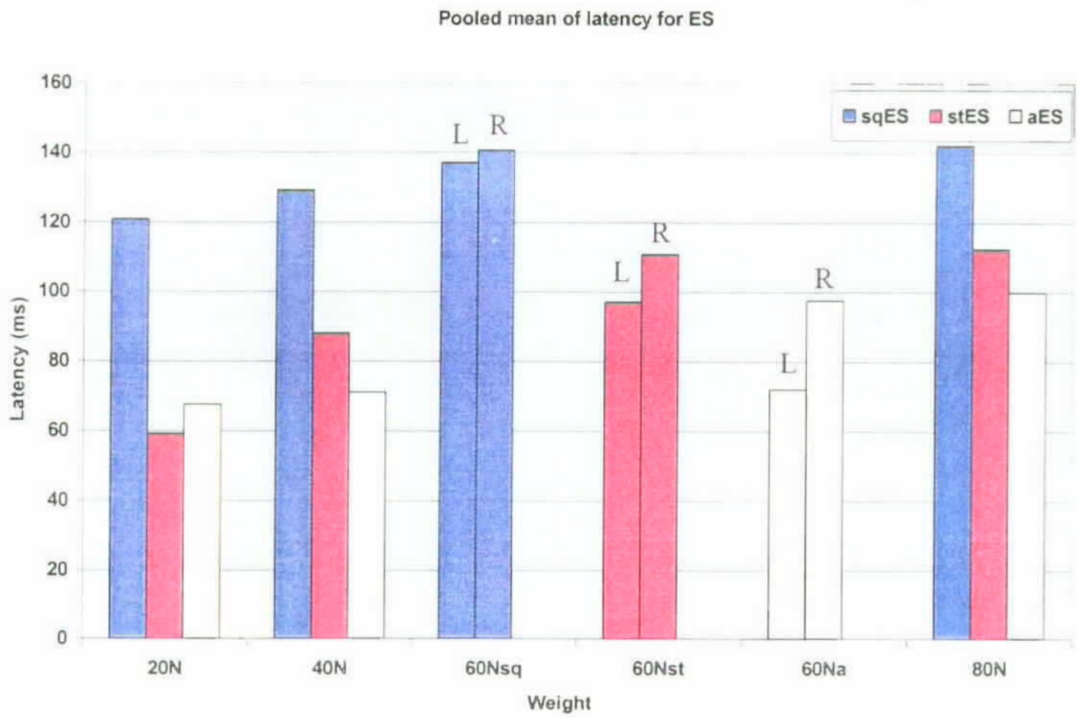


Figure 4.14 Pooled mean of latency for ES with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

At the lifting weight of 60N, both posture and side have significant effect on latency with p=0.001 and p=0.046 respectively. Contrast tests showed that the latency of symmetric squat lift was significantly longer than symmetric stoop and asymmetric stoop lift (p=0.001 and p=0.031, respectively). The latency of symmetric stoop lift was longer than asymmetric stoop lift, but the difference was not statistically significant (p=0.065). At lifting weights of 20N and 80N, the latency of symmetric squat lift was significantly longer than symmetric stoop lift and asymmetric stoop lift (both p=0.001 for 20N and with p=0.001 and p=0.028 respectively for 80N). The latency of asymmetric stoop lift for 20N load was longer than symmetric stoop lift, but the difference was not significant (p=0.051). for 80N load, the latency of asymmetric stoop lift was not significantly shorter than symmetric stoop lift (p=0.266). At the lifting weight of 40N, the latency of symmetric squat lift was significant longer than symmetric stoop lift and asymmetric stoop lift (p=0.002 and p=0.001, respectively). The latency of symmetric stoop lift was also significantly longer than asymmetric stoop (p=0.017). Trends where by the latency increased as the lifting weight increased and the latency of symmetric squat lift was longest and latency of asymmetric stoop lift shortest were found among the three lifting postures for 40N, 60N and 80N.

For IO, the main effect of posture was significant for 20N, 40N and 60N with p=0.001, 0.006 and 0.009 respectively. The main effect of side was not significant for 20N, 40N and 60N (p=0.993, 0.732 and 0.148, respectively). The interaction between the two left within-subject factors (posture and side) was still statistically significant with p=0.026 for lifting weight of 80N and therefore the data was analysed according to side. Posture was found to have significant effect on latency for both left and right IO. The pooled mean of latency of IO is shown (Figure 4.15):

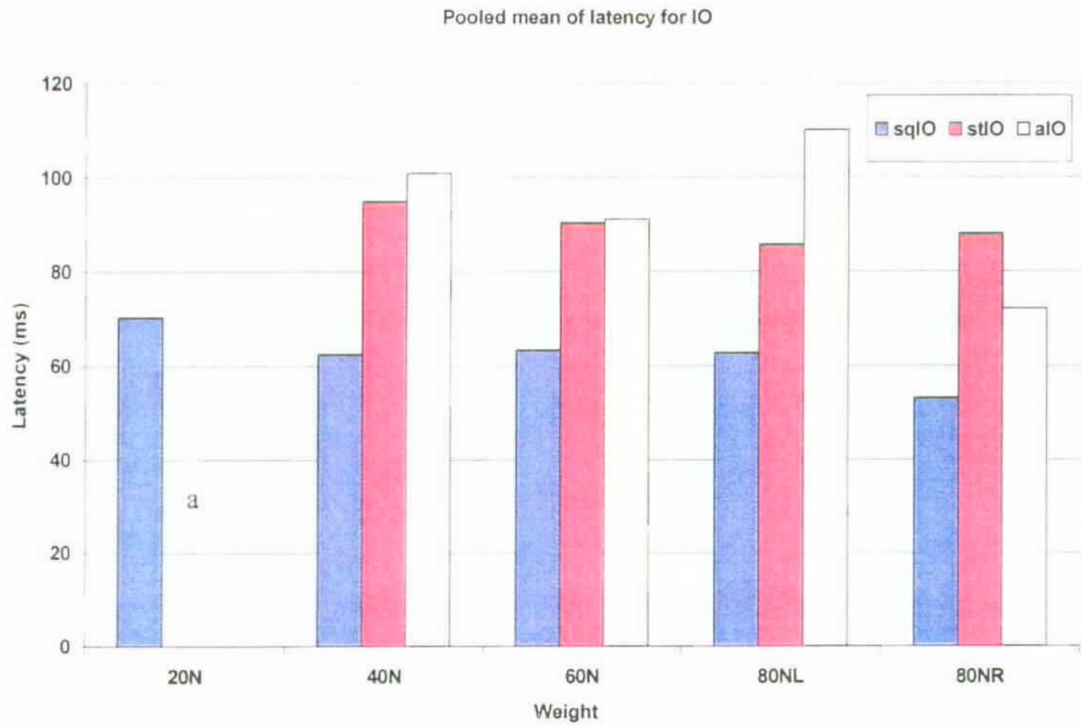


Figure 4.15 Pooled mean of latency for IO with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right
a=IO has no response to sudden release of load at the weight of 20N when symmetric stoop lift and asymmetric stoop lift was adopted

At the lifting weight of 20N, IO has no response to the sudden release of load when symmetric stoop or asymmetric stoop lifting postures were adopted. The latency of symmetric squat lift was significantly different from symmetric stoop lift and asymmetric stoop lift (both p=0.001). For lifting weights of 40N and 60N, the latency of symmetric squat was significantly shorter than symmetric stoop lift and asymmetric stoop lift with p=0.005 and 0.029 respectively for 40N, and p=0.018 and p=0.012 respectively for 80N. The latency of symmetric stoop lift was also shorter than asymmetric stoop, however, the difference was not significant with p=0.529 for 40N and p=0.934 for 60N. At the lifting weight of 80N, the latency of asymmetric stoop lift for LIO was significant longer than symmetric squat lift and symmetric stoop lift (p=0.015 and p=0.038, respectively). The latency of symmetric stoop is also longer than symmetric squat, however, the difference was not significant (p=0.105). For RIO, the latency of symmetric stoop lifting was found to be significantly longer than symmetric squat lifting (p=0.015). But for LIO and other lifting weights, the latency of asymmetric stoop lifting was shorter than symmetric stoop lifting, although the difference was not significant (p=0.13). It was shown that asymmetric stoop lifting had the longest latency and symmetric squat lifting the shortest latency among the three lifting postures for 40N, 60N and 80N load.

For G, the main effect of posture was significant for lifting weights of 20N and 60N (p=0.011 and p=0.009, respectively) while no significance was found for 40N (p=0.083). The main effect of side was not significant for lifting weights of 20N, 40N and 60N (p=0.573, 0.326 and 0.221, respectively). Similar to IO, the interaction between the two left within-subject factors (posture and side) was still statistically significant with p=0.034 for lifting weight of 80N, and therefore the data was analysed according to side. Posture was found to have significant effect on latency for left G with p=0.006 but not significant for right G (p=0.918). The pooled mean of latency of G was plotted as follows (Figure 4.16):

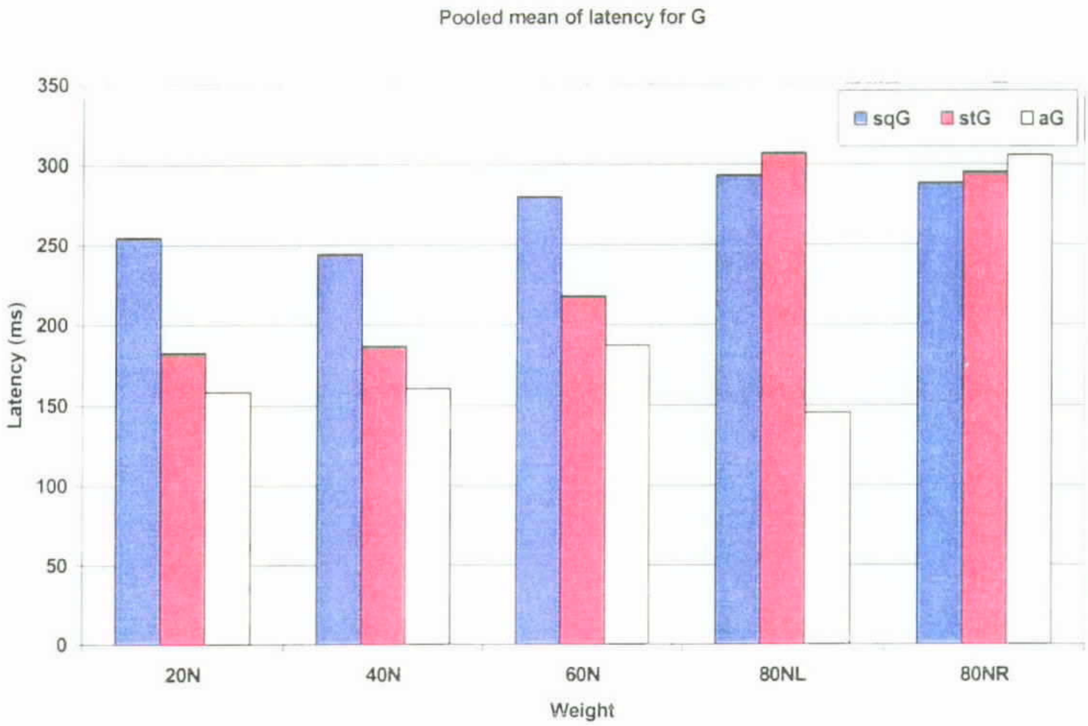


Figure 4.16 Pooled mean of latency for G with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

At the lifting weight of 20N and 60N, the latency of symmetric squat lifting was significantly longer than symmetric stoop and asymmetric stoop lifting (p=0.009 and p=0.015 respectively for 20N and p=0.046 and p=0.006, respectively for 60N). Also, the latency of symmetric stoop lifting was longer than asymmetric stoop lifting for both 20N and 60N, however, these differences were not significant (p=0.471 for 20N and p=0.305 for 60N). At lifting weight of 40N, it was found that the latency of symmetric squat lifting was longer than symmetric stoop and asymmetric stoop lifting and that the latency of symmetric stoop lifting was longer than asymmetric stoop lifting, however, repeated measures ANOVA showed that both posture and side had no significant effect on the latency (p=0.083 and 0.326, respectively). At the lifting weight of 80N, the latency of asymmetric stoop lifting was significantly shorter than symmetric squat and symmetric stoop lifting for LG (p=0.009 and p=0.001, respectively). In contrast to the latency of other lifting weights, the latency of LG in symmetric stoop lifting is longer than symmetric squat lifting for 80N weight although the difference was not significant (p=0.825). For RG, posture was found to have no significant effect on latency (p=0.918). It was shown the latency of symmetric squat lifting was the longest and the latency of asymmetric stoop lifting was the shortest among the three lifting postures for 20N, 40N and 60N.

With a few exceptional cases, the latency for most of the contraction group muscles (TA, EO, RF and RA) was found to be the longest when symmetric stoop lifting posture was adopted. The latency of TA and RF was longer when asymmetric stoop lift was adopted than when symmetric squat lift was adopted. The latency of EO was similar using either the asymmetric stoop lift or the symmetric squat lift. For IO, the latency was longest when asymmetric stoop lift was adopted, followed by symmetric stoop lift and shortest when symmetric squat lift was adopted. Posture was found to have no significant effect on the latency of LD. For relaxation group muscles, the latency was found to be the longest when the symmetric squat lift was adopted for all muscles (BF, ES and G), and shortest for BF and G when the asymmetric stoop lift was adopted. The latency of ES was similar when symmetric stoop lifting or asymmetric stoop lifting was adopted. For all lifting postures, a trend of latency decrease with increase in lifting weight for most of the contraction group muscles (EO, RF, LD and RA) was found. Lifting weight had no significant effect on latency for TA, and the latency of IO was similar for all lifting weights. For relaxation group muscles, a trend of latency increase with increase in lifting weight was found for ES and G. Lifting weight had no significant effect on the latency of BF. Side had no significant effect on latency for all muscles except RA and ES at the lifting weight of 60N.

4.2.2 Duration of response

The mean and standard deviation of the duration of response under different load release conditions were determined for each muscle (Tables 4.16-4.18 and Figures

4.17-4.19). For the muscles responding by contraction, duration of response refers to the duration of contraction response. For the muscles responding by relaxation, duration of response refers to the duration of the relaxation response. Generally, the erector spinae (ES) had the shortest duration of response, followed by rectus abdominis (RA), gastronemius (G), internal oblique (IO) or latissimus dorsi (LD), biceps femoris (BF), rectus femoris (RF) and tibialis anterior (TA) in ascending order. For all three lifting postures (symmetric squat lift, symmetric stoop lift and asymmetric stoop lift), the duration of response increased as the lifting weight increased for all muscles, except for BF in symmetric squat lifting. The duration of response was similar for left side and right side for all muscles when the symmetric squat lift or symmetric stoop lift was adopted. For contraction group muscles, the duration of the response of left side muscles was usually shorter than right side when the asymmetric stoop lift was adopted. However, for relaxation group muscles, the duration of response of left side muscles was usually longer than the right side when the asymmetric stoop lift was adopted.

Table 4.16 Mean (standard deviation) duration of response times of muscles for symmetric squat lift

Type of response	Duration of response (S.D) in ms				
	Muscle	20N	40N	60N	80N
Contraction	SqLTA	235.2(49.5)	288.4(59.8)	350.0(79.3)	540.4(203.7)
Contraction	SqRTA	208.7(43.6)	322.7(118.2)	416.5(157.7)	637.2(233.2)
Contraction	SqLEO	270.0(105.8)	287.5(65.2)	290.3(67.9)	321.3(141.6)
Contraction	SqREO	284.5(74.3)	331.7(78.7)	328.8(74.9)	381.4(80.8)
Contraction	SqLIO	105.0(32.4)	139.6(51.5)	175.7(77.4)	300.3(123.1)
Contraction	SqRIO	119.3(31.3)	140.5(43.5)	180.8(43.6)	264.5(130.0)
Contraction	SqLLD	139.5(32.4)	158.5(25.4)	187.7(70.1)	251.6(75.3)
Contraction	SqRLD	120.4(31.1)	159.7(40.6)	164.4(68.3)	298.4(89.3)
Contraction	SqLRA	97.5(30.4)	99.5(40.0)	140.5(52.5)	216.3(126.4)
Contraction	SqRRA	81.9(25.8)	107.8(47.3)	147.5(70.9)	227.1(125.9)
Contraction	SqLRF	214.3(77.4)	254.1(80.1)	424.2(154.5)	549.5(143.8)
Contraction	SqRRF	261.5(79.9)	299.6(90.9)	398.3(99.1)	461.5(160.0)
Relaxation	SqLES	89.6(25.5)	115.4(55.6)	104.1(42.1)	137.8(63.2)
Relaxation	SqRES	61.5(13.8)	88.7(31.6)	136.8(77.9)	135.0(111.2)
Relaxation	SqLBF	361.2(111.7)	339.5(107.2)	347.5(121.3)	174.3(55.6)
Relaxation	SqRBF	292.3(85.7)	250.6(91.1)	262.8(57.6)	168.1(54.1)
Relaxation	SqLG	130.2(34.4)	133.0(25.0)	192.7(77.3)	245.4(78.1)
Relaxation	SqRG	125.3(31.7)	138.5(37.3)	240.7(102.2)	305.6(115.5)

For explanation of abbreviations refer to page xvii; NR=No Response

Table 4.17 Mean (standard deviation) duration of response times of muscles for symmetric stoop lift

Type of response	Duration of response (S.D) in ms				
	Muscle	20N	40N	60N	80N
Contraction	StLTA	152.9(52.2)	252.9(73.6)	355.8(204.1)	628.0(241.7)
Contraction	StRTA	176.5(79.9)	302.5(122.2)	447.1(162.4)	585.9(231.4)
Contraction	StLEO	85.0(19.9)	135.0(36.5)	168.0(50.8)	320.0(106.8)
Contraction	StREO	126.5(32.7)	131.5(37.9)	176.8(58.1)	376.5(198.8)
Contraction	StLIO	NR	135.3(42.4)	214.0(41.3)	350.5(78.0)
Contraction	StRIO	NR	128.0(55.7)	204.7(54.6)	341.3(101.3)
Contraction	StLLD	NR	134.0(44.7)	187.7(104.3)	255.8(218.1)
Contraction	StRLD	NR	153.2(36.7)	171.4(47.9)	218.4(74.5)
Contraction	StLRA	45.3(11.1)	114.0(41.0)	189.0(56.8)	153.0(47.8)
Contraction	StRRA	40.0(12.3)	69.0(18.9)	169.0(55.2)	230.0(69.8)
Contraction	StLRF	281.3(56.8)	307.2(82.7)	344.0(161.6)	483.3(208.2)
Contraction	StRRF	251.4(118.8)	283.6(126.0)	331.1(162.1)	527.3(231.5)
Relaxation	StLES	132.5(42.0)	123.1(52.3)	130.5(42.4)	107.1(33.0)
Relaxation	StRES	121.0(32.3)	115.8(41.4)	89.0(42.2)	84.0(36.1)
Relaxation	StLBF	149.1(99.2)	177.7(113.6)	216.3(78.6)	237.1(71.2)
Relaxation	StRBF	140.8(47.1)	214.2(91.1)	188.6(102.6)	218.1(50.6)
Relaxation	StLG	120.0(36.9)	122.0(41.7)	154.0(39.7)	235.1(111.7)
Relaxation	StRG	100.0(14.5)	102.0(42.3)	200.0(78.9)	261.6(161.9)

For explanation of abbreviations refer to page xvii; NR=No Response

Table 4.18 Mean (standard deviation) duration of response times of muscles for asymmetric stoop lift

Type of response	Duration of response (S.D) in ms				
	Muscle	20N	40N	60N	80N
Contraction	ATA	208.0(48.5)	256.6(69.5)	270.2(74.2)	452.1(160.6)
Contraction	ARTA	283.0(109.3)	346.0(235.7)	456.4(257.3)	636.9(308.5)
Contraction	ALEO	120.0(47.9)	122.0(40.6)	269.0(79.2)	427.0(134.8)
Contraction	AREO	140.0(48.0)	142.0(41.2)	325.0(64.2)	503.0(156.2)
Contraction	ALIO	NR	135.2(50.5)	169.3(67.8)	254.7(146.4)
Contraction	ARIO	NR	395.0(96.4)	402.0(120.4)	436.0(108.1)
Contraction	ALLD	82.0(17.9)	133.0(50.3)	110.8(41.6)	160.2(61.7)
Contraction	ARLD	105.0(29.7)	158.0(46.5)	145.1(33.9)	193.5(79.5)
Contraction	ALRA	NR	NR	122.0(37.3)	165.3(80.7)
Contraction	ARRA	NR	NR	143.0(43.8)	150.8(54.1)
Contraction	ALRF	132.0(47.8)	226.3(66.6)	281.0(94.7)	348.0(127.6)
Contraction	ARRF	143.2(16.4)	267.4(15.6)	275.8(12.5)	370.8(48.5)
Relaxation	ALES	50.0(14.0)	52.0(14.2)	87.0(13.1)	150.4(32.2)
Relaxation	ARES	40.0(15.2)	43.0(11.6)	73.0(26.6)	114.5(41.7)
Relaxation	ALBF	199.5(22.7)	286.3(52.2)	256.2(34.0)	389.9(53.8)
Relaxation	ARBF	233.0(59.7)	189.0(68.2)	293.0(60.7)	301.5(41.2)
Relaxation	ALG	132.0(65.6)	171.0(57.1)	215.0(67.1)	257.0(43.2)
Relaxation	ARG	116.7(69.6)	99.7(85.7)	165.9(94.2)	230.3(113.8)

For explanation of abbreviations refer to page xvii; NR=No Response

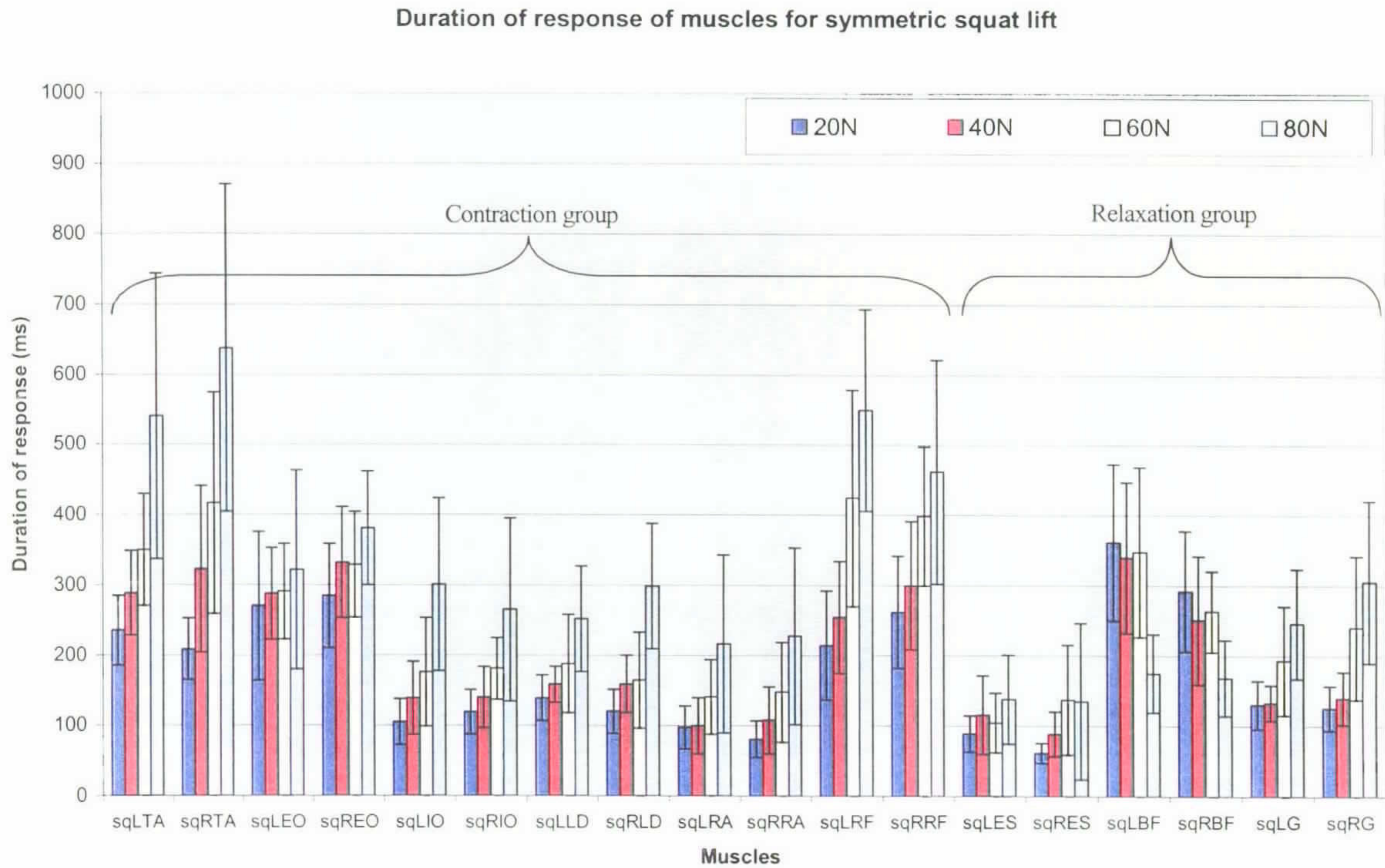


Figure 4.17 Mean and S.D. of muscle duration of response with different lifting weights after sudden release for symmetric squat lift

Note: For explanation of abbreviations refer to page xvii

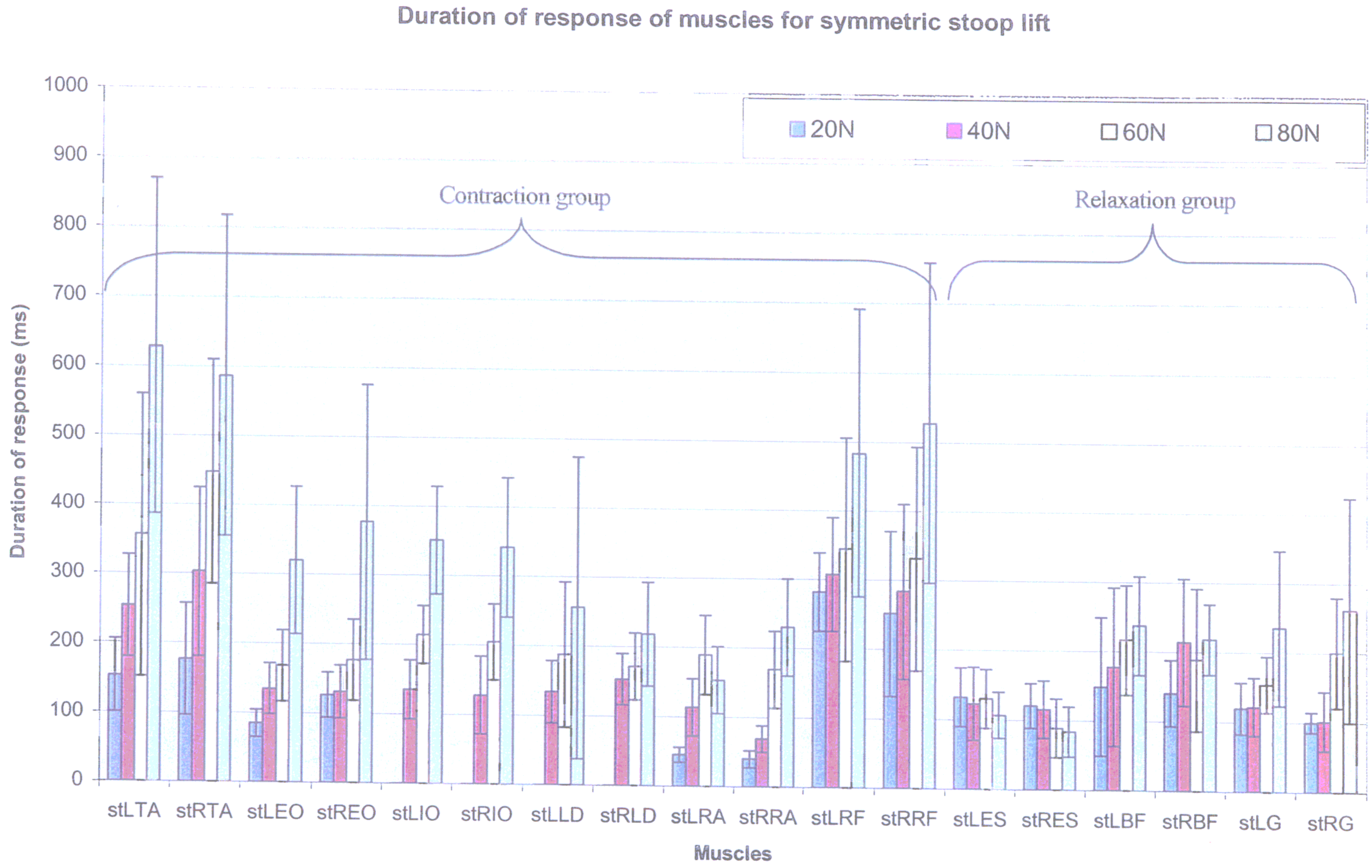


Figure 4.18 Mean and S.D. of muscle duration of response with different lifting weights after sudden release for symmetric stoop lift

Note 1: For explanation of abbreviations refer to page xvii

Note 2: Both bilateral IO and LD has no response to the stimulation at the lifting weight of 20N

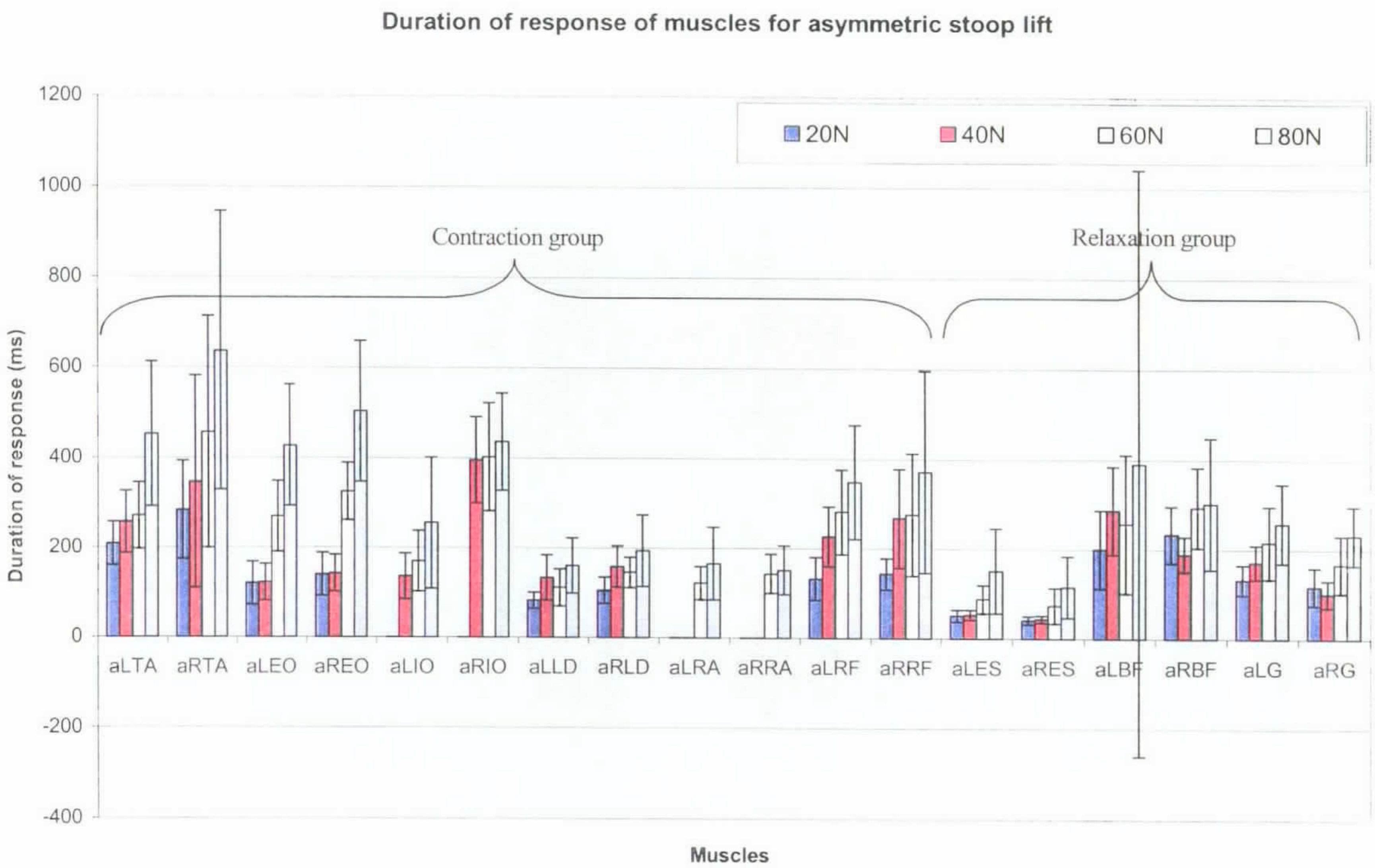


Figure 4.19 Mean and S.D. of muscle duration of response with different lifting weights after sudden release for asymmetric stoop lift

Note 1: For explanation of abbreviations refer to page xvii

Note 2: Bilateral IO and RA has no response to the stimulation at the lifting weight of 20N and 20N and 40N respectively

Similar to the analysis of muscle latency, repeated measures ANOVA was conducted to compare the duration of response between different conditions. According to the pattern of response, the nine bilateral muscles were divided into contraction and relaxation groups for data analysis. The analyses also included four within-subject factors: six levels of muscle for the contraction group and three levels of muscle for the relaxation group; three levels of lifting postures; four levels of lifting weights and two levels of side. It was shown that there were significant interactions between the within-subject factors for both the contraction group and relaxation group muscles and therefore the data were analysed according to the muscle.

Among all muscles, the three left within-subject factors (posture, weight and side) were not statistically significant for TA only. Posture has no significant effect on latency (p=0.949), but both weight and side had significant effect on latency (p=0.001 and p=0.032, respectively). Contrast tests were performed to compare the differences between the four levels of weight and it was found that the duration of response for 80N weight was significantly longer than all other weights (p=0.001); the duration of response for 60N weight was significantly longer than 20N and 40N (both p=0.001); the duration of response of weight 40N was significantly longer than 20N with also p=0.001. The duration of response was found to be longer for right side than the left side (p=0.032) for all lifting weights. The pooled mean of duration of response for TA is plotted in figure 4.20:

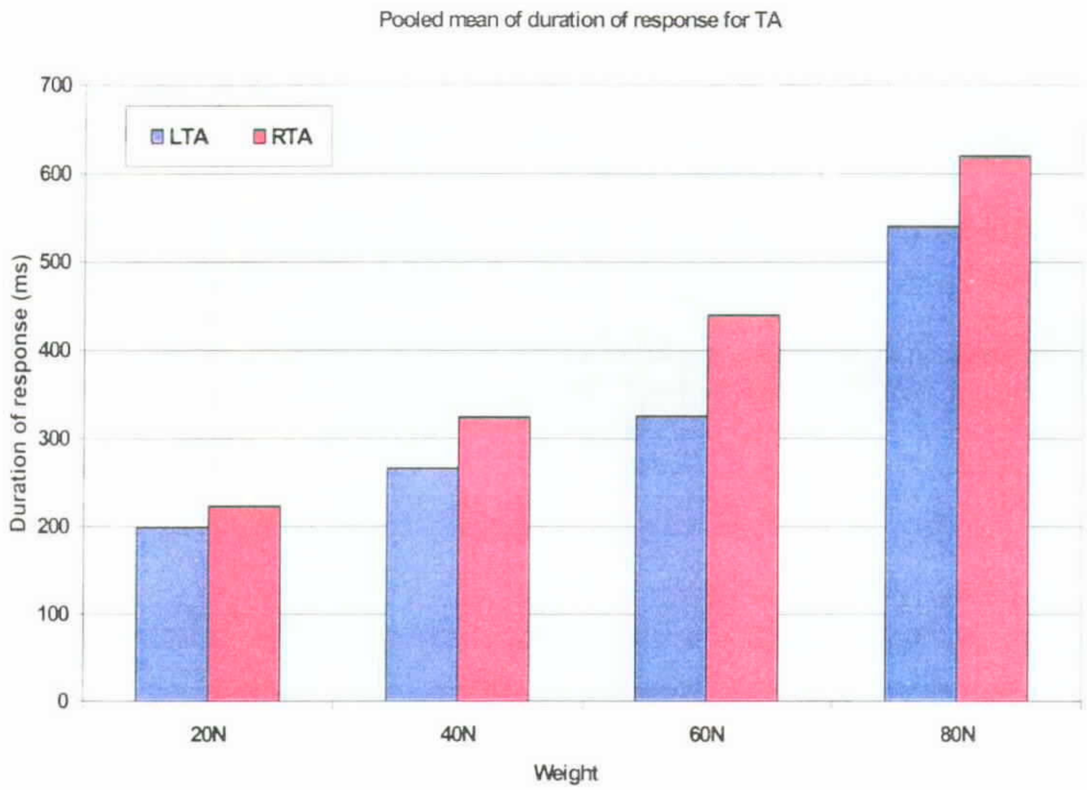


Figure 4.20 Pooled mean of duration of response for TA with different lifting weights
Note: L=left and R=right

For all other muscles (EO, IO, LD, RA, RF, ES, BF and G), the interactions between the three left within-subject factors (posture, weight and side) were still statistically significant and therefore they were analysed according to lifting weights. Data were analysed according to side (left and right) if significant interaction was still found between the within-subject factors (posture and side) (Table 4.19).

Table 4.19 Muscle duration of response comparison using repeated measures ANOVA with two within-subject factors (posture and side) and contrast tests comparing the duration of response among the three levels of posture

<table>
<tr><th colspan="4">Repeated measures ANOVA</th><th colspan="3">Contrast tests comparing the duration of response among the three levels of posture</th></tr>
<tr><th>Muscles</th><th>Weight</th><th>Within-subject factors</th><th>P value</th><th>sq-a</th><th>st-a</th><th>sq-st</th></tr>
<tr><td rowspan="8">EO</td><td rowspan="2">20N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.029*</td><td>0.001*</td></tr>
<tr><td>Side</td><td>0.116</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">40N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.914</td><td>0.001*</td></tr>
<tr><td>Side</td><td>0.105</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">60N</td><td>Posture</td><td>0.001*</td><td>0.508</td><td>0.001*</td><td>0.001*</td></tr>
<tr><td>Side</td><td>0.066</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">80N</td><td>Posture</td><td>0.076</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Side</td><td>0.157</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="8">IO</td><td rowspan="2">20N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.001*</td></tr>
<tr><td>Side</td><td>0.357</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">40N</td><td>Left</td><td>0.982</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Right</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.586</td></tr>
<tr><td rowspan="2">60N</td><td>Left</td><td>0.095</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Right</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.391</td></tr>
<tr><td rowspan="2">80N</td><td>Left</td><td>0.147</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Right</td><td>0.003*</td><td>0.002*</td><td>0.065</td><td>0.094</td></tr>
<tr><td rowspan="8">LD</td><td rowspan="2">20N</td><td>Left</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.001*</td></tr>
<tr><td>Right</td><td>0.001*</td><td>0.187</td><td>0.001*</td><td>0.001*</td></tr>
<tr><td rowspan="2">40N</td><td>Posture</td><td>0.592</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Side</td><td>0.155</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">60N</td><td>Posture</td><td>0.009*</td><td>0.005*</td><td>0.005*</td><td>0.874</td></tr>
<tr><td>Side</td><td>0.921</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">80N</td><td>Posture</td><td>0.019*</td><td>0.001*</td><td>0.016*</td><td>0.313</td></tr>
<tr><td>Side</td><td>0.678</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="8">RA</td><td rowspan="2">20N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.001*</td></tr>
<tr><td>Side</td><td>0.109</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">40N</td><td>Left</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.184</td></tr>
<tr><td>Right</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.06</td></tr>
<tr><td rowspan="2">60N</td><td>Posture</td><td>0.023*</td><td>0.586</td><td>0.01*</td><td>0.014*</td></tr>
<tr><td>Side</td><td>0.79</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">80N</td><td>Left</td><td>0.117</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Right</td><td>0.026*</td><td>0.01*</td><td>0.001*</td><td>0.92</td></tr>
<tr><td rowspan="8">RF</td><td rowspan="2">20N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.248</td></tr>
<tr><td>Side</td><td>0.587</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">40N</td><td>Posture</td><td>0.384</td><td colspan="3">Effect was not significant</td></tr>
<tr><td>Side</td><td>0.073</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">60N</td><td>Posture</td><td>0.030*</td><td>0.001*</td><td>0.26</td><td>0.223</td></tr>
<tr><td>Side</td><td>0.61</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="2">80N</td><td>Posture</td><td>0.002*</td><td>0.001*</td><td>0.02*</td><td>0.994</td></tr>
<tr><td>Side</td><td>0.899</td><td colspan="3">Effect was not significant</td></tr>
<tr><td rowspan="4"></td><td rowspan="2">20N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.001*</td></tr>
<tr><td>Side</td><td>0.004*</td><td colspan="3">L was significantly different from R</td></tr>
<tr><td rowspan="2">40N</td><td>Posture</td><td>0.001*</td><td>0.001*</td><td>0.001*</td><td>0.121</td></tr>
<tr><td>Side</td><td>0.068</td><td colspan="3">Effect was not significant</td></tr>
</table>

ES	60N	Left	0.038*	0.299	0.029*	0.098
		Right	0.053	Effect was not significant		
	80N	Posture	0.087	Effect was not significant		
		Side	0.26	Effect was not significant		
BF	20N	Posture	0.001*	0.002*	0.001*	0.001*
		Side	0.573	Effect was not significant		
	40N	Left	0.009*	0.231	0.062	0.009*
		Right	0.197	Effect was not significant		
	60N	Posture	0.003*	0.159	0.066	0.001*
		Side	0.43	Effect was not significant		
	80N	Posture	0.186	Effect was not significant		
		Side	0.571	Effect was not significant		
G	20N	Posture	0.173	Effect was not significant		
		Side	0.175	Effect was not significant		
	40N	Left	0.002*	0.001*	0.013*	0.944
		Right	0.046*	0.035*	0.869	0.078
	60N	Posture	0.375	Effect was not significant		
		Side	0.477	Effect was not significant		
	80N	Posture	0.527	Effect was not significant		
		Side	0.5	Effect was not significant		
*for p<0.05						

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For EO, the main effect of side was not significant for all lifting weights but the main effect of posture was significant for all lifting weights except for 80N (p=0.076). The pooled mean of duration of response of EO (Figure 4.21) shows that at lifting weights of 20N and 40N, the duration of response of symmetric squat lifting was significantly longer than symmetric stoop and asymmetric stoop lifting (both p=0.001). The duration of response of symmetric stoop lifting was also significantly longer than asymmetric stoop lifting (p=0.029) for 20N load. At the lifting weight of 60N, both the duration of response of symmetric squat lifting and asymmetric stoop lifting were significantly longer than symmetric stoop lifting (p=0.001). The duration of response of symmetric squat lifting was longer than asymmetric stoop lifting, however, the difference was not significant (p=0.508). For lifting weight of 80N, the duration of response of asymmetric stoop lifting was longer than symmetric squat and stoop lifting, but again this was not statistically significant since both posture and side had no significant effect on the duration of response (p=0.076 and p=0.157, respectively).

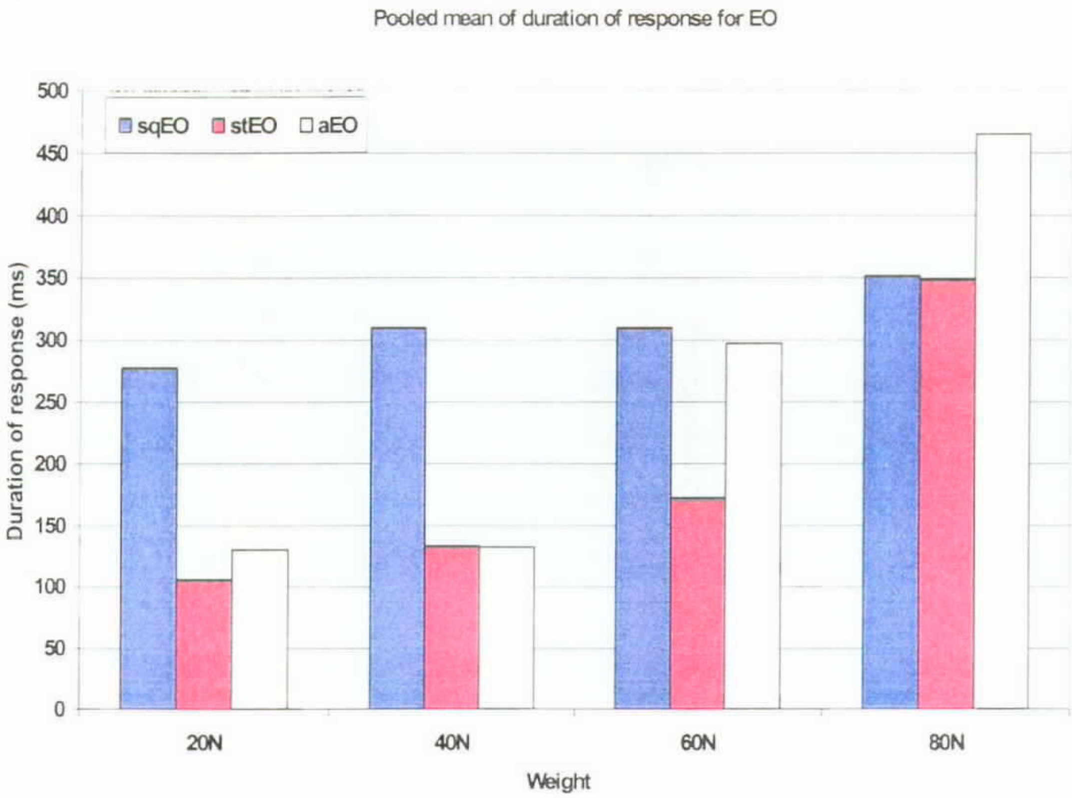


Figure 4.21 Pooled mean of duration of response for EO with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For IO, the main effect of posture was significant (p=0.001) for lifting weight of 20N but the main effect of side was not significant (p=0.357). The interaction between the two left within-subject factors (posture and side) was still statistically significant (p=0.001) for lifting weights of 40N and 60N and for lifting weight of 80N (p=0.003) also. Therefore, the data were analysed according to side. Posture was found to have significant effect on the duration of response for right IO with p=0.001 for 40N and 60N and p=0.003 for 80N. However, posture was found to have no significant effect on the duration of response for left IO (p=0.982, 0.095 and 0.147 for 40N, 60N and 80N, respectively). Contrast tests were performed to compare the three levels of posture within right IO. The pooled mean of duration of response of IO was plotted (Figure 4.22):

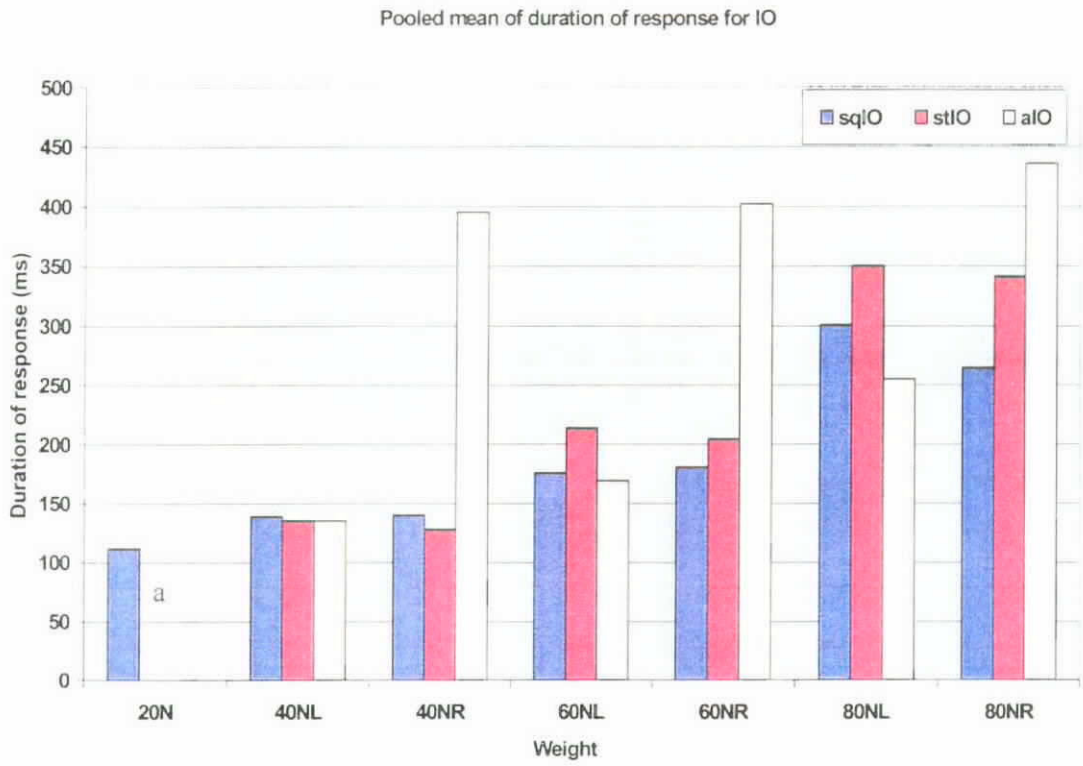


Figure 4.22 Pooled mean of duration of response for IO with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right
a= IO has no response to sudden release of load at the weight of 20N when symmetric stoop lift or asymmetric stoop lift was adopted

At the lifting weight of 20N, IO has no response to the sudden release of load when symmetric stoop or asymmetric stoop lifting posture was adopted. The duration of response of symmetric squat lifting was significantly different from symmetric stoop and asymmetric stoop lifting (both p=0.001). A trend of increasing duration of response increased with increasing lifting weight was observed from 40N to 80N load for all lifting postures. At lifting weights of 40N and 60N, the duration of response of right IO was significantly longer for asymmetric stoop lifting than symmetric stoop and squat lifting (both p=0.001), but the duration of response of symmetric squat lifting was not significantly different from symmetric stoop lifting (p=0.586 for 40N and p=0.391 for 60N). At the lifting weight of 80N, the right IO duration of response of asymmetric stoop lifting was significantly longer than symmetric squat lifting (p=0.002) but not significantly longer than symmetric stoop lifting (p=0.065). The duration of response of symmetric stoop lifting was also longer than symmetric squat lifting, but not significantly (p=0.094).

For LD, side had no significant effect on the duration of response at lifting weights of 40N, 60N and 80N (p=0.155, 0.921 and 0.678). Posture had no significant effect on the duration of response at the lifting weight of 40N (p=0.592), however, it had a significant effect at the lifting weight of 60N and 80N (p=0.009 and p=0.019, respectively). The interaction between the two left within-subject factors (posture and side) was still statistically significant with p=0.034 for lifting weight of 20N. Therefore the data were analysed according to side. Posture was found to have significant effect on the duration of response for both left and right LD with p=0.001. The pooled mean of duration of response for LD (Figure 4.23) shows that at the lifting weight of 20N, LD has no response to the sudden release of load when the symmetric stoop lift was adopted. The duration of response of symmetric squat lifting was significantly longer than asymmetric stoop lifting (p=0.001) for left LD but not significantly for right LD (p=0.187). For lifting weights of 60N and 80N, both the duration of response of symmetric squat lifting and symmetric stoop lifting were significantly longer than asymmetric stoop lift (p=0.005 for 60N and p=0.001 and p=0.016 for 80N, respectively). For symmetric squat and stoop lifting, a trend of increasing duration of response with increasing lifting weight was found.

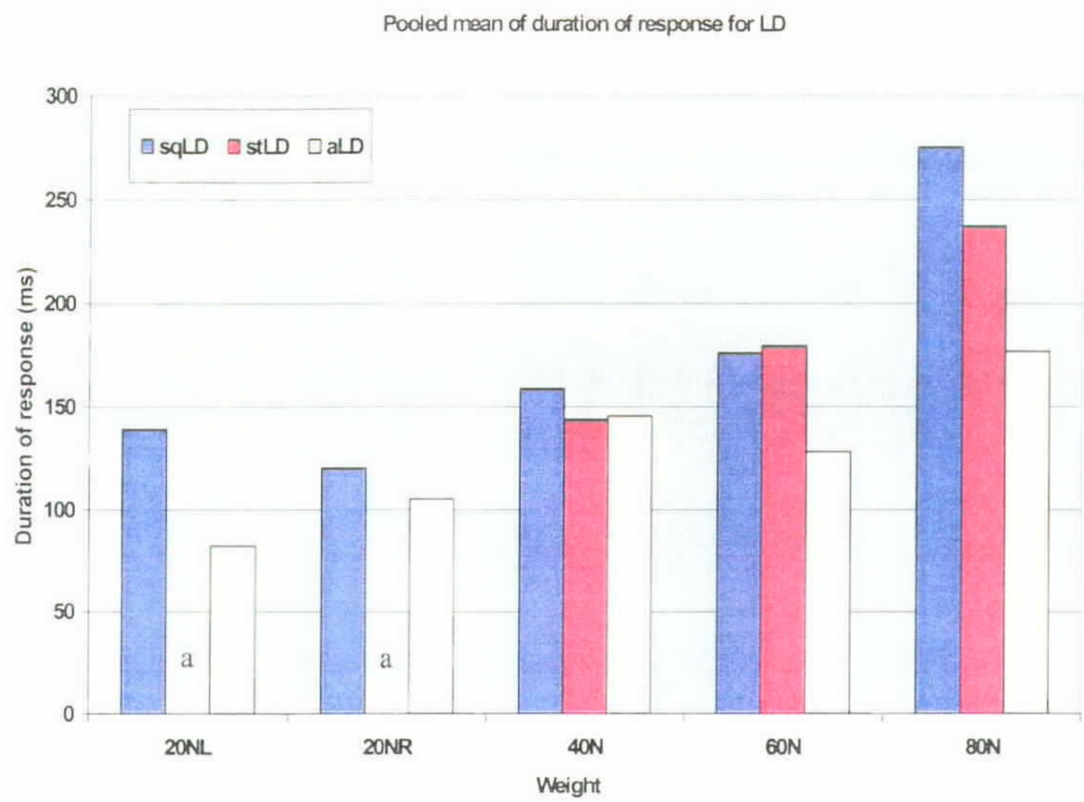


Figure 4.23 Pooled mean of duration of response for LD with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift
a=LD has no response to the sudden release of load at the weight of 20N when symmetric stoop lift was adopted

For RA, the main effect of posture was significant with p=0.001 and p=0.023 for lifting weights of 20N and 60N, respectively while the main effect of side was not significant (p=0.109 and p=0.79 for 20N and 60N, respectively). The interaction between the two left within-subject factors (posture and side) was still statistically significant for lifting weights of 40N and 80N with p=0.013 and p=0.002. Therefore the data were analysed according to side. Posture was found to have significant effect on the duration of response for both left and right RA (p=0.001) at the lifting weight of 40N, and for right RA (p=0.026) at the lifting weight of 80N. The pooled mean of duration of response for RA is plotted in figure 4.24.

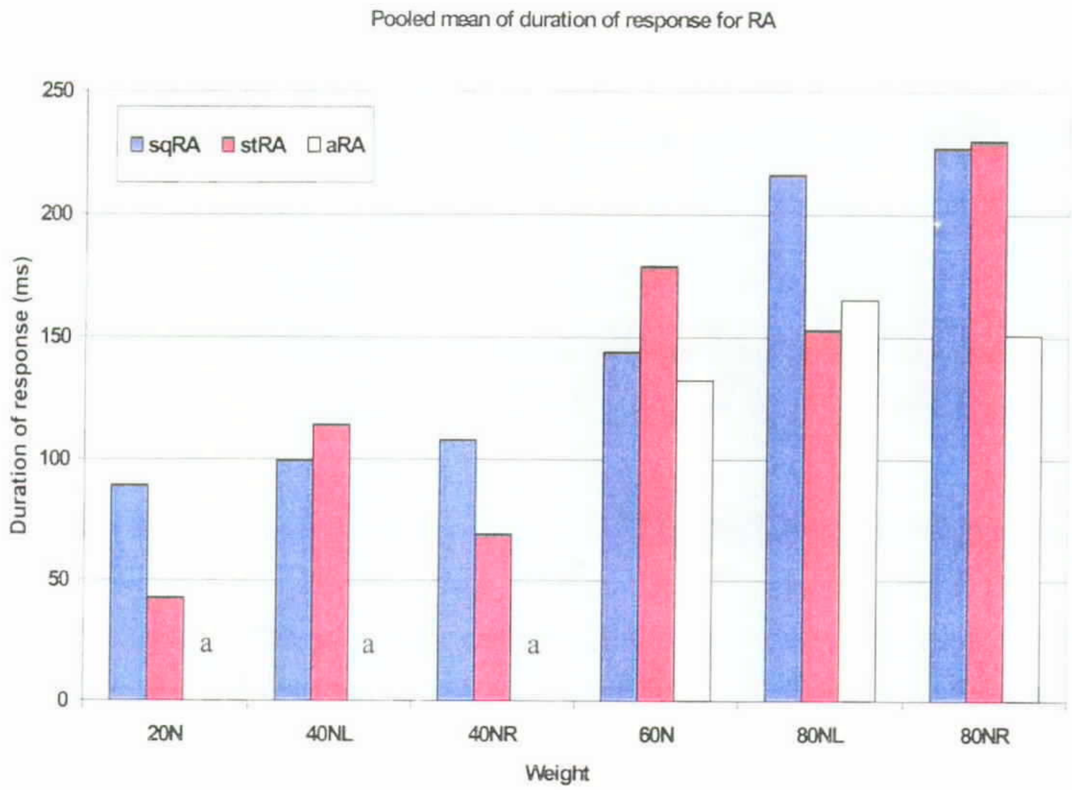


Figure 4.24 Pooled mean of duration of response for RA with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right a=RA has no response to the sudden release of load at the weight of 20N and 40N when asymmetric stoop lift was adopted

At the lifting weights of 20N and 40N, RA showed no response to the sudden release of load when the asymmetric stoop lift was adopted. At the lifting weight of 20N, the duration of response for symmetric squat lifting was significantly longer than symmetric stoop lifting (p=0.001). The duration of response for symmetric squat lifting was also longer than symmetric stoop lifting for the right RA at the lifting weight of 40N, however, the difference was not significant (p=0.06). In contrast to the right RA, the duration of response for symmetric squat lifting was shorter than symmetric stoop lifting for left RA, although the difference was also not significant (p=0.184). At the lifting weight of 60N, the duration of response for symmetric stoop lifting was found to be significantly longer than symmetric squat and asymmetric stoop lifting (p=0.014 and p=0.001, respectively). For the lifting weight of 80N, posture had no significant effect on duration of response for left RA (p=0.117). The right RA had shorter duration of response for asymmetric stoop lifting than symmetric squat and symmetric stoop lifting with p=0.04 and p=0.001 respectively. A trend of increasing duration of response with increasing lifting weight was found for symmetric squat lifting.

For RF, side had no significant effect on the duration of response for all lifting weights but posture had no significant effect on the duration of response for 40N only (p=0.384). The pooled mean of duration of response for RF was plotted (Figure 4.25):

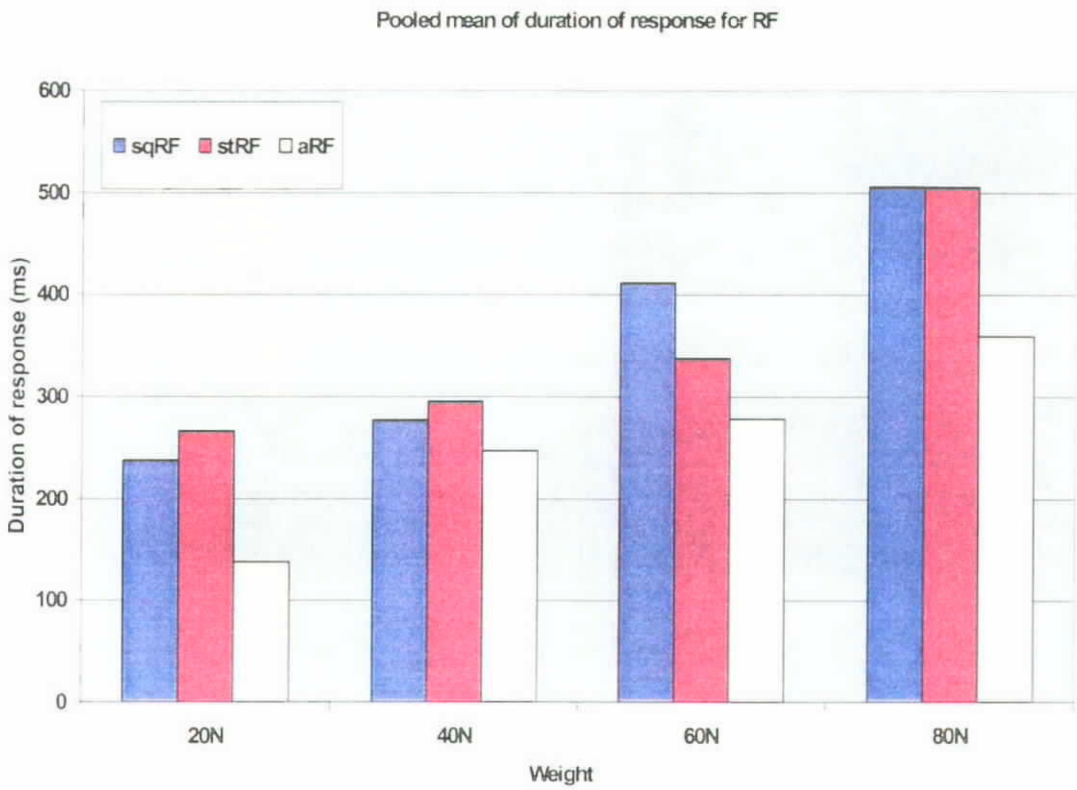


Figure 4.25 Pooled mean of duration of response for RF with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For both 20N and 80N, the duration of response of symmetric squat and symmetric stoop lifting were longer than asymmetric stoop lifting with both p=0.001 for 20N and p=0.001 and p=0.02 respectively for 80N. At the lifting weight of 40N, both posture and side had no significant effect on the duration of response (p=0.384 and p=0.073, respectively). At the lifting weight of 60N, symmetric squat lifting had a longer duration of response than symmetric and asymmetric stoop lifting, but his was insignificant (p=0.223) for symmetric stoop lifting, and significant (p=0.001) for asymmetric stoop lifting. The duration of response for symmetric stoop lifting was also longer than asymmetric stoop lifting, however, the difference was insignificant (p=0.26). A trend of increasing duration of response as the lifting weight increased was found for all lifting postures. In addition, the duration of response of symmetric squat and symmetric stoop lifting were longer than asymmetric stoop lifting for all lifting weights.

For ES, side had significant effect on the duration of response at the lifting weight of 20N (p=0.004) but not significant at the lifting weights of 40N and 80N (p=0.068 and 0.26, respectively). The main effect of posture was significant for the lifting weights of 20N and 40N (both p=0.001) and not significant at the lifting weigh of 80N (p=0.087). The interaction between the two left within-subject factors (posture and side) was still statistically significant for the lifting weight of 60N (p=0.008). Therefore, the data were analysed according to side. Posture was found to have significant effect on the duration of response for left ES (p=0.038) but not significant (p=0.053) for right ES. The pooled mean of duration of response for ES (Figure 4.26) shows that at lifting weight of 20N, symmetric stoop lifting had a significantly longer duration of response than symmetric squat and asymmetric stoop lifting (both p=0.001), and symmetric squat lifting also had significantly longer duration of response than asymmetric stoop lifting (p=0.001). It was also found that the duration of response was longer for left ES than right ES (p=0.004) for all lifting postures. At the lifting weight of 40N, symmetric squat and symmetric stoop lifting had a longer duration of response than asymmetric stoop lifting (both p=0.001), and symmetric stoop was also found to have longer duration of response than symmetric squat lifting, but the difference was not significant (p=0.121). At the lifting weight of 60N, left ES also had a longer duration of response for symmetric stoop lifting than symmetric squat and asymmetric stoop lifting, which was significant (p=0.029) for symmetric squat lifting but insignificant (p=0.098) for symmetric stoop lifting. Both posture and side had no significant effect on the duration of response at the lifting weight of 80N (p=0.087 and p=0.26, respectively). A trend of increasing duration of response with increasing lifting weight was seen for symmetric squat and asymmetric stoop lifting.

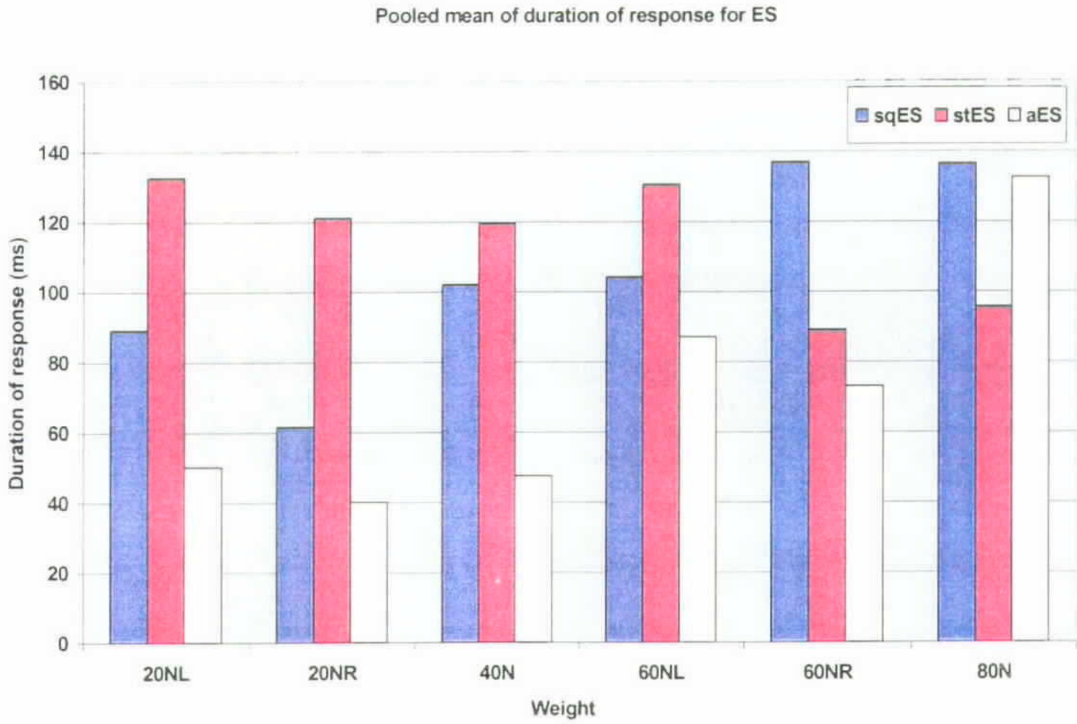


Figure 4.26 Pooled mean of duration of response for ES with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

For BF, the main effect of posture was significant for 20N and 60N (p=0.001 and 0.003, respectively). The main effect of side was not significant for weights of 20N, 60N and 80N (p=0.573, 0.43 and 0.571, respectively). The interaction between the two left within-subject factors (posture and side) was still statistically significant for the lifting weight of 40N (p=0.042) and therefore the data were analysed according to side. Posture was found to have a significant effect on the duration of response for left BF (p=0.009) but this was not significant (p=0.197) for right BF. The pooled mean of duration of response for BF was plotted (Figure 4.27):

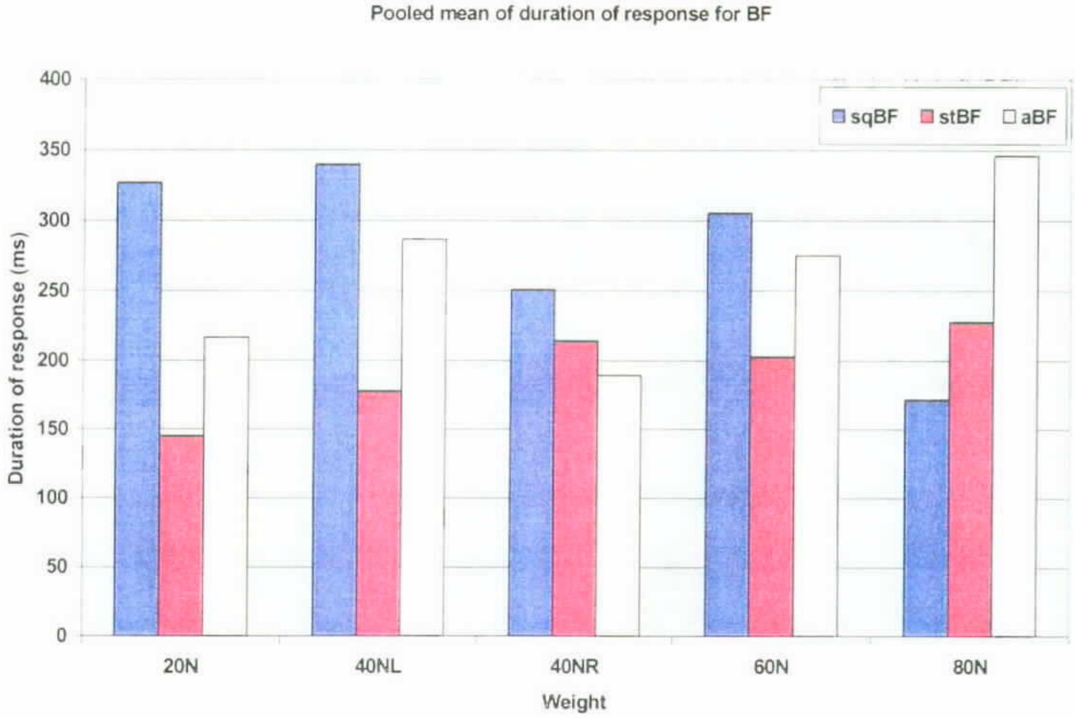


Figure 4.27 Pooled mean of duration of response for BF with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

The duration of response of symmetric stoop lifting was shorter than symmetric squat and asymmetric stoop lifting at lifting weights of 20N and 60N, however, this difference was significant for 20N (p=0.002 and p=0.001, respectively) but insignificant for 60N (p=0.159 and 0.066, respectively). For both 20N and 60N, the duration of response for symmetric squat lifting was significantly longer than symmetric stoop lifting (p=0.001). At the lifting weight of 40N, posture had significant effect on the duration of response for left BF (p=0.009) but no significant effect (p=0.197) for right BF. Similar to the results at 20N and 60N, at 40N, left BF had the shortest duration of response when symmetric stoop lifting was adopted. However, the response duration of symmetric stoop lifting was not significantly shorter than the symmetric squat and asymmetric stoop lifting (p=0.231 and p=0.062, respectively). Both posture and side had no significant effect on the duration of response (p=0.186 and p=0.571, respectively). A trend of increasing duration of response increased with increasing lifting weight was found for the symmetric squat lifting posture.

For G, both posture and side had no significant effect on the duration of response with p=0.173 and p=0.175 respectively for 20N, p=0.375 and p=0.477 respectively for 60N and p=0.527 and p=0.5 respectively for 80N. The interaction between the two left within-subject factors (posture and side) was still statistically significant for the lifting weight of 40N (p=0.001) and therefore the data were analysed according to side. Posture was found to have a significant effect on the duration of response for left and right G (p=0.002 and p=0.046, respectively). The pooled mean of duration of response for G (Figure 4.28) shows that at the lifting weight of 40N, the duration of asymmetric stoop lifting was significantly longer than the symmetric squat and symmetric stoop lifting with p=0.001 and p=0.013 for left G. For right G, the duration of response of symmetric squat lifting was longer than the symmetric stoop and asymmetric stoop lifting, however, this was only significant (p=0.035) for asymmetric stoop lifting.

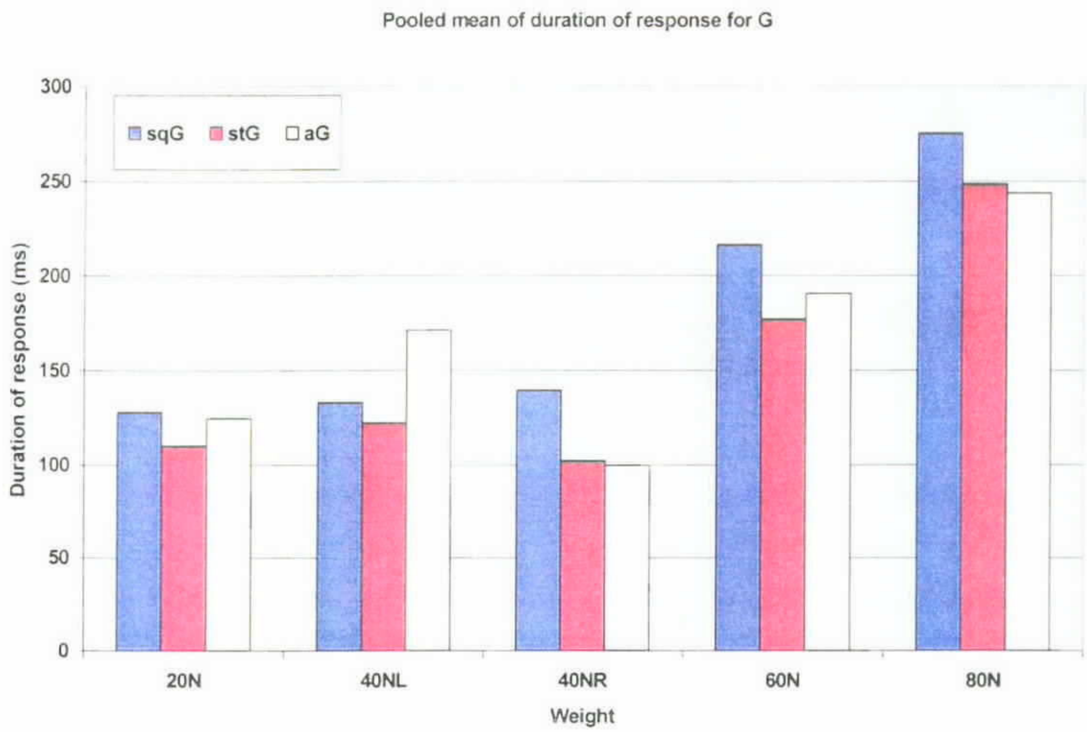


Figure 4.28 Pooled mean of duration of response for G with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

With the exception of a few exceptional cases, posture had no significant effect on the duration of response for TA, and EO had the shortest duration of response when symmetric stoop lift was adopted for all lifting weights. LD, RF and ES had the shortest duration of response when the asymmetric stoop lifting posture was adopted for all lifting weights. For BF and G, the duration of response was found to be the longest when symmetric squat lifting was adopted at the lifting weights of 20N, 40N and 60N for BF and weights of 20N, 60N and 80N for G. A trend of increasing duration of response with increase in lifting weight was found for TA, EO, IO and RF in all lifting postures. The duration of response of LD and RA increased with the lifting weight only when symmetric squat or symmetric stoop lifting was adopted. For ES, the duration of response increased as the lifting weight increased only when symmetric squat or asymmetric stoop lifting was adopted. The duration of response increased with the lifting weight when symmetric stoop lifting was adopted for BF and symmetric squat lifting for G. Side had no significant effect on the duration of response for EO and RA for all lifting weights and postures. The duration of response of right TA was found to be longer than left TA for all lifting weights and postures. Side was also found to have no significant effect on the duration of response for LD (40N, 60N and 80N), BF (20N, 60N and 80N), G (20N, 60N and 80N), RA (20N and 60N) and ES (40N and 80N).

4.2.3 Co-contraction duration

There were five co-contraction muscle couples identified, namely ES-EO (Erector Spinae-External Oblique), ES-IO (Erector Spinae-Internal Oblique), ES-RA (Erector Spinae-Rectus Abdominis), BF-RF (Biceps Femoris-Rectus Femoris) and G-TA (Gastronemius-Tibialis Anterior). The co-contraction duration was defined as the duration of simultaneous contraction of the two muscles before the relaxation response of the posterior muscles. The mean and standard deviation of co-contraction duration were determined for each co-contraction couples for all three lifting postures (Table 4.20-4.22 and Figure 4.29-4.31).

Table 4.20 Mean (standard deviation) of co-contraction duration of muscles couples for symmetric squat lift

Muscles couples	20N	40N	60N	80N
LES-EO	50.0 (29.3)	57.2(33.4)	80.1(45.8)	75.0(15.4)
RES-EO	50.6(26.5)	65.5(26.9)	73.3(33.2)	74.4(24.7)
LES-IO	51.8(30.1)	65.8(33.1)	70.7(36.1)	86.1(26.9)
RES-IO	49.5(22.4)	67.5(36.9)	80.3(15.4)	82.2(45.1)
LES-RA	20.0(11.6)	47.2(11.5)	56.3(23.1)	76.8(12.9)
RES-RA	23.6(15.3)	46.3(13.2)	58.3(22.1)	67.8(10.0)
LBF-RF	96.4(53.9)	100.5(33.1)	129.7(53.1)	126.4(22.7)
RBF-RF	93.8(66.1)	94.0(22.0)	117.4(34.0)	118.4(56.8)
LG-TA	203.7(102.5)	198.8(98.7)	226.2(89.4)	236.9(139.7)
RG-TA	195.9(115.3)	185.4(130.8)	219.4(109.1)	235.8(140.7)

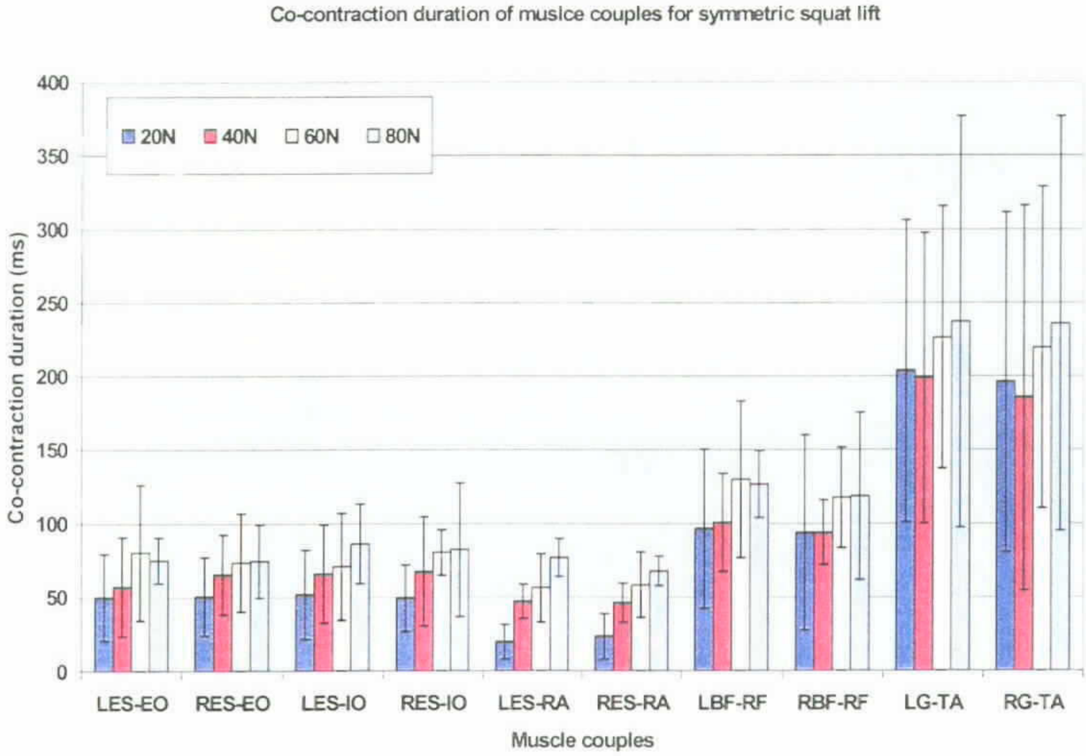


Figure 4.29 Co-contraction duration of muscle couples for symmetric squat lifting with different lifting weights

Table 4.21 Mean (standard deviation) of co-contraction duration of muscles couples for symmetric stoop lift

Muscles couples	20N	40N	60N	80N
LES-EO	NC	NC	12.8(3.4)	32.7(15.2)
RES-EO	NC	NC	27.2(11.4)	31.3(12.6)
LES-IO	NA	NC	2.0(1.3)	23.9(15.4)
RES-IO	NA	NC	25.3(14.2)	27.0(16.7)
LES-RA	NC	NC	6.5(3.3)	5.7(3.1)
RES-RA	NC	NC	8.8(4.1)	14.3(8.7)
LBF-RF	54.1(33.4)	71.5(23.1)	78.1(47.0)	103.2(22.1)
RBF-RF	53.3(23.4)	58.6(15.9)	89.8(19.8)	111.0(45.8)
LG-TA	113.7(25.9)	112.8(33.8)	156.8(46.9)	237.3(105.2)
RG-TA	108.4(12.4)	112.7(23.8)	143.5(47.9)	226.4(59.8)

Note: NC=No Co-contraction; NA=Not Applicable since one of the muscle in the muscle couples showed no response for that particular lifting weight

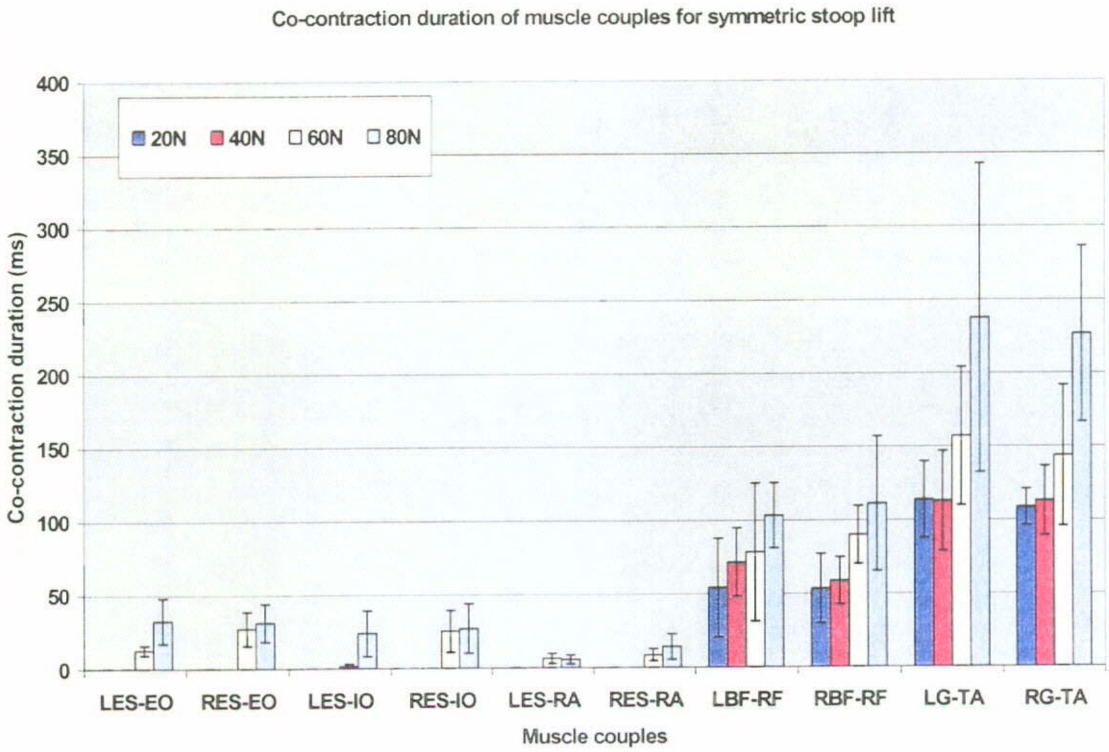


Figure 4.30 Co-contraction duration of muscle couples for symmetric stoop lift with different lifting weights

Table 4.22 Mean (standard deviation) of co-contraction duration of muscles couples for asymmetric stoop lift

Muscles couples	20N	40N	60N	80N
LES-EO	NC	6.3(3.7)	4.0(2.8)	17.8(6.8)
RES-EO	NC	14(5.4)	33.5(3.9)	51.8(22.1)
LES-IO	NA	NC	NC	NC
RES-IO	NA	NC	15.5(5.9)	39.8(6.8)
LES-RA	NA	NA	5.2(3.3)	11.7(7.8)
RES-RA	NA	NA	3.8(2.1)	32.8(6.9)
LBF-RF	24.7(12.5)	45.8(23.7)	32.0(21.0)	49.3(11.0)
RBF-RF	71.5(23.4)	73.2(33.7)	64.0(25.8)	70.1(30.1)
LG-TA	68.4(25.6)	54.5(24.8)	61.4(21.9)	77.8(20.7)
RG-TA	124.9(59.6)	125.2(69.7)	187.8(83.4)	245.6(103.4)

Note: NC=No Co-contraction; NA=Not Applicable since one of the muscle in the muscle couples showed no response for that particular lifting weight

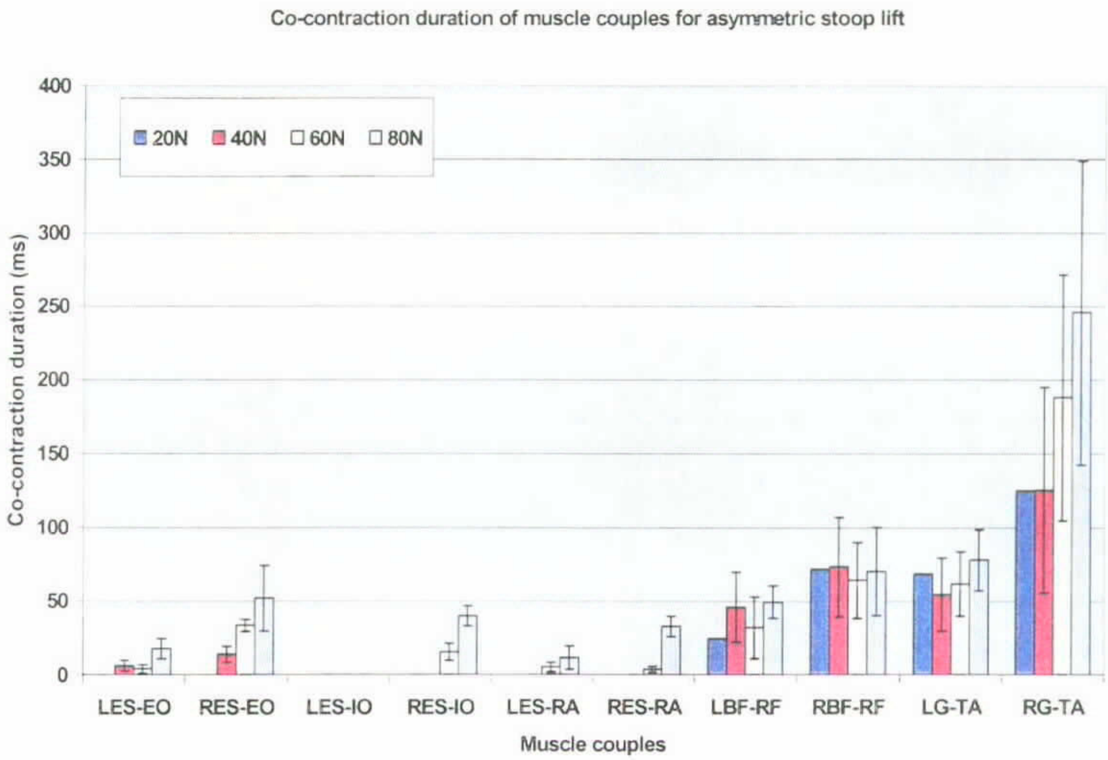


Figure 4.31 Co-contraction duration of muscle couples for asymmetric stoop lifting with different lifting weights

BF-RF and G-TA were the two muscle couples that had the longest co-contraction duration among all muscle couples for all lifting postures. The co-contraction duration of left and right side were similar for both symmetric squat and symmetric stoop lifting. For asymmetric stoop lifting, the right side had a longer co-contraction duration than the left side for all muscle couples. Generally, symmetric squat lifting had a longer co-contraction duration than symmetric stoop lifting, and symmetric stoop lifting had longer co-contraction duration than asymmetric stoop lifting for muscle couples BF-RF and G-TA. For ES-EO, ES-IO and ES-RA, symmetric squat lifting had longer co-contraction duration than symmetric and asymmetric stoop lifting. Also, a trend of increasing co-contraction duration with increasing lifting weight was found for all lifting postures and muscle couples.

Repeated measures ANOVA was conducted to compare the co-contraction duration among different conditions. The analyses included four within-subject factors; muscle couples, lifting weights, lifting postures and sides. There were five levels of muscle couple (ES-EO, ES-IO, ES-RA, BF-RF and G-TA), four levels of lifting weight, three levels of lifting posture and two levels of side. It was found that there were significant interactions between the within-subject factors and therefore the data were analysed according to the muscle couples. For ES-EO and ES-IO, the interactions between the left three within-subject factors (posture, weight and side) were not significant but the main effect of posture and weight were significant for both muscle couples with both $p=0.001$ for ES-EO, and $p=0.001$ and $p=0.002$ respectively for ES-IO. The main effect of side was only found to be significant with $p=0.03$ for ES-EO. Contrast tests were performed to compare the co-contraction duration of the three levels of posture and four levels of weight (Tables 4.23-4.24). The co-contraction duration of ES-EO and pooled mean of co-contraction duration of ES-IO was plotted (Figures 4.32-4.33).

Table 4.23 Contrast tests comparing the co-contraction duration among the three postures for muscle couples ES-EO and ES-IO

Muscle Couples	Contrast tests comparing the latency among the three levels of posture		
	Symmetric squat and asymmetric stoop	Symmetric stoop and asymmetric stoop	Symmetric squat and symmetric stoop
ES-EO	0.001*	0.258	0.001*
ES-IO	0.001*	0.007*	0.001*
* for $p<0.05$			

Table 4.24 Contrast tests comparing the co-contraction duration among the four weights for muscle couples ES-EO and ES-IO

ES-EO	20N	40N	60N	80N	ES-IO	20N	40N	60N	80N
20N					20N				
40N	0.194				40N	0.061			
60N	0.003*	0.041*			60N	0.001*	0.019*		
80N	0.002*	0.001*	0.278		80N	0.001*	0.011*	0.126	
* for p < 0.05									

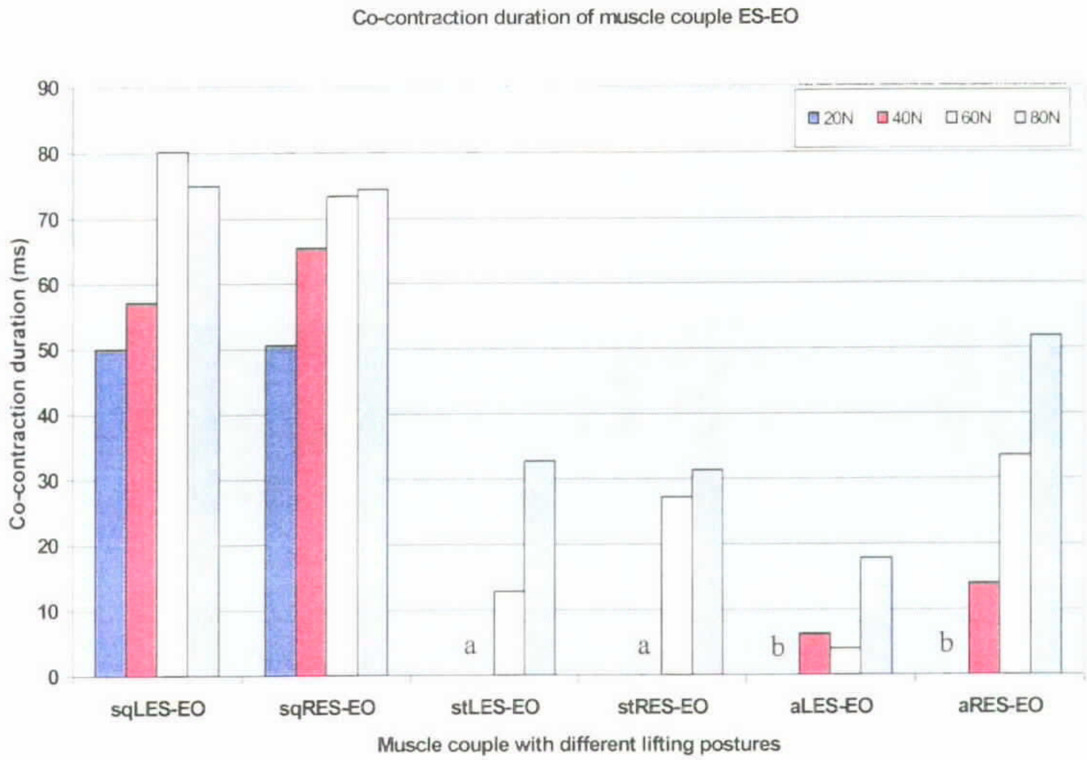


Figure 4.32 Co-contraction duration of muscle couple ES-EO with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

a=There was no co-contraction since ES relaxed before EO started to contract at the lifting weights of 20N and 40N for symmetric stoop lifting

b=There was no co-contraction since ES relaxed before EO started to contract at the lifting weight of 20N for asymmetric stoop lifting

It was found that the co-contraction duration was significantly longer for the ES-EO muscle couple for symmetric squat lifting than symmetric stoop and asymmetric stoop lifting (both p=0.001). For all lifting postures, the co-contraction duration was significantly longer at the lifting weight of 80N than 40N and 20N (p=0.001 and p=0.002, respectively). The co-contraction duration was also significantly longer at the lifting weight of 60N than 40N and 20N (p=0.041 and p=0.003, respectively). Generally, there was a trend whereby the co-contraction duration increased as the lifting weight increased for all lifting postures.

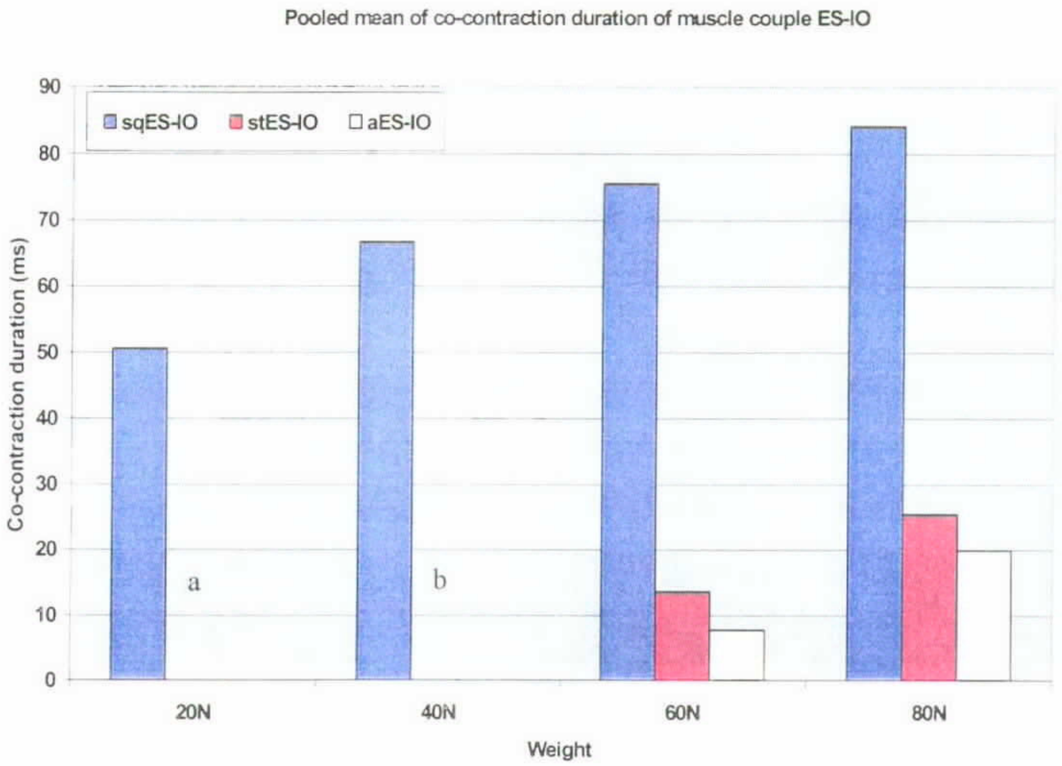


Figure 4.33 Pooled mean of co-contraction duration of muscle couple ES-IO with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift
a=There was no co-contraction since IO has no response to the sudden release of load at the lifting weight of 20N for symmetric stoop or asymmetric stoop lifting
b=There was no co-contraction since ES relaxed before IO started to contract at the lifting weight of 40N for asymmetric stoop or asymmetric stoop lifting

The co-contraction duration of symmetric squat lifting was found to be significantly longer than symmetric stoop and asymmetric stoop lifting (both p=0.001) and the co-contraction duration of symmetric stoop was also found to be significantly longer than asymmetric stoop lifting (p=0.007). For all lifting postures, the longest co-contraction duration was found at the lifting weight of 80N. Similar to ES-EO, the co-contraction duration was found to be significantly longer at the lifting weight of 80N than 40N and 20N (p=0.011 and p=0.001, respectively) and again the co-contraction duration was significantly longer at the lifting weight of 60N than 40N and 20N (p=0.019 and P=0.001, respectively). A trend of increasing duration of co-contraction with increasing lifting weight was found for all lifting postures.

For muscle couple ES-RA, BF-RF and G-TA, the interactions between the three left within-subject factors (posture, weight and side) were still statistically significant and therefore the data were analysed according to lifting weights. For muscle couple G-TA, the interaction between the two left within-subject factors was still significant at the lifting weight of 80N with p=0.01 and therefore the data were analysed according to side (Table 4.25).

Table 4.25 Muscle co-contraction duration comparison using repeated measures ANOVA with two within-subject factors (posture and side) and contrast tests comparing the co-contraction duration among the three levels of posture

Repeated measures ANOVA				Contrast tests comparing the co-contraction duration among the three levels of posture		
Muscle couples	Weights	Within-subject factors	p value	sq-a	st-a	sq-st
ES-RA	20N	Posture	0.001*	0.002*	0.001*	0.002*
		Side	0.942	Effect was not significant		
	40N	Posture	0.001*	0.001*	0.001*	0.001*
		Side	0.941	Effect was not significant		
	60N	Posture	0.001*	0.001*	0.144	0.001*
		Side	0.843	Effect was not significant		
	80N	Posture	0.001*	0.001*	0.024	0.001*
		Side	0.338	Effect was not significant		
BF-RF	20N	Posture	0.006*	0.001*	0.523	0.022*
		Side	0.086	Effect was not significant		
	40N	Posture	0.022*	0.028*	0.506	0.073
		Side	0.766	Effect was not significant		
	60N	Posture	0.004*	0.001*	0.001*	0.074
		Side	0.271	Effect was not significant		
	80N	Posture	0.001*	0.001*	0.001*	0.36
		Side	0.706	Effect was not significant		
G-TA	20N	Posture	0.001*	0.001*	0.292	0.006*
		Side	0.294	Effect was not significant		
	40N	Posture	0.001*	0.001*	0.137	0.001*
		Side	0.232	Effect was not significant		
	60N	Posture	0.005*	0.005*	0.35	0.032*
		Side	0.084	Effect was not significant		
	80N	Left	0.001*	0.003*	0.003*	0.99
		Right	0.917	Effect was not significant		
* for p < 0.05						

For muscle couples ES-RA and BF-RF, the main effect of posture was significant, but the main effect of side was not significant for all lifting weights. Pooled mean of co-contraction duration for ES-RA and BF-RF were plotted (Figures 4.34-4.35):

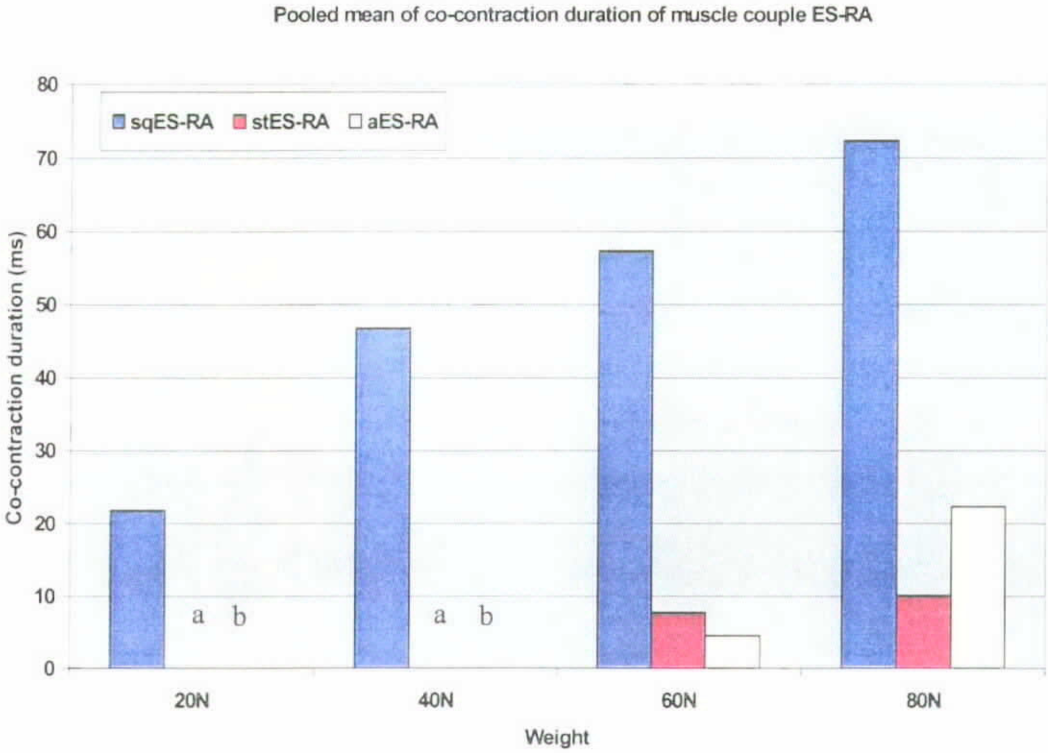


Figure 4.34 Pooled mean of co-contraction duration of muscle couple ES_RA with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift
a=There was no co-contraction since ES relaxed before RA started to contract at the lifting weights of 20N and 40N for asymmetric stoop lifting
b=There was no co-contraction since RA has no response to the sudden release of load at the lifting weights of 20N and 40N for asymmetric stoop lifting

For muscle couple ES-RA, posture had significant effect on the co-contraction duration at all lifting weights (all with p=0.001) while side had no significant effect at all lifting weights, with p=0.942, 0.941, 0.843 and 0.338 for 20, 40, 60 and 80N respectively. At the lifting weights of 20N and 40N, the co-contraction duration was found to be significantly longer for symmetric squat lifting than for symmetric and asymmetric stoop lifting (p=0.002 and p=0.001, respectively in both cases). At the lifting weights of 60N and 80N, the co-contraction duration of symmetric squat lift was significantly longer than the two other lifting postures (all with p=0.001). For all three lifting postures, there was a trend where the co-contraction duration increased as the lifting weight increased.

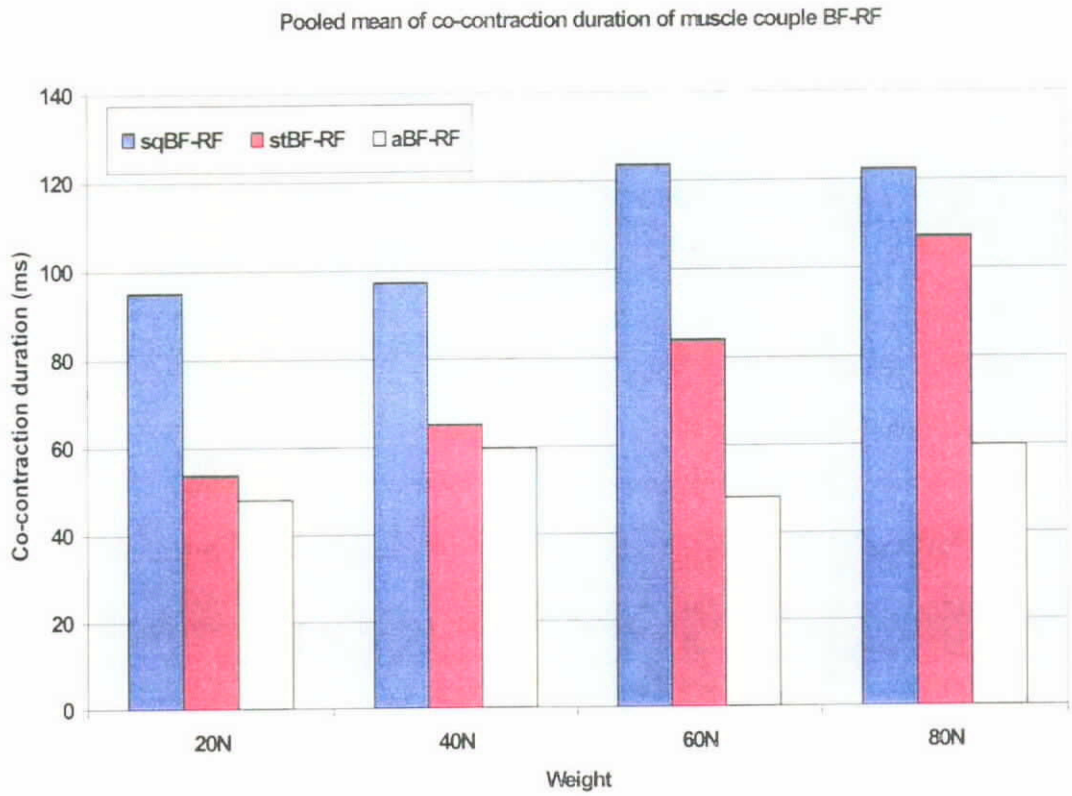


Figure 4.35 Pooled mean of co-contraction duration of muscle couple BF-RF with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift and a=asymmetric stoop lift

For muscle couple BF-RF, side has no significant effect on the co-contraction duration, while posture had a significant effect on the co-contraction duration at all lifting weights. For all lifting weights, symmetric squat lifting was found to have the longest co-contraction period followed by symmetric stoop lifting and asymmetric stoop lifting with the shortest co-contraction duration. At the lifting weight of 20N, symmetric squat lifting had significantly longer co-contraction duration than symmetric and asymmetric stoop lifting (p=0.001 and 0.022, respectively). Symmetric squat lifting had longer co-contraction duration than asymmetric stoop lifting, but the difference was not significant (p=0.523). At the lifting weight of 40N, symmetric squat lifting had a significantly longer than asymmetric stoop lift (p=0.028) but not significantly longer than that of symmetric stoop lifting ($p=0.073$). At the lifting weights of 60N and 80N, both symmetric squat lifting and symmetric stoop lift were found to have significantly longer co-contraction durations than asymmetric stoop lifting (both with $p=0.001$). The trend of increasing co-contraction duration with increase in lifting weight applied to symmetric squat and stoop lifting only.

For muscle couple G-TA (Figure 4.36) at the lifting weights of 20N, 40N and 60N symmetric squat lift had significantly longer co-contraction duration than symmetric stoop and asymmetric stoop lift with $p=0.006$ and $p=0.001$ respectively for 20N, both $p=0.001$ for 40N and $p=0.032$ and $p=0.005$ respectively for 60N). Symmetric stoop lift also had longer co-contraction duration than asymmetric stoop lift, but this was not significant, with $p=0.292$ for 20N, $p=0.137$ for 40N and $p=0.35$ for 60N. At the lifting weight of 80N, only posture had significant effect on the co-contraction duration for LG-TA with $p=0.001$. Asymmetric stoop lifting was found to have shorter co-contraction duration than the other two lifting postures (both with $p=0.003$). A trend where the co-contraction duration increased as the lifting weight increased was only found for symmetric stoop lifting.

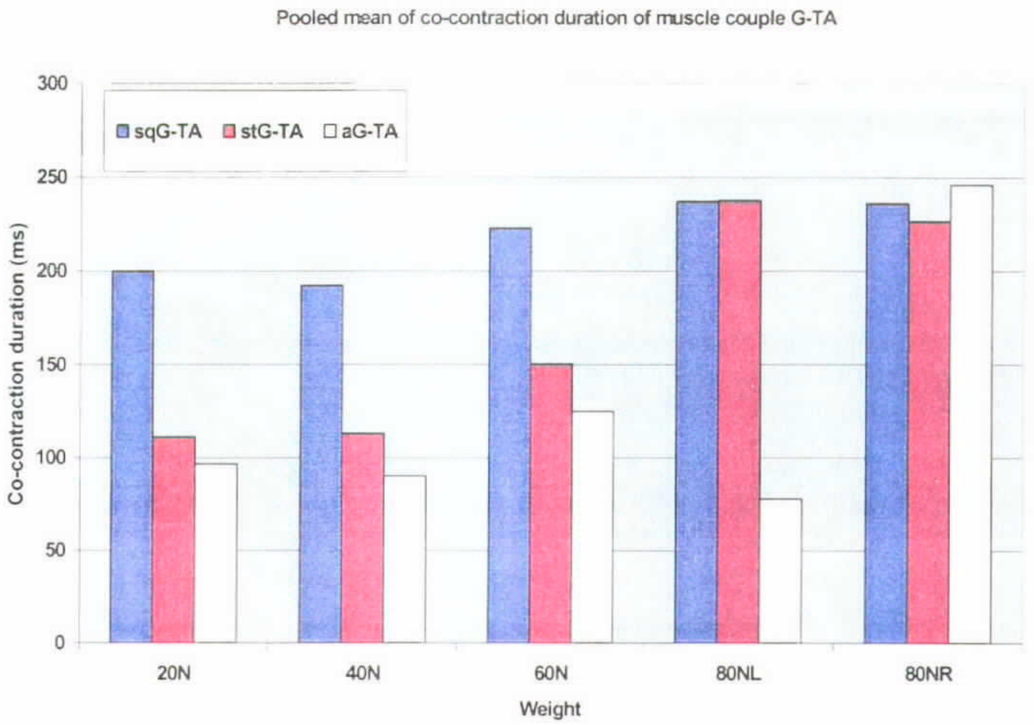


Figure 4.36 Pooled mean of co-contraction duration of muscle couple G-TA with different lifting weights and postures

Note: sq=symmetric squat lift, st=symmetric stoop lift, a=asymmetric stoop lift, L=left and R=right

Generally, symmetric squat lifting had the longest co-contraction duration compared to symmetric stoop and asymmetric stoop lifting for all muscle couples. For muscle couples BF-RF and G-TA, symmetric squat lifting had the longest co-contraction duration followed by symmetric stoop lifting and asymmetric stoop lift having the shortest co-contraction duration. For muscle couples ES-EO, ES-IO and ES-RA, there was a trend that the co-contraction duration increased as the lifting weight increased for all lifting postures. For BF-RF and G-TA, a similar trend was found in symmetric squat or stoop lifting for BF-RF and in symmetric stoop lifting only for G-TA. Side has no significant effect on the co-contraction duration for muscle couple ES-IO, ES-RA and BF-RF but has a significant effect for ES-EO and G-TA (80N).

4.3 Net Moment At Lumbosacral Joint

For the measurement of net moment at the lumbosacral joint (L5/S1), the following parameters were determined for each trial:

Peak axial rotation moment during lifting (M_{PAR})

Peak lateral bending moment during lifting (M_{PLB})

Peak flexion-extension moment during lifting (M_{PFE})

Rate of change of axial rotation moment after sudden release of load (R_{AX})

Rate of change of lateral bending moment after sudden release of load (R_{LB})

Rate of change of flexion-extension moment after sudden release of load (R_{FE})

Definition of the above parameters can be found in section 3.7.5. The moment profile at the L5/S1 level was similar for all subjects and for all lifting conditions. As the lifting weight and lifting posture changed, the moment profile changed in magnitude but not in characteristic shape. All of the defined parameters were studied and compared under different lifting postures (symmetric squat, symmetric stoop and asymmetric stoop lifting) and lifting weights (20, 40, 60 and 80N).

4.3.1 Peak moment

The mean and standard deviation of the peak axial rotation moment, peak lateral bending moment and peak flexion-extension moment under different release of load conditions were determined (Table 4.26 and Figure 4.37-39). All the absolute value of peak moments were normalized by body weight (N) and body height (m) and are therefore presented and analysed as a percentage of the product of body weight and body height. It was found that both peak axial rotation moments, peak lateral bending moment and peak flexion-extension moment were affected by lifting posture and lifting weight. The results showed that the value of peak moments were highest when asymmetric stoop lifting was adopted compared with the other two lifting postures. The peak flexion-extension moment value also seems to increase with the weight lifted under all lifting postures. However, for peak axial rotation and lateral bending moment, peak moment values increased with the weight lifted only when asymmetric stoop lift was adopted.

Table 4.26 Mean and standard deviation of peak axial rotation moment, lateral bending moment and flexion-extension moment under different lifting conditions

Lifting conditions	Peak axial rotation moment (M_{PAR})		Peak lateral bending moment (M_{PLB})		Peak flexion-extension moment (M_{PFE})	
	Mean (%Nm)	S.D. (%Nm)	Mean (%Nm)	S.D. (%Nm)	Mean (%Nm)	S.D. (%Nm)
Sq20N	2.4	0.5	2.4	0.4	13.1	0.9
Sq40N	2.1	0.3	3.0	0.6	14.8	1.2
Sq60N	2.3	0.7	2.9	0.7	15.5	1.7
Sq80N	2.6	0.5	3.1	0.6	16.3	2.8
St20N	1.9	0.3	1.8	0.4	16.7	2.1
St40N	2.3	0.5	2.7	0.5	16.8	1.5
St60N	2.0	0.3	1.9	0.5	17.9	1.0
St80N	2.6	0.6	2.2	0.5	18.7	2.4
A20N	3.7	0.7	7.0	1.0	17.7	2.2
A40N	4.0	0.8	8.1	1.2	17.6	1.7
A60N	4.7	0.9	8.5	0.8	19.4	1.4
A80N	4.9	0.7	9.3	1.4	19.8	1.1

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift
%Nm=All peak moments were expressed as a percentage of the product of body weight (N) and body height (m)

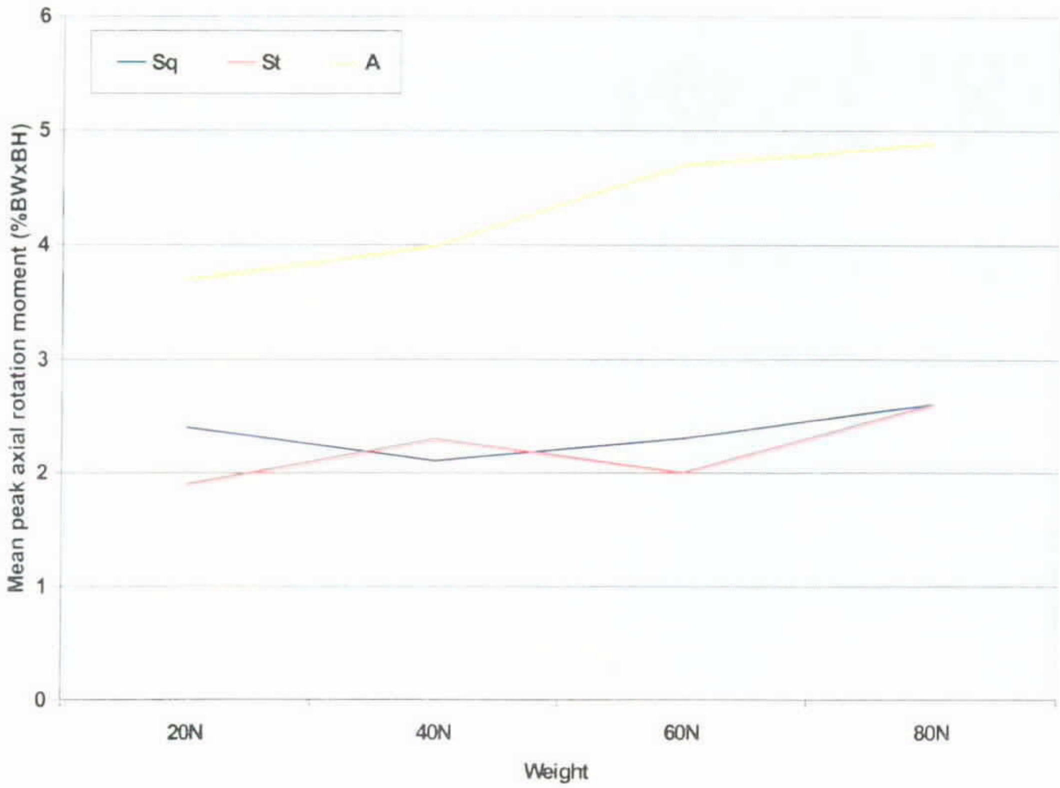


Figure 4.37 Mean peak axial rotation moment under different lifting conditions
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

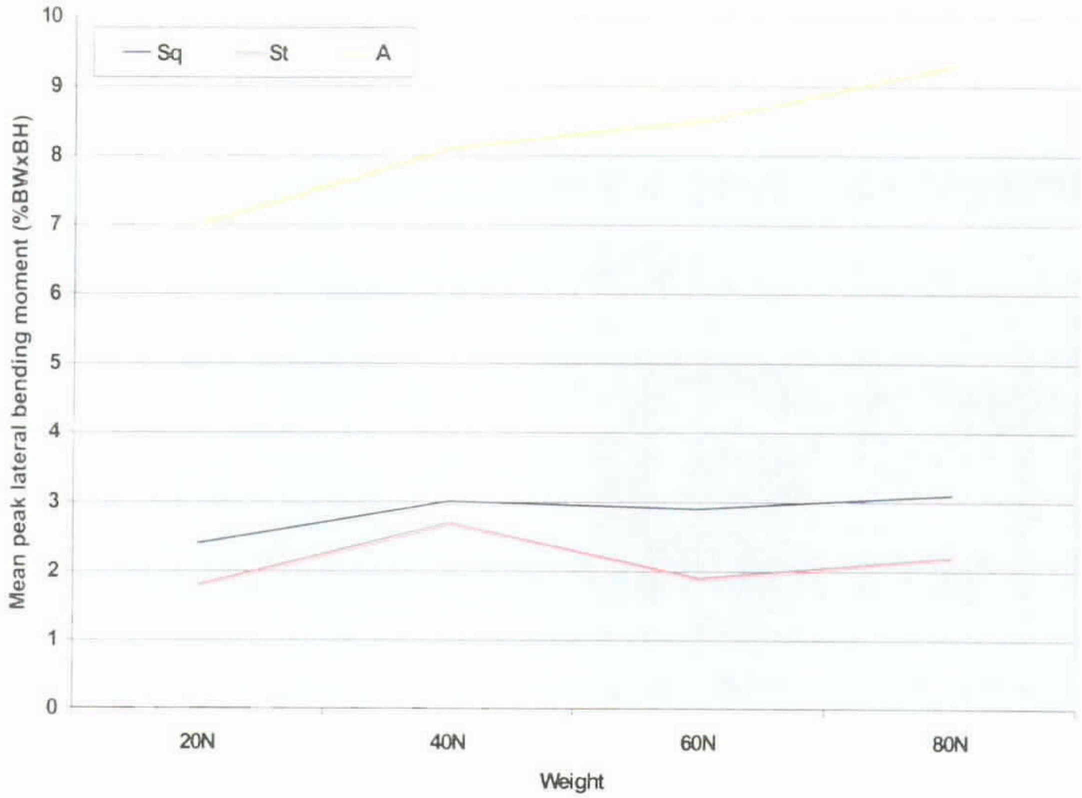


Figure 4.38 Mean peak lateral bending moment under different lifting conditions
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

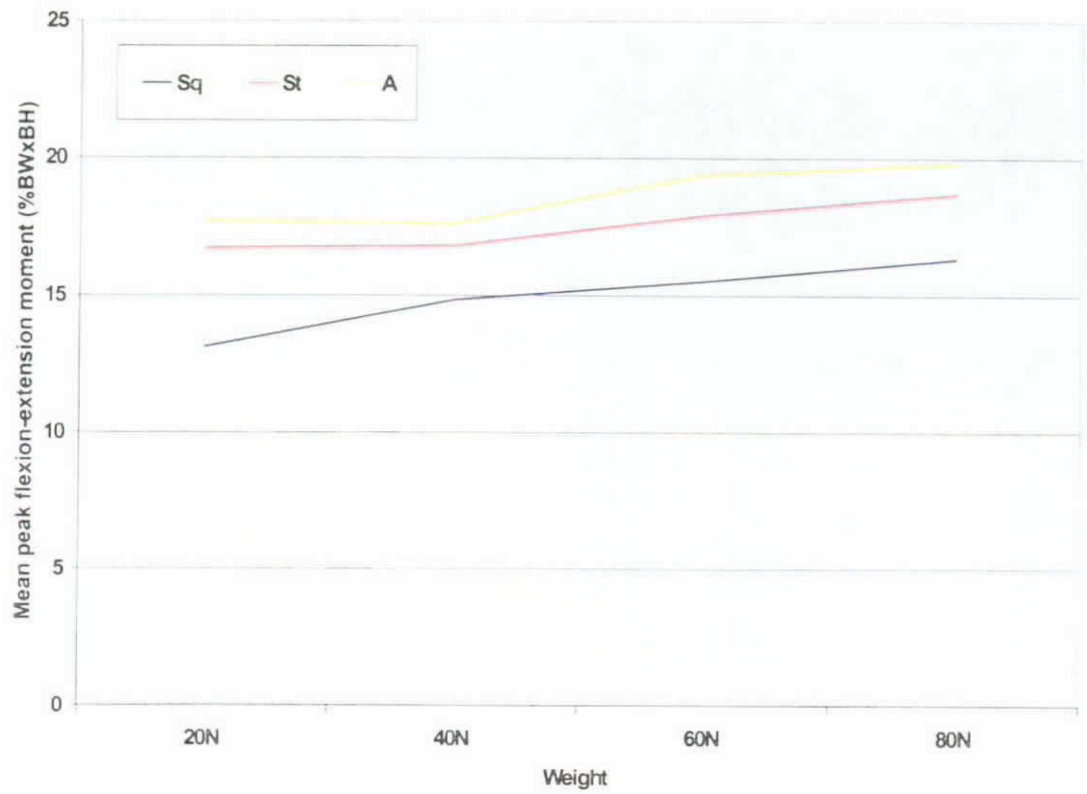


Figure 4.39 Mean peak flexion-extension moment under different lifting conditions
Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

Repeated measures ANOVA with two within-subject factors, lifting posture and lifting weight, were performed. Factor one consisted of three levels denoting three different lifting postures (symmetric squat, symmetric stoop and asymmetric stoop) while factor two consisted of four levels denoting four different lifting weights (20, 40, 60 and 80N). It was shown that there was statistically significant interaction between within-subject factors for both peak axial rotation moment, peak lateral bending moment and peak flexion-extension moment with p=0.016, 0.003 and 0.001 respectively. Therefore, data was analysed according to lifting weight. It was found that posture had significant effect on peak axial rotation moment, peak lateral bending moment and peak flexion-extension moment for all lifting weights (all with p=0.001 except p=0.006 at the lifting weight of 40N for peak flexion-extension moment). Table 4.27 shows the result of contrast tests comparing the three levels within posture factor.

For peak axial rotation moment, the peak moment for all the subjects ranged from 21.2 Nm to 54.6 Nm under different lifting postures and lifting weights. Asymmetric stoop lifting showed a significantly higher peak moment value a symmetric squat and stoop lifting. However, the peak moment value difference between symmetric squat and symmetric stoop lifting was not significant with p=0.121, 0.411, 0.419 and 0.929 for lifting weights of 20, 40, 60 and 80N respectively.

The peak lateral bending moment for all the subjects ranged from -31.7 Nm to -98.6 Nm under different lifting postures and lifting weights. Similar to peak axial rotation moment, asymmetric stoop lifting had a significantly higher peak lateral bending moment then symmetric squat and stoop lifting at all lifting weights. The peak moment value difference between symmetric squat lifting and symmetric stoop lifting was not significant (p=0.215) at the lifting weight of 40N, but at lifting weights of 20, 60 and 80N, symmetric squat lifting had a significantly higher peak moment value than symmetric stoop lifting (p=0.017,0.009 and 0.031, respectively).

For the peak flexion-extension moment, values ranged from 146.7 Nm to 260.9 Nm under different lifting postures and lifting weights. In contrast to the peak axial rotation and lateral bending moments, symmetric squat lifting had a significantly lower peak flexion-extension moment than the other lifting postures at all lifting weights. The differences in peak moment between symmetric and asymmetric stoop lifting were not statistically significant with p=0.204, 0.351, 0.057 and 0.155 for lifting weights from 20 to 80N, respectively.

Table 4.27 Contrast tests comparing peak axial rotation moment, peak lateral bending moment and peak flexion-extension moment among the three levels of posture

Lifting weight	Contrast tests comparing peak moment among the three levels of posture		
	Symmetric squat and asymmetric stoop	Symmetric stoop and asymmetric stoop	Symmetric squat and symmetric stoop
M_{PAR} 20N	0.001*	0.001*	0.121
M_{PAR} 40N	0.001*	0.001*	0.410
M_{PAR} 60N	0.001*	0.001*	0.419
M_{PAR} 80N	0.001*	0.001*	0.929
M_{PLB} 20N	0.001*	0.001*	0.017*
M_{PLB} 40N	0.001*	0.001*	0.215
M_{PLB} 60N	0.001*	0.001*	0.009*
M_{PLB} 80N	0.001*	0.001*	0.031*
M_{PFE} 20N	0.001*	0.204	0.001*
M_{PFE} 40N	0.003*	0.351	0.030*
M_{PFE} 60N	0.001*	0.057	0.018*
M_{PFE} 80N	0.001*	0.155	0.001*
* for p<0.05			

In general, asymmetric stoop lifting had the highest peak axial rotation moment and peak lateral bending moment in comparison with symmetric squat and stoop lift; however, symmetric stoop lift had the highest peak flexion-extension moment in comparison with symmetric squat and asymmetric stoop lift.

4.3.2 Rate of change of moment

In order to examine the effect of absolute rate of change of moment after sudden release of load on balance preservation, the mean and standard deviation of the absolute change of moment and rate of change of moment from the onset of sudden release of load to the first trough was determined for the axial rotation moment, lateral bending moment and flexion-extension moment under different release of load conditions (Tables 4.28-30 and figures 4.40-45). The results showed that for both absolute change of axial rotation moment (M_{CAR}), absolute change of lateral bending moment (M_{CLB}), absolute rate of change of axial rotation moment (R_{AR}) and absolute rate of change of lateral bending moment (R_{LB}), the highest values were found for asymmetric stoop lifting at all lifting weights. The absolute change of moment and absolute rate of change of moment also increased as the lifting weight increased under the asymmetric stoop lifting posture. The absolute change of flexion-extension moment (M_{CFE}) and absolute rate of change of flexion-extension moment (R_{FE}) increased as the lifting weight increased for all lifting postures.

Table 4.28 Mean and standard deviation of the absolute change and rate of change of axial rotation moment, after sudden release of load under different lifting conditions

Lifting conditions	Absolute change of axial rotation moment from onset of sudden unload to trough (M_{ARC})		Rate of absolute change of axial rotation moment from onset of sudden unload to trough (R_{AR})	
	Mean (Nm)	S.D. (Nm)	Mean (Nm/s)	S.D. (Nm/s)
Sq20N	5.5	0.8	17.6	0.9
Sq40N	8.6	1.2	22.4	1.3
Sq60N	6.4	1.1	19.9	0.8
Sq80N	7.9	0.7	23.7	2.1
St20N	4.8	0.9	13.7	0.5
St40N	6.3	0.6	15.4	0.3
St60N	5.1	0.5	17.5	0.7
St80N	4.9	0.3	16.7	1.2
A20N	17.1	0.8	72.6	4.7
A40N	17.6	1.3	75.3	3.4
A60N	19.8	0.9	81.6	3.9
A80N	25.6	2.1	86.7	5.1

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

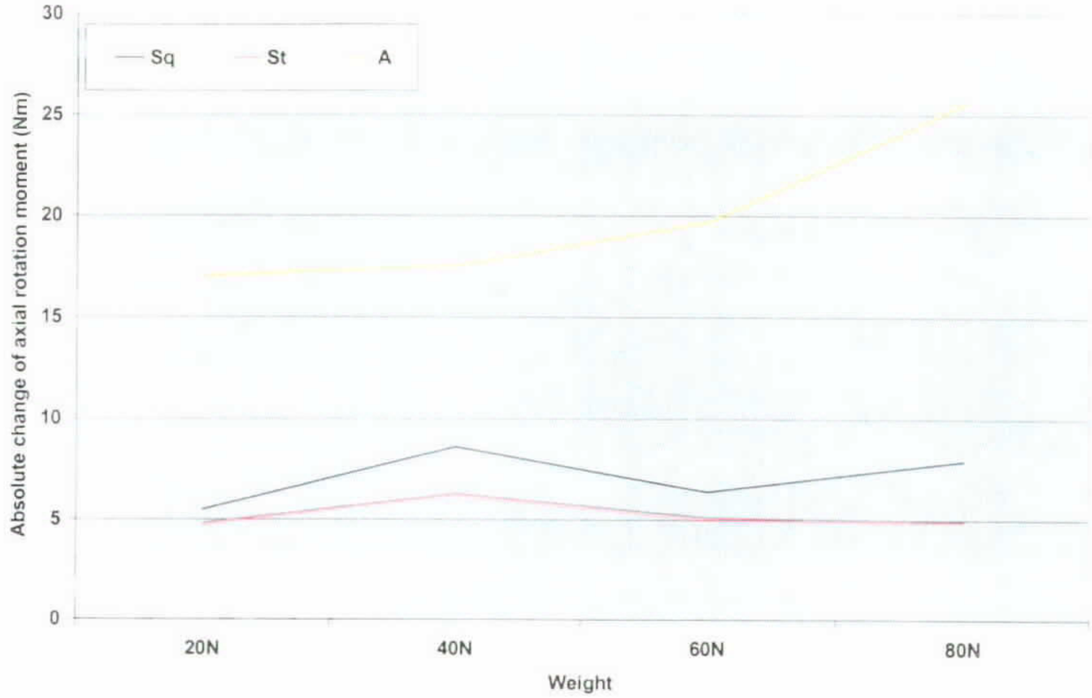


Figure 4.40 Absolute change of axial rotation moment after sudden release of load under different lifting conditions

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

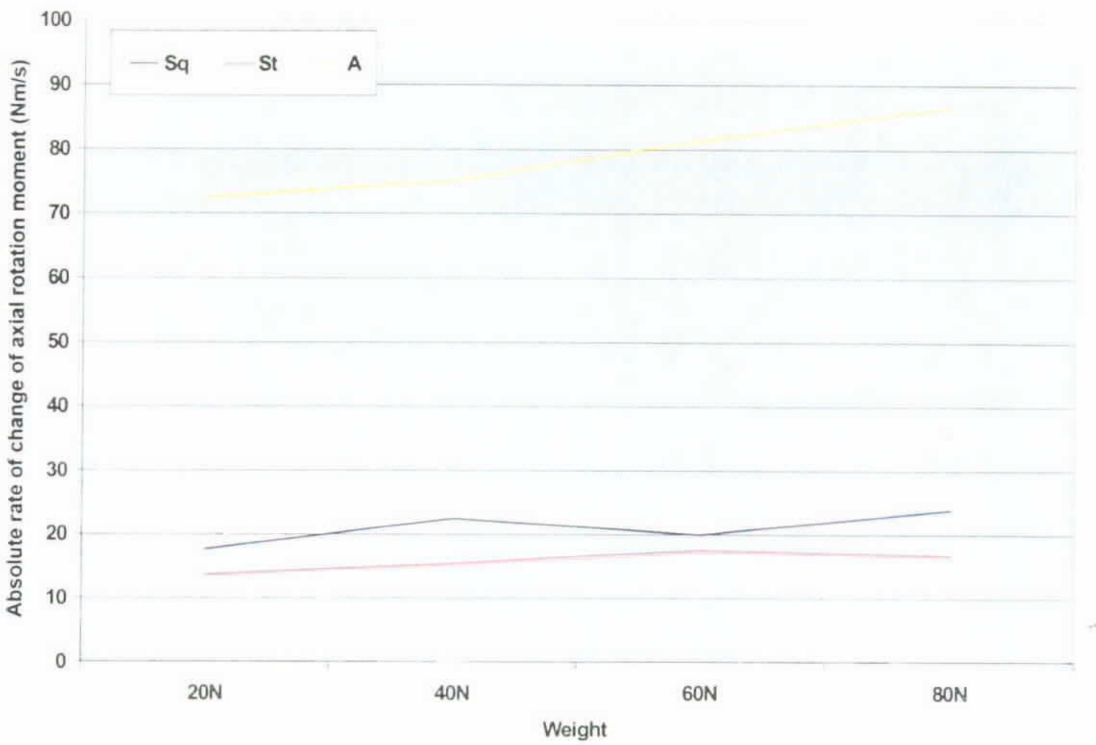


Figure 4.41 Absolute rate of change of axial rotation moment after sudden release of load under different lifting conditions

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

Table 4.29 Mean and standard deviation of the absolute change and rate of change of lateral bending moment after sudden release of load under different lifting conditions

Lifting conditions	Absolute change of lateral bending moment from onset of sudden unload to trough (M_{LBC})		Rate of absolute change of lateral bending moment from onset of sudden unload to trough (R_{LB})	
	Mean (Nm)	S.D. (Nm)	Mean (Nm/s)	S.D. (Nm/s)
Sq20N	5.1	1.3	38.9	6.4
Sq40N	6.7	0.9	43.4	7.8
Sq60N	7.5	1.5	65.6	5.2
Sq80N	6.9	0.7	59.9	7.9
St20N	3.4	0.9	22.8	5.7
St40N	4.6	0.5	21.3	3.4
St60N	4.1	0.8	19.9	3.1
St80N	5.7	1.2	26.7	2.4
A20N	43.5	5.6	131.6	8.7
A40N	49.7	4.1	156.2	11.2
A60N	61.2	3.9	168.7	9.8
A80N	58.6	5.7	187.8	15.3

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

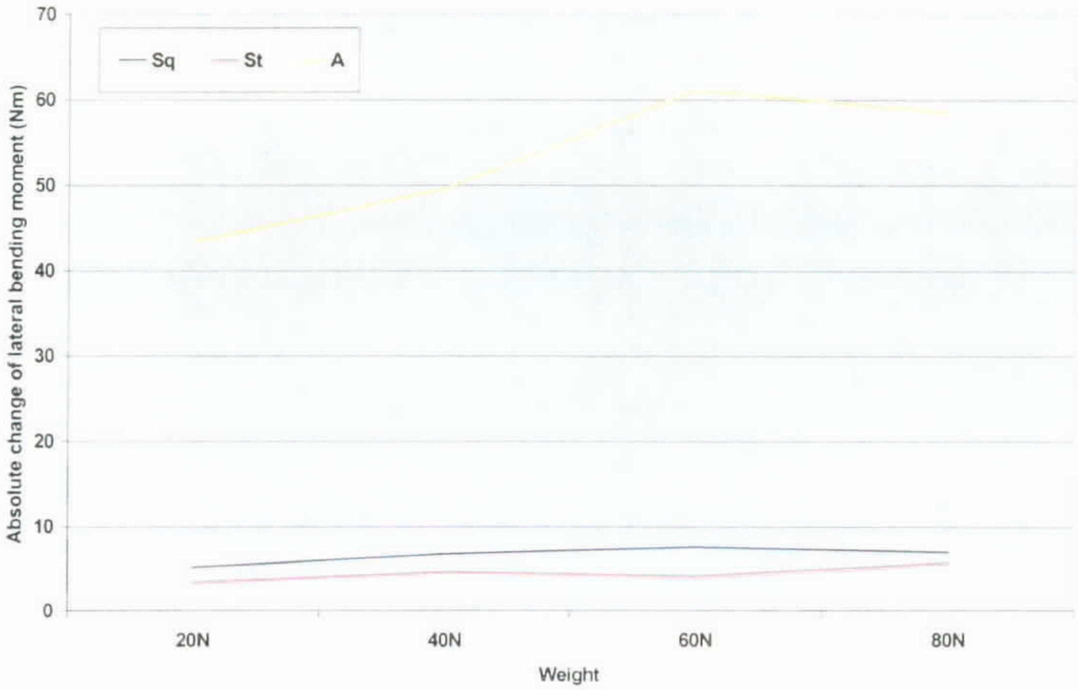


Figure 4.42 Absolute change of lateral bending moment after sudden release of load under different lifting conditions

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

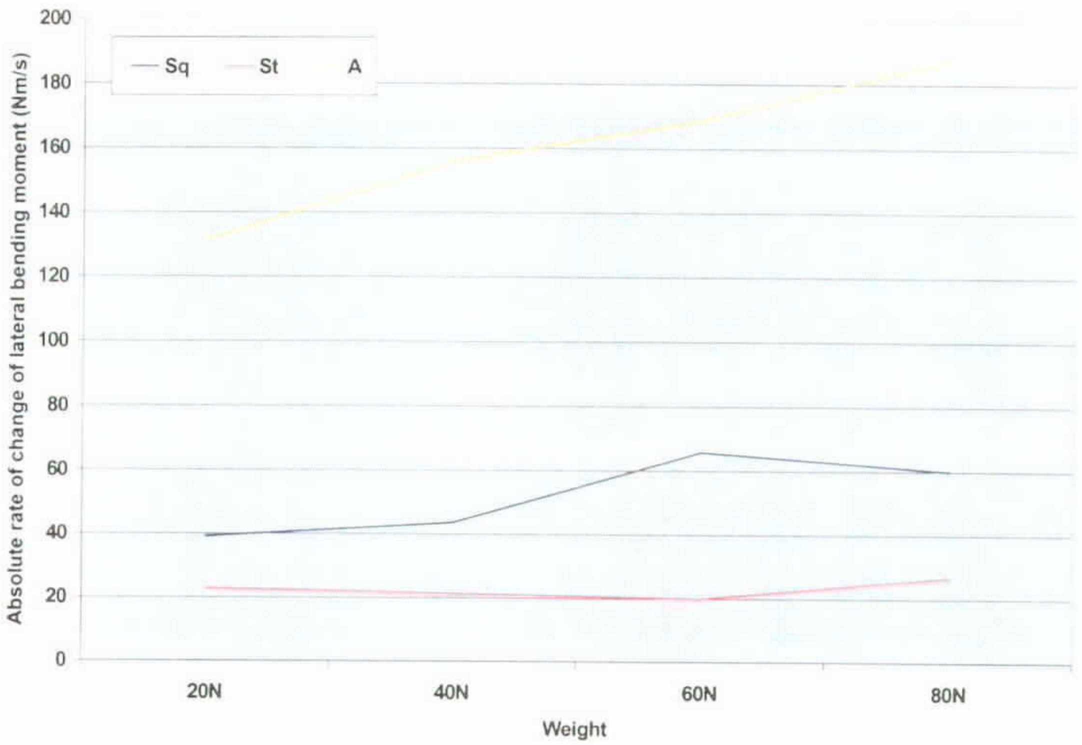


Figure 4.43 Absolute rate of change of lateral bending moment after sudden release of load under different lifting conditions

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

Table 4.30 Mean and standard deviation of the absolute change and rate of change of flexion-extension moment after sudden release of load under different lifting conditions

Lifting conditions	Absolute change of flexion-extension moment from onset of sudden unload to trough (M_{FEC})		Rate of absolute change of flexion-extension moment from onset of sudden unload to trough (R_{FE})	
	Mean (Nm)	S.D. (Nm)	Mean (Nm/s)	S.D. (Nm/s)
Sq20N	157.4	13.5	436.1	33.4
Sq40N	179.1	19.1	556.4	41.2
Sq60N	187.7	16.4	598.4	28.6
Sq80N	205.1	13.3	677.8	29.7
St20N	123.4	11.9	288.3	41.3
St40N	131.7	9.8	301.4	28.9
St60N	137.8	12.4	359.8	23.1
St80N	152.9	7.9	466.1	19.9
A20N	146.3	14.2	417.3	46.7
A40N	173.4	13.7	489.4	38.8
A60N	163.2	8.9	534.2	33.4
A80N	186.7	13.2	593.4	41.9

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

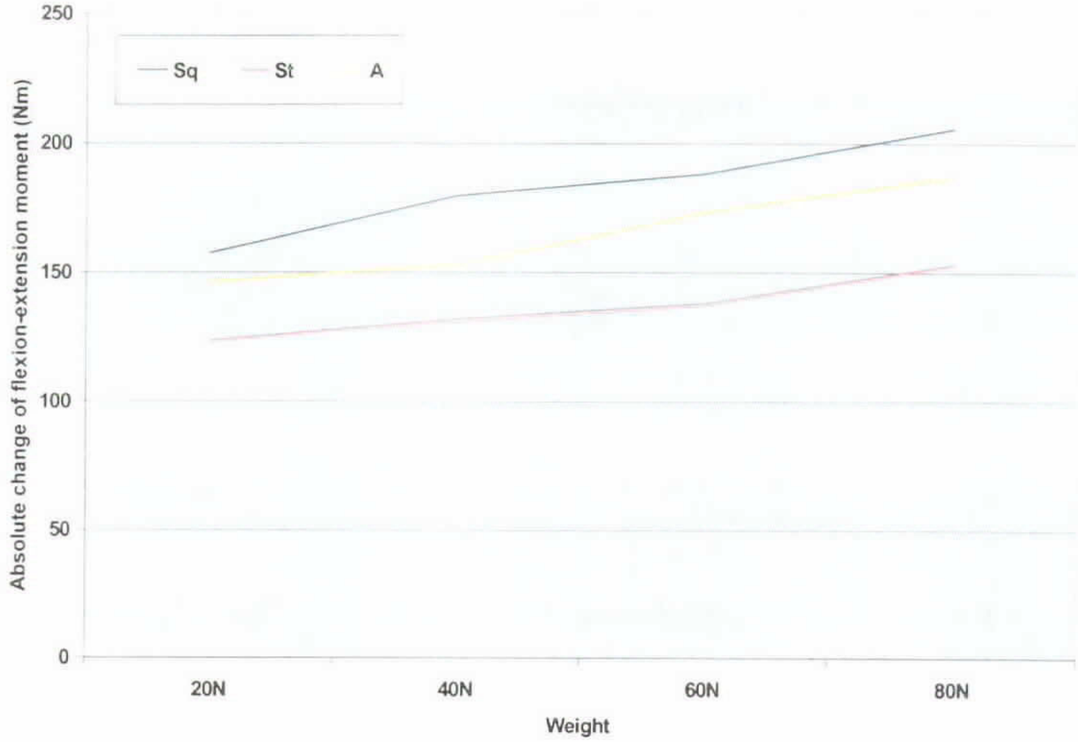


Figure 4.44 Absolute change of flexion-extension moment after sudden release of load under different lifting conditions

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

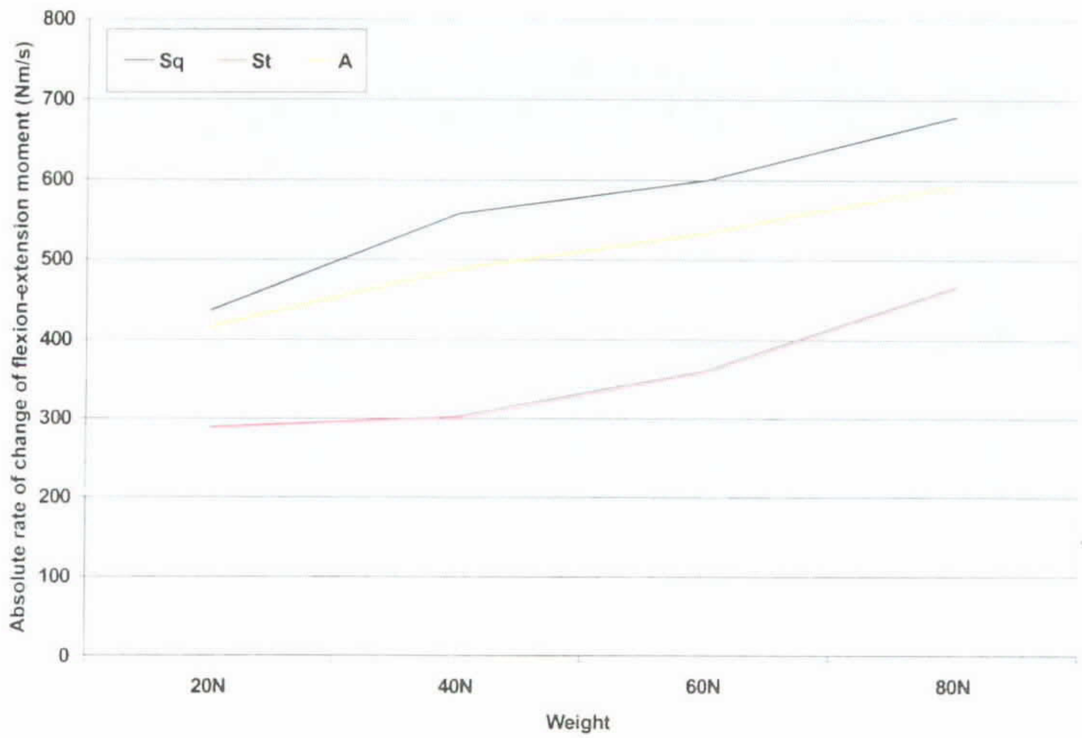


Figure 4.45 Absolute rate of change of flexion-extension moment after sudden release of load under different lifting conditions

Note: Sq=Symmetric squat lift; St=symmetric stoop lift; A=asymmetric stoop lift

Repeated measures ANOVA with two within-subject factors, lifting posture and lifting weight were performed for all absolute change and rate of change of moments. It was shown that there was statistically significant interaction between within-subject factors for M_{CAR}, M_{CLB}, M_{CFE}, R_{AR}, R_{LB}, and R_{FE} with p=0.017, 0.013, 0.001, 0.007, 0.041 and 0.039 respectively. Therefore, data was analysed according to lifting weight.

Asymmetric stoop lifting had a significantly higher M_{ARC} and R_{AR} than symmetric squat and symmetric stoop lifting at all lifting weights. Symmetric squat lifting had higher M_{ARC} and R_{AR} than symmetric stoop lifting; however, the difference was not statistically significant. Table 4.31 showed the contrast tests comparing M_{ARC} and R_{AR} after sudden release of load between the three levels of posture.

Table 4.31 Contrast tests comparing absolute change of axial rotation moment and absolute rate of change of axial rotation moment after sudden release of load among the three levels of posture

Lifting weight	Contrast tests comparing M_{ARC} and R_{AR} among the three levels of posture		
	Symmetric squat and asymmetric stoop	Symmetric stoop and asymmetric stoop	Symmetric squat and symmetric stoop
M_{ARC} 20N	0.012*	0.019*	0.121
M_{ARC} 40N	0.033*	0.021*	0.410
M_{ARC} 60N	0.014*	0.015*	0.419
M_{ARC} 80N	0.031*	0.016*	0.337
R_{AR} 20N	0.009*	0.004*	0.171
R_{AR} 40N	0.031*	0.022*	0.215
R_{AR} 60N	0.011*	0.007*	0.119
R_{AR} 80N	0.043*	0.041*	0.110
* for $p<0.05$			

For M_{CLB} and R_{LB}, (similar to M_{CAR} and R_{AR}), higher values were found when asymmetric stoop lifting was adopted and the difference between asymmetric stoop lifting and the other two lifting postures was statistically significant. Symmetric squat lifting also showed a higher M_{CLB} and R_{LB} than symmetric stoop lifting with the difference being insignificant for M_{CLB}, but significant for R_{LB} at lifting weights of 60N and 80N (Table 4.32).

Table 4.32 Contrast tests comparing absolute change of lateral bending moment and absolute rate of change of lateral bending moment after sudden release of load among the three levels of posture

Lifting weight	Contrast tests comparing M_{CLB} and R_{LB} among the three levels of posture		
	Symmetric squat and asymmetric stoop	Symmetric stoop and asymmetric stoop	Symmetric squat and symmetric stoop
M_{LBC} 20N	0.022*	0.029*	0.321
M_{LBC} 40N	0.013*	0.019*	0.410
M_{LBC} 60N	0.009*	0.015*	0.319
M_{LBC} 80N	0.017*	0.021*	0.337
R_{LB} 20N	0.009*	0.004*	0.115
R_{LB} 40N	0.031*	0.022*	0.153
R_{LB} 60N	0.011*	0.007*	0.041*
R_{LB} 80N	0.043*	0.041*	0.045*
* for $p<0.05$			

Unlike the axial rotation and lateral bending moments, symmetric squat lifting had a higher M_{CAR} and R_{AR} than symmetric and asymmetric stoop lifting, while asymmetric stoop lifting had a higher M_{CAR} and R_{AR} than symmetric stoop lifting. Symmetric squat had a significantly higher M_{CFE} and R_{FE} than symmetric stoop lifting at all lifting weights. Symmetric squat lifting also had a higher M_{FEC} and R_{FE} than asymmetric stoop lifting, however, this was not significant except at the lifting weight of 40N for M_{CFE}. Symmetric stoop lifting had a significantly lower M_{CFE} and R_{FE} than asymmetric stoop at all lifting weights except 20N and 40N for M_{CFE}. Table 4.33 shows the contrast tests comparing M_{CFe} and R_{FE} after sudden release of load among the three levels of posture.

Table 4.33 Contrast tests comparing absolute change of flexion-extension moment and absolute rate of change of flexion-extension moment after sudden release of load among the three levels of posture

Lifting weight	Contrast tests comparing M_{CFE} and R_{FE} among the three levels of posture		
	Symmetric squat and asymmetric stoop	Symmetric stoop and asymmetric stoop	Symmetric squat and symmetric stoop
M_{FEC} 20N	0.121	0.067	0.037*
M_{FEC} 40N	0.041*	0.081	0.022*
M_{FEC} 60N	0.134	0.045*	0.021*
M_{FEC} 80N	0.132	0.043*	0.016*
R_{FE} 20N	0.324	0.033*	0.021*
R_{FE} 40N	0.215	0.012*	0.009*
R_{FE} 60N	0.127	0.008*	0.017*
R_{FE} 80N	0.101	0.035*	0.015*
* for $p<0.05$			

In general, asymmetric stoop lifting had higher absolute change of axial rotation moment, absolute rate of change of axial rotation moment, absolute change of lateral bending moment and absolute rate of change of lateral bending moment than symmetric squat and symmetric stoop lifting. However, symmetric squat lifting had higher absolute change of flexion-extension moment and absolute rate of change of flexion-extension moment than symmetric and asymmetric stoop lifting.

CHAPTER 5 DISCUSSION

The results presented indicate the changes in COP, muscle activity and moment at the L5/S1 joint after sudden release during a lifting task. Unlike sudden application of load, which causes an anterior motion and allows the balance to be regained by calf muscle contraction, sudden release of load causes a posterior motion, which increase the risk of falling and resultant back injuries. Biomechanically, humans are better adapted to deal with a sudden flexion moment because of the counteracting role of the lever arm of the foot (Magnusson et al., 1996). However, sudden unloading accidents happen in leisure activities and more importantly, in the workplace. Therefore, a clear understanding of the mechanisms involved under sudden release of load condition is necessary for occupational safety guidelines.

5.1 Centre Of Pressure

The results showed that the sudden release of load during a lifting task causes an obvious COP deflection in both antero-posterior and medio-lateral directions. The response in antero-posterior and medio-lateral directions was found to occur between 70 ms to 124 ms, and 84 ms to 251 ms respectively. These results are comparable to those obtained by Magnusson et al (1996). The pattern of medio-lateral COP response time is very similar for all lifting postures, but symmetric squat and stoop lift have closer values than asymmetric stoop lift.

One of the objectives of this study is to investigate the ability of balance preservation under sudden unexpected release of load with different lifting postures and lifting weights by examining the COP excursion in antero-posterior and medio-lateral directions. For the dynamic situation, body weight, ground reaction force and inertia effects due to the motion are involved. The ground reaction force will be equal but opposite to the vector sum of the body weight and the inertial effects. With regard to the movement of the centre of pressure (COP), it should be within the base of support if the subject is able to maintain his equilibrium. Imbalance was judged to occur when the heels lost contact with the ground or when a compensatory step was made to prevent falling. If the subject does not move his feet, the more the COP moves towards the boundary of the base of support, i.e. the line joining the heels, the more likely the subject is to fall. If the distance between the COP and the posterior boundary of the base of support is inversely proportional to the lifting weight following sudden release, then the maximum weight that can be lifted without loss of balance under sudden release conditions can be predicted by means of extrapolation. The results presented indicate that ability of preserving balance was affected by lifting weight and lifting posture, and the results showed that the stoop lifting was comparatively more stable than the squat lifting during a sudden unload situation. The heavier the lifting weight, the higher the chance of losing balance and the higher the risk of injury. It was predicted that a subject would lose balance under a sudden unload condition for a squat lifting posture with the lifting weight over 114N, asymmetric stoop lifting posture with the lifting weight over 173N, or symmetric stoop lifting posture with the lifting weight over 205N.Our findings agree with those of Commissaris and Toussaint's (1997) concerning balance preservation. Their study investigated the effect of the presence or absence of load knowledge on the low-back loading and the control of balance in lifting tasks. They also found that preserving balance seemed easier while picking up load with stoop lift than with a squat lift.

It is generally accepted that the squat lifting technique, as opposed to the stoop technique, best minimizes the risk of low back injury when handling loads. However, unexpected sudden release of load during lifting and handling happens in everyday life and it seems that the squat lifting posture may not be ideal for people who are more likely to encounter sudden release of load such as refuse collectors, luggage dispatchers and movers.

5.2 Electromyography Measurement

Muscle recruitment and timing pattern play an important role in maintaining lumbar spine stability. In order to maintain a normal level of stability, trunk muscles must compensate by altering the typical activation pattern. Due to the multiple degrees of freedom and mechanical redundancy, both in structure and function, the neuromuscular control of the spinal system is extremely complex. Therefore, an adequate response to sudden loading depends not only on a sufficient muscle force but also on correct recruitment and timing patterns to assure the mechanical stability of the lumbar spine (Radebold et al., 2000).

Among the nine pairs of muscles studied, the latissimus dorsi, external oblique, internal oblique, rectus abdominis, rectus femoris and tibialis anterior responded by contraction after sudden release of load in all trials, while the posterior muscles (lumbar erector spinae, biceps femoris and gastrocnemius) all responded by relaxation. This result was as expected since the posterior muscle groups act as the prime movers during the lifting tasks, and therefore have to response by relaxation to prevent overshoot of the body when the weight lifted was suddenly released. The anterior muscle groups act as the antagonist and therefore respond by contraction while latissimus dorsi is a posterior back muscle, it responds to sudden release by contraction rather than relaxation. The reason for this may be due to its anatomical musculature and location, as the latissimus dorsi runs downwards across the trunk from the scapula to the thoracolumbar fascia. The thoracolumbar fascia extends across the lumbar region and is believed to provided stability to the spine (Thelen et al. 1995).

The internal oblique and latissimus dorsi showed no response to the sudden release stimulation at low weight in symmetric stoop lifting, while the rectus abdominis and internal oblique showed no response to the stimulation at low weight in asymmetric stoop lifting. This result may be because at low weights, the stability of the trunk was less threatened, and therefore muscles that have a primary function of trunk stabilization do not need to be activated. Another interpretation for the RA showing no response at low weight when asymmetric stoop lift was adopted was in terms of the requirements for trunk stabilization through the antagonistic forces these muscles generate. The frequency of physical oscillation of the trunk during symmetric conditions appears to be dependent on preview time. During unexpected and limited preview conditions, each subject's trunk displayed more oscillatory behaviour than during full preview conditions. Within these oscillations, the abdominal acted in a stabilizing role and pulled the body forward, counteracting the posterior musculature. The peak EMG activity of these antagonistic contractions increased as preview time decreased. This indicates that the trunk behaves like an underdamped system under conditions of limited or no preview time at all (i.e. sudden release). As preview time increased, behavior shifts towards a more heavily damped system, wherein little oscillatory behavior is exhibited. However, in the asymmetric conditions, the rectus abdominus muscles do not play the same antagonistic stabilizing role since the box was positioned at 30 degrees to the left of the subjects. The lines of action of both the left and right abdominal muscles were not directly opposite to the right posterior trunk muscles in the oblique direction and therefore the line of action of the rectus abdominus muscles was not optimal to counteract agonistic forces on the subject's right side. As a result, antagonistic muscle activity was not observed (Lavender et al., 1989).

In the current study, the physiological response times during the sudden release of load were mainly the M2 polysynaptic (50-90ms) for those quick onset muscle groups (i.e. tibialis anterior and rectus femoris) and mixture of triggered reaction (80-120ms), M3 voluntary response (120-180ms) with proprioceptive visual and the vestibular responses triggered by the postural changes (150ms) for other muscles (Schmidt, 1991). Tibialis anterior and rectus femoris were found to be the first two muscles to respond to the stimulation, followed by external oblique, internal oblique, rectus abdominis or erector spinae almost in all trials. Following the above muscles were the biceps femoris and then the latissimus dorsi and the last muscle to respond was the gastronemius. The results show that the sequence of muscle response consistently started from the distal ankle and then to the knee and finally to the proximal trunk. This type of muscle recruitment sequence can be described as an "ankle strategy" for preserving balance, which is characterized by the sequence of muscle recruitment and also the long period of co-contraction between tibialis anterior and the gastronemius. Ankle strategy starts preservation balance from the ankle joint to the joints closer to the perturbation point. Since muscles around the ankle were the first to respond to the perturbation by stretching, they might act as an inertia damper in this case, as they are already stretched. However, this finding is in contrast to the results of a study conducted by Oddsson et al. (1999), who found the sequence of muscle recruitment consistently started from the proximal trunk muscles down to the hip muscles and finally to the distal ankle joint. This difference is probably due to the experimental design, as in their study; the perturbation was induced to the foot via a force platform, while in the present study, the perturbation was applied directly to the trunk through upper limbs. In Dodson's study, the already stretched trunk muscles seem to act as the inertia damper.

For all lifting postures, the latency of muscles which responded by contraction decreased with increasing lifting weight and the latency of muscles which responded by relaxation increased with increasing lifting weight. It is possibly that at high lifting weights, longer co-contraction is needed to stabilize the trunk and maintain the body balance. With the exception of a few exceptional cases, the latency was found to be the longest when symmetric stoop lifting posture was adopted for most of the contraction group muscles (TA, EO, RF and RA). The latency of TA and RF was longer for asymmetric stoop lifting than symmetric squat lifting. For relaxation group muscles (BF, ES and G), the latency was found to be the longest for symmetric squat lifting. The latency of BF and G was found to be the shortest when asymmetric stoop lift was adopted. This may again indicate that symmetric stoop lifting allows comparatively easier preservation of balance under sudden release of load than asymmetric stoop or symmetric squat lifting posture, as the body will respond less quickly when there is less threat of loss of balance.

The duration of response was similar for left and right side for all muscles when symmetric squat lift or symmetric stoop lift was adopted. For contraction group muscles, the duration of response of the left side muscle is usually shorter than the right side when asymmetric stoop lift was adopted. However, for relaxation group muscles, the duration of response of the left side muscle is usually longer than the right side when asymmetric stoop lift was adopted. This may again be attributed to the "inertia damper" theory since under asymmetric stoop lifting posture, the load was set to the left side of the subject and therefore the right side muscles will respond earlier than the left and with a longer duration of response. Posture seems to have has no significant effect on the duration of response for TA, probably because it is the always the first muscle to response to the perturbation.

Co-contraction of agonistic and antagonistic muscle groups was shown to stiffen different joints and consequently stabilize the lumbar spine (Cholewicki and McGill, 1996). In this study, symmetric squat lifting had the longest co-contraction duration compared to symmetric stoop and asymmetric stoop lifting for all muscle couples. For muscle couple BF-RF and G-TA, symmetric squat lifting had the longest co-contraction duration followed by symmetric stoop lift, and asymmetric stoop lifting which had the shortest co-contraction duration. For muscle couple ES-EO, ES-IO and ES-RA, there is a trend for the co-contraction duration to increase as the lifting weight increases in all lifting postures. Overall these results show that the heavier the lifting weight is the higher chance of losing balance when encountering sudden release of load. Posture is another factor that can affect the ability of balance preservation, and it seems that stoop lifting is the most stable posture under sudden unloading since the least effort is required to maintain balance.

5.3 Loading About The Lumbosacral Joint

To control a lift appropriately, people prepare themselves before the actual lift. These anticipatory preparations concern increased levels of trunk activation and postural adjustments such as shifting the centre of mass backwards (De Looze et al., 2000). A prerequisite for a proper preparation is knowledge about the actual mass of the load before the attempt to lift. In a situation where people were misled by a sudden change of mass, an inadequate preparation was evoked, leading to a higher L5/S1 torque and a higher risk of falling, as compared with lifting with appropriate preparation (Commissaris and Toussaint, 1997). In the present study, the load mass to be lifted was suddenly released which the subject had no time to prepare for the change of mass and therefore causing rapid change of torque at the low back region resulting possible momentary loss of balance or uncontrolled segmental rotation.

Both peak axial rotation moments, peak lateral bending moment and peak flexion-extension moment were found to be affected by lifting posture and lifting weight. The results showed that the peak moments were highest when asymmetric stoop lifting was adopted compared with the other two lifting postures. Peak flexion-extension moments also seem to increase with the weight lifted under all lifting postures. However, for peak axial rotation and lateral bending moment, peak moment values increased with the weight lifted only when asymmetric stoop lifting was adopted.

The peak axial rotation moment for all the subjects ranged from 21.2 Nm to 54.6 Nm under different lifting postures and lifting weights while the peak lateral bending moment ranged from -31.7 Nm to -98.6 Nm and the peak flexion-extension moment ranged from 146.7 Nm to 260.9 Nm. In the current study, when the subject was lifting a weight of 60 N and asymmetric stoop lift was adopted, the average peak axial rotation moment and lateral bending moment were found to be 50.9 Nm and 85 Nm respectively. A study conducted by Schipplein in 1996 found values of 22.5 Nm for the peak axial rotation moment and 76.2 Nm for the peak lateral bending moment with an external load of 50 N. The peak axial rotation and lateral bending moment in our study was comparatively high and the discrepancy may due to the way the twisting task was performed. In our study, the subject was twisting while lifting a load while in the Schipplein's study, the subject was purely performing a twisting task and that may generate less axial rotation and lateral bending moment. A study conducted by Lavender et al. in 1999 concerning the effects of lifting speed on bending and twisting moment also found that in the full turn task the twisting moment was 27% smaller than that observed in the lift and twist task. For the peak flexion extension moment, our data is similar to those reported by Leskinen et al., 1983 (295-341 Nm with an external load of 150 N); Schipplein, 1996 (225-344 Nm with an external load of 50-250N) and Commissaris and Toussaint, 1997 (309.8 Nm with an external load of 160 N).

Sudden release of load caused an "overshoot" in L5/S1 moment. As the back muscles are under tension and resisting the flexion moment, which ends suddenly with unexpected release, their contraction activity initially continues without resistance, accelerating the trunk backwards and unbalancing the body. At this point, the muscles respond with a decreasing activity or relaxation before the abdominal muscles correct the body position. A study conducted by Commissaris and Toussaint in 1997 investigated the effect of the presence or absence of load knowledge on the low-back loading and the control of balance in lifting tasks stated that overestimating load mass can cause an "overshoot" in the linear momenta. In their study, subjects lifted a 6kg box, which they expected to be 16 kg, since, in a series of lifts, the load mass was changed from 16 to 6 kg without their knowledge. Just after box lift-off, the negative (backward) instantaneous horizontal momentum of the centre of gravity of the body reached a larger value in the 6kg trials than in the 16kg trials. Likewise, the positive (upward) instantaneous vertical momentum of the centre of gravity of the body increased more when the 6kg box was lifted. The positive instantaneous angular momentum of the body around a rotation axis situated in the body's centre of gravity showed an overshoot too when the overestimated load mass was lifted. Thus, the linear and angular momenta were larger than expected, leading to imbalance and compensatory reactions to regain balance. This explanation may also imply to the current study since at the moment when the load was suddenly released, the load being lifted was suddenly changed from the real load (20N to 80N) to the fake load (8N). Postural reactions required to regain balance could be hazardous to the low-back musculoskeletal system (Oddsson, 1990). Also, epidemiological studies have shown that workers exerting sudden unexpected maximal efforts are particularly vulnerable to low-back disorders (Magora, 1973).

The lifting event and moment at the L5/S1 can be divided into three phases. The first phase occurs as the subject bends down or kneels down to grasp the box; the resultant L5/S1 moment is due to the subject's upper body. Secondly, as the subject lifts the box and returns to an upright position, the L5/S1 moment results from the body mass of the subject and the weight lifted, showing a further rise and reaching a peak value as the pulling force on the box increases. The third phase occurs as the subject encounters sudden release of load and returns to a balanced position, where the L5/S1 moment shows a steep drop and fluctuates for a period of time after the sudden release of load while the subject is trying to preserve balance, still holding the fake load. The absolute change of axial rotation moment, lateral bending moment, flexion-extension moment, and the absolute rate of change of axial rotation moment, lateral bending moment, and flexion-extension moment were studied. The results showed that in general, asymmetric stoop lifting had the highest absolute change and rate of change of axial rotation and lateral bending moment, while symmetric squat lifting had the highest absolute change and rate of change of flexion-extension moment. Since the magnitude of flexion-extension changes were considerably higher than axial-rotation or lateral bending, squat lifting posture is therefore considered to be more hazardous in comparison with the other two lifting postures.

Although stoop and squat lifting have often been subjected to research to identify and study differences in, for instance, low-back loading (Toussaint et al., 1992), metabolic energy expenditure (De Looze, et al., 1992) or spinal shrinkage (Van Dieen et al., 1994), difference in control o of balance have rarely been assessed. In Commissaris and Toussaint's study (1997), they found preserving balance seemed easier (i.e. less risk of injury) while picking up a load with a stoop lift than with a squat lift. According to their study, picking up a load in front of the body induced a risk of toppling forward, because the body's centre of gravity quickly shifted forward and the counter-clockwise angular momentum of the body towards an erect posture was braked. Hence, the projection of the centre of gravity on the ground approached the front margin of the base of support and a smooth extending movement of the subject was hampered. Apparently, the subjects successfully minimized the adverse effects of these balance-threatening events, since balance was not lost. This was accomplished by specific preparations prior to load pick-up. In the first place, the adverse effect of the forward centre of gravity shift was reduced by the preparatory change in the instantaneous horizontal momentum of the centre of gravity of the body. During squat lifting, a profound backward instantaneous horizontal momentum of the centre of gravity of the body was created prior to load pick up, to brake the forward centre of gravity shift and thus prevent that the horizontal position of the centre of gravity of the body crossing the front margin of the base of support after load pick-up. During stoop lifting, a decrease in the posteriorly directed instantaneous horizontal momentum of the centre of gravity of the body occurred close to load pick-up, but it was not significant. Without preparation (i.e. sudden release of load condition), the backward instantaneous horizontal momentum of the centre of gravity of the body would have been smaller or would even have been directed forward. In the second place, the adverse effect of a braked counter-clockwise angular body momentum was reduced by a preparatory increase in instantaneous angular momentum of the body for both lifting techniques. Without the preparatory increase in angular momentum of the body, the angular momentum of the body would probably be too small at load pick up and the counter-clockwise rotation might then even reversed to clockwise rotation, which could induce a fall forward. In short, loss of balance when picking up a load was prevented by specific preparations that yielded a large backward centre of gravity momentum and a considerable counter-clockwise angular momentum at load pick up. It is important to note that load knowledge was required to execute these preparations, for sudden release of load yielded inadequate preparations.

The preparatory changes described above were not exactly the same for the squat lifting and the stoop lifting. This is because a difference in preparatory actions implies that the threat to balance at pick up of load was not the same for both techniques. A significant difference in preparation between techniques was found for the instantaneous horizontal momentum of the centre of gravity of the body, suggesting a differential effect of load pick up on the body's centre of gravity position. Without preparation, the horizontal position of the centre of gravity of the body was positioned closer to the front margin of the base of support in squat lifting compared with stoop lifting. The forward centre of gravity shift at load pick up was, therefore, more threatening when squat lift was adopted. Furthermore, a technique difference in the direction and magnitude of horizontal momentum of the centre of gravity at load pick up was observed. Without preparation, the horizontal momentum of the centre of gravity was about zero at load pick up in squat lifting, while the stoop lifting, horizontal momentum of the centre of gravity was negative, that is, directed backward. Hence, in stoop lifting, the forward centre of gravity shift at load pick up would be reversed by the backward horizontal momentum of the centre of gravity, even without preparatory actions, whereas in squat lifting, the centre of gravity would shift forward without being braked or reversed. Thus, these results suggest that preserving balance at load pick up was easier with stoop lifting than while picking up the same load using a squat lift (Commissaris and Toussaint, 1997). The similarity between Commissaris and Toussaint's study and the current study was due to subjects not having enough time for adequate preparation for the sudden decrease of load being lifted. Therefore, the mechanism described in Commissaris and Toussaint (1997) can possibly be applied to the present study in explaining why it is comparatively easier to preserve balance in stoop lifting than squat lifting under sudden release of load condition.

Traditionally, when considering compression on the lumbar spine during the performance of a lifting task, the recommended posture has always been the squat lift. The stoop lift approach has been shunned as it is believed to present greater risk of injury during lifting and while this traditional view of lifting posture is in general appropriate, some important exceptions exist. Results supporting the squat posture as safer have used loads, which can be lifted between the knees, which is the most common type industrial lifting. However, some loads are simply too bulky to fit between the knees and cannot be lifted using the squat posture. In this study, both the COP displacement analysis and EMG muscle activity showed that squat lifting may not be applicable to all people, especially those who are likely be exposed to sudden release of load. Revised safety guidelines and procedures should be established with these considerations in mind.

5.4 Limitations

Lack of consideration of the muscle effects in determining the moment at L5/S1 is the major drawback of this study. Without considering the muscle effects, the net moment on the L5/S1 level may be underestimated somewhat, since the muscle forces generated to stabilize the spine are often several times larger than the external load and body weight combined. These large muscle forces are responsible for most compressive and shear forces placed on the spine. As suggested by Thelen et al. (1995), co-contraction of lumbar muscles may contribute to 16-19% of the total muscles forces at the lumbar spine. Granta & Marras (1995) demonstrated that neglect of muscles co-contraction during lifting may underestimate spinal load compression by 45% and shear forces by 70%. Muscle co-contraction put extra load on the lumbar spine. For high sudden release loading, several co-contraction muscle couples, including RA-ES, EO-ES and LD-RA, were activated which may result in possible overloading of the spine. At the same time, the co-contraction duration of the co-contraction muscles couples in the lower limbs (i.e. the G-BF and TA-G) also significantly increased, which put extra load on the joints of the lower limbs. However, calculation of muscle force of free dynamic lifting involves the determination of force-length and force-velocity relationship and is therefore extremely complicated for sudden release conditions. Further biomechanical studies are required to examine the effects of muscles in order to estimate the joint loading more precisely and accurately under sudden release of load conditions.

Another limitation of this study is the training effect. A trained response will inevitably begin to appear during the experiment and affects the result to a certain extent. The response of the subject upon sudden release of load in this study was mostly experienced response rather than the first experience response which may not simulate the real situation and therefore, the sudden release loading and the repetition of sudden release stimulation were randomised in order to minimize the training effect in this study. Also, since fatigue may delay the muscle's response time and affect the results (Magnusson et al. 1996, Wilder et al. 1996), subjects were allowed to rest between trials to minimize muscle fatigue.

CHAPTER 6 CONCLUSION

The safety limit for different lifting postures were predicted by linear extrapolation and extra attention should be paid when squat lifting a weight over 114N, asymmetric stoop lifting a weight over 173N, or symmetric stoop lifting a weight over 205N. Since training will affect the results of a sudden release experiments, both the sudden release loading and the repetition cycle on which sudden release was stimulated were randomised to minimize the training effect. However, after the first release, anticipation of further release will inevitably occur to some extent, and the results presented in this study reflect those mostly experienced response rather than the first experience, and therefore the load limit will be lower for real situation.

Lifting imposes high mechanical loads on the musculoskeleltal structures of the lower back. The trunk musculature, which has several functions in lifting, plays a crucial role. The large superficial back muscles are activated to extend the trunk, while the abdominal muscles, in cooperation with the deep intersegmental back muscles, are involved in stabilizing the spine (Tesh et al. 1987, Panjabi et al. 1989). Activity of the back and abdominal muscles causes a large increase in the load on spinal motion segments (de Looze et al. 1999). This produces a neuromuscular challenge to adequately control the lifting movement and at the same time ensure that musculoskeletal loads are kept within safe margins.

Sudden release of load caused an "overshoot" in L5/S1 moment in all anatomical axes. When sudden release of load happened unexpectedly during a lifting task, the flexion moment ended suddenly and the exerted muscle force created to maintain equilibrium would generate an unexpected backward acceleration, unbalancing the body. Immediately, the body will respond by generating a very large muscle force, possibly larger than needed to regain the balance, and these postural reactions required to regain balance could be hazardous to the lumbar musculoskeletal system. (Oddsson, 1990; Magnusson et al., 1996 and Lavender et al., 1993). Therefore, sudden release of load may contribute to a higher risk of low-back injury in manual materials handling tasks since sudden maximal efforts could strain the soft tissues, especially when the worker is assuming an unfavourable posture for this type of effort. Additionally, many physical causes of low back pain such as bending and twisting were sudden maximal efforts incidentally carried out at the moment of incident (Magora, 1973).

Efforts to reduce the incidence of low-back disorder at the workplace are often based upon the evaluation of manual materials handling tasks, in which several load determining factors included the load location, the displacement of the load, the asymmetry of lifting, the lifting frequency and the coupling between load and hands are involved. The NIOSH equation provides a method for computing a weight limit for manual lifting from these factors (Waters et al., 1993). However, sudden release of load is a factor that is not accounted for in this equation.

The safety limit for squat lifting is lower than other postures, but it has been commonly adopted as the preferred posture for lifting, and therefore the current ergonomic guidelines for proper lifting should be carefully addressed if sudden release conditions are to be taken into account. The results of this study are valuable for establishing guidelines for manual material handlings, especially for people who are exposed to sudden unload, such as garbage collectors, truck unloaders and luggage dispatchers. Further biomechanical studies are required to determine the force generated by each muscle and therefore achieve a quantitative study of the load on the lumbar spine under different lifting conditions. Further biomechanical studies are required to include the effects of muscles in order to estimate the joint loading more precisely and accurately under sudden release of load conditions.

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APPENDIX 1

Data Sheet

Effect of Lifting Posture on the Load on the Lumbar Spine under an Unexpected Sudden Unloading Condition

A. Personal Particulars

Subject No. ______________
Age: ______________
Sex: ______________

B. Anthropometric Measurements

Body Weight: ______________
Body Height: ______________
Knee Height: ______________
Lifting distance: ______________
Shank circumference: ______________
Thigh circumference: ______________

C. Skin Resistance of Muscles (k ohms)

LES: ________	RES: ________	LRA: ________	RRA: ________
LIO: ________	EIO: ________	IEO: ________	REO: ________
LLD: ________	RLD: ________	LBF: ________	RBF: ________
LRF: ________	RRF: ________	LTA: ________	RTA: ________
LG: ________	RG: ________		

		Symmetric		Asymmetric
	Lifting Weight	Squat	Stoop	Stoop
Unload Cycle				

Remark: ____
____

Date: ______________________ Measured by: ______________________

APPENDIX 2

Equations for calculating centre of pressure

Parameter	Calculation	Description
Fx	=fx12+fx34	Medio-lateral force
Fy	=fy14+fy23	Anterior-posterior force
Fz	=fz1+fz2+fz3+fz4	Vertical force
Mx	=b*(fz1+fz2-fz3-fz4)	Plate moment about X-axis
My	=a*(-fz1+fz2+fz3-fz4)	Plate moment about Y-axis
Mx'	=b*(fz1+fz2-fz3-fz4)+Fy*az0	Plate moment about top plate surface
My'	=a*(-fz1+fz2+fz3-fz4)+Fx*az0	Plate moment about top plate surface
Mz	=b*(-fx12+fx34)+a*(fy14-fy23)	Plate moment about z-axis
Tz	Mz-Fy*ax+Fx*ay	Free moment, Vertical torque
Ax	=(Fx*az0-My)/Fz	X-Coordinate of force application point (COP)
ay	=(Fy*az0+Mx)/Fz	Y-Coordinate of force application point (COP)

APPENDIX 3

Subject Consent Form

I, ________________ (Name of Subject), hereby consent to participate in, as a subject; the research titled "Effects of Lifting Posture on the Load on the Lumbar Spine under an Unexpected Unloading Condition".

I understand the objectives and details of the experimental procedures, which have been explained to me by the investigator, Cheng Yung Wa

I also understand that I have the right to discontinue the experiment, with no reasons given, during the experiment.

I have given the opportunity to ask questions related to the test and the investigator has answered me satisfactory.

I realize that all of the findings will only be used for research purpose and the data may be published in different peer-reviewed journals or conference proceedings.

Results of this research are the properties of the Hong Kong Polytechnic University.

________________ ________________

Subject Signature Investigator Signature

Name of Witness: __________________ Witness Signature: ________________

Date: ______________

APPENDIX 4

Definition of Abbreviation

Symbol	Definition
m_1	Mass of right foot
m_2	Mass of left foot
m_3	Mass of right shank
m_4	Mass of left shank
m_5	Mass of right thigh
m_6	Mass of left thigh
m_7	Mass of pelvis
I_{1x}	Moment of inertia of right foot about AR axis
I_{1y}	Moment of inertia of right foot about LB axis
I_{1z}	Moment of inertia of right foot about FE axis
I_{2x}	Moment of inertia of left foot about AR axis
I_{2y}	Moment of inertia of left foot about LB axis
I_{2z}	Moment of inertia of left foot about FE axis
I_{3x}	Moment of inertia of right shank about AR axis
I_{3y}	Moment of inertia of right shank about LB axis
I_{3z}	Moment of inertia of right shank about FE axis
I_{4x}	Moment of inertia of left shank about AR axis
I_{4y}	Moment of inertia of left shank about LB axis
I_{4z}	Moment of inertia of left shank about FE axis
I_{5x}	Moment of inertia of right thigh about AR axis
I_{5y}	Moment of inertia of right thigh about LB axis
I_{5z}	Moment of inertia of right thigh about FE axis
I_{6x}	Moment of inertia of left thigh about AR axis
I_{6y}	Moment of inertia of left thigh about LB axis
I_{6z}	Moment of inertia of left thigh about FE axis
I_{7x}	Moment of inertia of pelvis about AR axis
I_{7y}	Moment of inertia of pelvis about LB axis

I_{7z}	Moment of inertia of pelvis about FE axis
$\vec{P}_{M1(t)}\left(x_{M1(t)}, y_{M1(t)}, z_{M1(t)}\right)$	Relative position of M1 at instant time t
$\vec{P}_{M2(t)}\left(x_{M2(t)}, y_{M2(t)}, z_{M2(t)}\right)$	Relative position of M2 at instant time t
$\vec{P}_{M3(t)}\left(x_{M3(t)}, y_{M3(t)}, z_{M3(t)}\right)$	Relative position of M3 at instant time t
$\vec{P}_{M4(t)}\left(x_{M4(t)}, y_{M4t)}, z_{M4(t)}\right)$	Relative position of M4 at instant time t
$\vec{P}_{M5}\left(x_{M5z(t)}, y_{M5(t)}, z_{M5(t)}\right)$	Relative position of M5 at instant time t
$\vec{P}_{M6(t)}\left(x_{M6(t)}, y_{M6(t)}, z_{M6(t)}\right)$	Relative position of M6 at instant time t
$\vec{P}_{M7(t)}\left(x_{M7(t)}, y_{M7(t)}, z_{M7(t)}\right)$	Relative position of M7 at instant time t
$\vec{P}_{M8(t)}\left(x_{M8(t)}, y_{M8(t)}, z_{M8(t)}\right)$	Relative position of M8 at instant time t
$\vec{P}_{M9(t)}\left(x_{M9(t)}, y_{M9(t)}, z_{M9(t)}\right)$	Relative position of M9 at instant time t
$\vec{P}_{M10(t)}\left(x_{M10(t)}, y_{M10(t)}, z_{M10(t)}\right)$	Relative position of M10 at instant time t
$\vec{P}_{M11(t)}\left(x_{M11(t)}, y_{M11(t)}, z_{M11(t)}\right)$	Relative position of M11 at instant time t
$\vec{P}_{M12(t)}\left(x_{M12(t)}, y_{M12(t)}, z_{M12(t)}\right)$	Relative position of M12 at instant time t
$\vec{P}_{M13(t)}\left(x_{M13(t)}, y_{M13(t)}, z_{M13(t)}\right)$	Relative position of M13 at instant time t
$\vec{P}_{M14(t)}\left(x_{M14(t)}, y_{M14(t)}, z_{M14(t)}\right)$	Relative position of M14 at instant time t
$\vec{P}_{M15(t)}\left(x_{M15(t)}, y_{M15(t)}, z_{M15(t)}\right)$	Relative position of M15 at instant time t
$\vec{P}_{Mpt(t)}\left(x_{Mpt(t)}, y_{Mpt(t)}, z_{Mpt(t)}\right)$	Relative position of mid point of M14M15 at instant time t
$\vec{P}_{M21(t)}\left(x_{M21(t)}, y_{M21(t)}, z_{M21(t)}\right)$	Relative position of M21 at instant time t
$\vec{P}_{M22(t)}\left(x_{M22(t)}, y_{M22(t)}, z_{M22(t)}\right)$	Relative position of M22 at instant time t
$\vec{P}_{C1}\left(x_{C1}, y_{C1}, z_{C1}\right)$	Spatial relationship between M6and the local coordinate system
$\vec{P}_{C2}\left(x_{C2}, y_{C2}, z_{C2}\right)$	Spatial relationship between M5 and the local coordinate system
$\vec{P}_{C3}\left(x_{C3}, y_{C3}, z_{C3}\right)$	Spatial relationship between M21 and the local coordinate system
$\vec{P}_{C4}\left(x_{C4}, y_{C4}, z_{C4}\right)$	Spatial relationship between M22 and the local coordinate system
$\vec{P}_{C5}\left(x_{C5}, y_{C5}, z_{C5}\right)$	Spatial relationship between L5/S1 centroid and the local coordinate system
$\vec{u}_{1(t)}\ \vec{v}_{1(t)}\ \vec{w}_{1(t)}$	Reference system for right foot at instant time t
$\vec{u}_{2(t)}\ \vec{v}_{2(t)}\ \vec{w}_{2(t)}$	Reference system for left foot at instant time t
$\vec{u}_{3(t)}\ \vec{v}_{3(t)}\ \vec{w}_{3(t)}$	Reference system for right shank at instant time t
$\vec{u}_{4(t)}\ \vec{v}_{4(t)}\ \vec{w}_{4(t)}$	Reference system for left shank at instant time t
$\vec{u}_{7(t)}\ \vec{v}_{7(t)}\ \vec{w}_{7(t)}$	Reference system for pelvis at instant time t
$\vec{u}_{D7}\ \vec{v}_{D7}\ \vec{w}_{D7}$	Reference system for pelvis of the skeleton

$\vec{P}_{RA(t)}\left(x_{RA(t)}, y_{RA(t)}, z_{RA(t)}\right)$	Relative position of right ankle joint centre at instant time t
$\vec{P}_{LA(t)}\left(x_{LA(t)}, y_{LA(t)}, z_{LA(t)}\right)$	Relative position of left ankle joint centre at instant time t
$\vec{P}_{RK(t)}\left(x_{RK(t)}, y_{RK(t)}, z_{RK(t)}\right)$	Relative position of right knee joint centre at instant time t
$\vec{P}_{LK(t)}\left(x_{LK(t)}, y_{LK(t)}, z_{LK(t)}\right)$	Relative position of left knee joint centre at instant time t
$\vec{P}_{RH(t)}\left(x_{RH(t)}, y_{RH(t)}, z_{RH(t)}\right)$	Relative position of right hip joint centre at instant time t
$\vec{P}_{LH(t)}\left(x_{LH(t)}, y_{LH(t)}, z_{LH(t)}\right)$	Relative position of left hip joint centre at instant time t
$\vec{P}_{1(t)}\left(x_{1(t)}, y_{1(t)}, z_{1(t)}\right)$	Relative position of CM of right foot at instant time t
$\vec{P}_{2(t)}\left(x_{2(t)}, y_{2(t)}, z_{2(t)}\right)$	Relative position of CM of left foot at instant time t
$\vec{P}_{3(t)}\left(x_{3(t)}, y_{3(t)}, z_{3(t)}\right)$	Relative position of CM of right shank at instant time t
$\vec{P}_{4(t)}\left(x_{4(t)}, y_{4(t)}, z_{4(t)}\right)$	Relative position of CM of left shank at instant time t
$\vec{P}_{5(t)}\left(x_{5(t)}, y_{5(t)}, z_{5(t)}\right)$	Relative position of CM of right thigh at instant time t
$\vec{P}_{6(t)}\left(x_{6(t)}, y_{6(t)}, z_{6(t)}\right)$	Relative position of CM of left thigh at instant time t
$\vec{P}_{7(t)}\left(x_{7(t)}, y_{7(t)}, z_{7(t)}\right)$	Relative position of CM of pelvis at instant time t
$\vec{a}_{1(t)}\left(a_{1x(t)}, a_{1y(t)}, a_{1z(t)}\right)$	Linear acceleration at CM of right foot at instant time t
$\vec{a}_{2(t)}\left(a_{2x(t)}, a_{2y(t)}, a_{2z(t)}\right)$	Linear acceleration at CM of left foot at instant time t
$\vec{a}_{3(t)}\left(a_{3x(t)}, a_{3y(t)}, a_{3z(t)}\right)$	Linear acceleration at CM of right shank at instant time t
$\vec{a}_{4(t)}\left(a_{4x(t)}, a_{4y(t)}, a_{4z(t)}\right)$	Linear acceleration at CM of left shank at instant time t
$\vec{a}_{5(t)}\left(a_{5x(t)}, a_{5y(t)}, a_{5z(t)}\right)$	Linear acceleration at CM of right thigh at instant time t
$\vec{a}_{6(t)}\left(a_{6x(t)}, a_{6y(t)}, a_{6z(t)}\right)$	Linear acceleration at CM of left thigh at instant time t
$\vec{a}_{7(t)}\left(a_{7x(t)}, a_{7y(t)}, a_{7z(t)}\right)$	Linear acceleration at CM of pelvis at instant time t
$\vec{v}_{1x(t)}\ \vec{v}_{1y(t)}\ \vec{v}_{1z(t)}$	Orientation of right foot at instant time t
$\vec{v}_{2x(t)}\ \vec{v}_{2y(t)}\ \vec{v}_{2z(t)}$	Orientation of left foot at instant time t
$\vec{v}_{3x(t)}\ \vec{v}_{3y(t)}\ \vec{v}_{3z(t)}$	Orientation of right shank at instant time t
$\vec{v}_{4x(t)}\ \vec{v}_{4y(t)}\ \vec{v}_{4z(t)}$	Orientation of left shank at instant time t
$\vec{v}_{5x(t)}\ \vec{v}_{5y(t)}\ \vec{v}_{5z(t)}$	Orientation of right thigh at instant time t
$\vec{v}_{6x(t)}\ \vec{v}_{6y(t)}\ \vec{v}_{6z(t)}$	Orientation of left thigh at instant time t
$\vec{v}_{7x(t)}\ \vec{v}_{7y(t)}\ \vec{v}_{7z(t)}$	Orientation of pelvis at instant time t
$\vec{\alpha}_{1(t)}\left(\alpha_{1x(t)}, \alpha_{1y(t)}, \alpha_{1z(t)}\right)$	Angular acceleration at CM of right foot at instant time t

$\vec{\alpha}_{2(t)}\left(\alpha_{2x(t)},\alpha_{2y(t)},\alpha_{2z(t)}\right)$	Angular acceleration at CM of left foot at instant time t
$\vec{\alpha}_{3(t)}\left(\alpha_{3x(t)},\alpha_{3y(t)},\alpha_{3z(t)}\right)$	Angular acceleration at CM right shank at instant time t
$\vec{\alpha}_{4(t)}\left(\alpha_{4x(t)},\alpha_{4y(t)},\alpha_{4z(t)}\right)$	Angular acceleration at CM of left shank at instant time t
$\vec{\alpha}_{5(t)}\left(\alpha_{5x(t)},\alpha_{5y(t)},\alpha_{5z(t)}\right)$	Angular acceleration at CM of right thigh at instant time t
$\vec{\alpha}_{6(t)}\left(\alpha_{6x(t)},\alpha_{6y(t)},\alpha_{6z(t)}\right)$	Angular acceleration at CM of left thigh at instant time t
$\vec{\alpha}_{7(t)}\left(\alpha_{7x(t)},\alpha_{7y(t)},\alpha_{7z(t)}\right)$	Angular acceleration at CM of pelvis at instant time t
$\vec{P}_{L5/S1(t)}\left(x_{L5/S1(t)}, y_{L5/S1(t)}, z_{L5/S1(t)}\right)$	Relative position of CM of L5/S1 at instant time t
$\vec{P}_{r(t)}\left(x_{r(t)}, y_{r(t)}, z_{r(t)}\right)$	Relative position of centre of pressure at instant time t
$\vec{F}_{L5/S1(t)}\left(F_{L5/S1x(t)}, F_{L5/S1y(t)}, F_{L5/S1z(t)}\right)$	Net force at the L5/S1 at instant time t
$\vec{F}_{r(t)}\left(F_{rx(t)}, F_{ry(t)}, F_{rz(t)}\right)$	Ground reaction force at instant time t
$\vec{M}_{L5/S1(t)}\left(M_{L5/S1x(t)}, M_{L5/S1y(t)}, M_{L5/S1z(t)}\right.$	Net moment at the L5/S1 at instant time t
$\vec{M}_{r(t)}\left(M_{rx(t)}, M_{ry(t)}, M_{rz(t)}\right)$	Ground reaction moment at instant time t
$\vec{T}_{(t)}\left(T_{x(t)}, T_{y(t)}, T_{z(t)}\right)$	Torque of force plate at instant time t
$[R_z]\ [R_y]\ [R_x]$	Three successive steps of rotation matrix
$[R]$	The resulting rotation matrix
$[R_1]$	The resulting rotation matrix for right foot
$[R_2]$	The resulting rotation matrix for left foot
$[R_3]$	The resulting rotation matrix for right shank
$[R_4]$	The resulting rotation matrix for left shank
$[R_5]$	The resulting rotation matrix for right thigh
$[R_6]$	The resulting rotation matrix for left thigh
$[R_7]$	The resulting rotation matrix for pelvis
$\theta_{11(t)}$, $\theta_{12(t)}$, $\theta_{13(t)}$	Carden angles of right foot at instant time t
$\theta_{21(t)}$, $\theta_{22(t)}$, $\theta_{13(t)}$	Carden angles of left foot at instant time t
$\theta_{31(t)}$, $\theta_{32(t)}$, $\theta_{33(t)}$	Carden angles of right shank at instant time t
$\theta_{41(t)}$, $\theta_{42(t)}$, $\theta_{43(t)}$	Carden angles of left shank at instant time t
$\theta_{51(t)}$, $\theta_{52(t)}$, $\theta_{53(t)}$	Carden angles of right thigh at instant time t
$\theta_{61(t)}$, $\theta_{62(t)}$, $\theta_{63(t)}$	Carden angles of left thigh at instant time t
$\theta_{71(t)}$, $\theta_{72(t)}$, $\theta_{73(t)}$	Carden angles for pelvis at instant time t
h	Sampling instant time, = 1/f (sampling freq.)

APPENDIX 5

Body Segment Mass and Moment of Inertia

Equations used to calculate the anthropometric parameters are shown as follows:

R. Foot (Vaughan *et al.*, 1992)

Mass:

$$m_1 = 0.0083\,(A_1) + 254.5\,(A_{13})\,(A_{15})\,(A_{17}) - 0.065$$

Moment of inertia about FE axis:

$$I_{1z} = 0.00023\,(A_1)\left[3\,(A_{13})^2 + 4\,(A_{15})^2\right] + 0.00022$$

Moment of inertia about LB axis:

$$I_{1y} = 0.00021\,(A_1)\left[3\,(A_{13})^2 + 4\,(A_{19})^2\right] + 0.00067$$

Moment of inertia about AR axis:

$$I_{1x} = 0.00141\,(A_1)\left[(A_{15})^2 + (A_{19})^2\right] - 0.00008$$

L. Foot (Vaughan *et al.*, 1992)

Mass:

$$m_2 = 0.0083\,(A_1) + 254.5\,(A_{14})\,(A_{16})\,(A_{18}) - 0.065$$

Moment of inertia about FE axis:

$$I_{2z} = 0.00023\,(A_1)\left[3\,(A_{14})^2 + 4\,(A_{16})^2\right] + 0.00022$$

Moment of inertia about LB axis:

$$I_{2y} = 0.00021\,(A_1)\left[3\,(A_{14})^2 + 4\,(A_{20})^2\right] + 0.00067$$

Moment of inertia about AR axis:

$$I_{2x} = 0.00141\,(A_1)\left[(A_{16})^2 + (A_{20})^2\right] - 0.00008$$

R. Shank (Vaughan *et al.*, 1992)

Mass:

$$m_3 = 0.0226\,(A_1) + 31.33\,(A_7)\,(A_9)^2 + 0.016$$

Moment of inertia about FE axis:

$$I_{3z} = 0.00347\,(A_1)\left[(A_7)^2 + 0.076\,(A_9)^2\right] + 0.00511$$

Moment of inertia of about LB axis:

$$I_{3y} = 0.00387\,(A_1)\left[(A_7)^2 + 0.076\,(A_9)^2\right] + 0.00138$$

Moment of inertia about AR axis:

$$I_{3x} = 0.00041\,(A_1)\,(A_9)^2 + 0.00012$$

L. Shank (Vaughan *et al.*, 1992)

Mass:

$$m_4 = 0.0226\,(A_1) + 31.33\,(A_8)\,(A_{10})^2 + 0.016$$

Moment of inertia about FE axis:

$$I_{4z} = 0.00347\,(A_1)\left[(A_8)^2 + 0.076\,(A_{10})^2\right] + 0.00511$$

Moment of inertia about LB axis:

$$I_{4y} = 0.00387\,(A_1)\left[(A_8)^2 + 0.076\,(A_{10})^2\right] + 0.00138$$

Moment of inertia about AR axis:

$$I_{4x} = 0.00041\,(A_1)\,(A_{10})^2 + 0.00012$$

R. Thigh (Vaughan *et al.*, 1992)

Mass:

$$m_5 = 0.1032\,(A_1) + 12.76\,(A_3)\,(A_5)^2 - 1.023$$

Moment of inertia about FE axis:

$$I_{5z} = 0.00762\,(A_1)\left[(A_3)^2 + 0.076\,(A_5)^2\right] + 0.01153$$

Moment of inertia about LB axis:

$$I_{5y} = 0.00726\,(A_1)\left[(A_3)^2 + 0.076\,(A_5)^2\right] + 0.01186$$

Moment of inertia about AR axis:

$$I_{5x} = 0.00151\,(A_1)\,(A_5)^2 + 0.00305$$

L. Thigh (Vaughan *et al.*, 1992)

Mass:

$$m_6 = 0.1032\,(A_1) + 12.76\,(A_4)\,(A_6)^2 - 1.023$$

Moment of inertia about FE axis:

$$I_{6z} = 0.00762\,(A_1)\left[(A_4)^2 + 0.076\,(A_6)^2\right] + 0.01153$$

Moment of inertia about LB axis:

$$I_{6y} = 0.00726\,(A_1)\left[(A_4)^2 + 0.076\,(A_6)^2\right] + 0.01186$$

Moment of inertia about AR axis:

$$I_{6x} = 0.00151\,(A_1)\,(A_6)^2 + 0.00305$$

Pelvis (Zheng *et al.*, 1998)

Mass:

$$m_7 = 0.13332\,(A_1)$$

Moment of inertia about FE axis:

$$I_{7z} = 0.02262$$

Moment of inertia about LB axis:

$$I_{7y} = 0.03218$$

Moment of inertia about AR axis:

$$I_{7x} = 0.02734$$

APPENDIX 6

Joint Centre

The following equations were used to define the uvw reference system and calculate the joint centres:

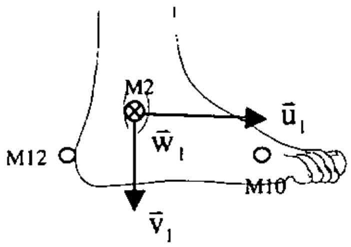
uvw reference system:

R. Foot (Vaughan *et al.*, 1992)

$$\vec{u}_{1(t)} = \frac{\vec{P}_{M10(t)} - \vec{P}_{M12(t)}}{\left|\vec{P}_{M10(t)} - \vec{P}_{M12(t)}\right|}$$

$$\vec{w}_{1(t)} = \frac{\left(\vec{P}_{M10(t)} - \vec{P}_{M2(t)}\right) \times \left(\vec{P}_{M12(t)} - \vec{P}_{M2(t)}\right)}{\left|\left(\vec{P}_{M10(t)} - \vec{P}_{M2(t)}\right) \times \left(\vec{P}_{M12(t)} - \vec{P}_{M2(t)}\right)\right|}$$

$$\vec{v}_{1(t)} = \vec{w}_{1(t)} \times \vec{u}_{1(t)}$$

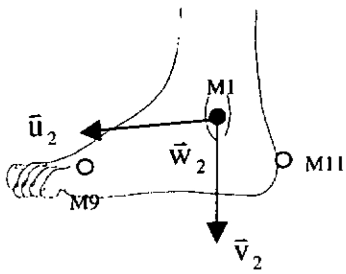


L. Foot (Vaughan *et al.*, 1992)

$$\vec{u}_{2(t)} = \frac{\vec{P}_{M9(t)} - \vec{P}_{M11(t)}}{\left|\vec{P}_{M9(t)} - \vec{P}_{M11(t)}\right|}$$

$$\vec{w}_{2(t)} = \frac{\left(\vec{P}_{M9(t)} - \vec{P}_{M1(t)}\right) \times \left(\vec{P}_{M11(t)} - \vec{P}_{M1(t)}\right)}{\left|\left(\vec{P}_{M9(t)} - \vec{P}_{M1(t)}\right) \times \left(\vec{P}_{M11(t)} - \vec{P}_{M1(t)}\right)\right|}$$

$$\vec{v}_{2(t)} = \vec{w}_{2(t)} \times \vec{u}_{2(t)}$$

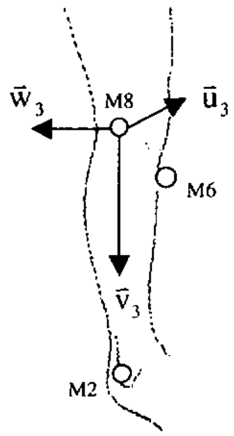


R. Shank (Vaughan *et al.*, 1992)

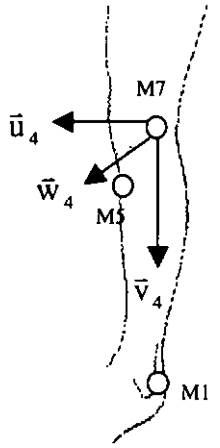
$$\vec{v}_{3(t)} = \frac{\vec{P}_{M2(t)} - \vec{P}_{M8(t)}}{\left|\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right|}$$

$$\vec{w}_{3(t)} = \frac{\left(\vec{P}_{M6(t)} - \vec{P}_{M8(t)}\right) \times \left(\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right)}{\left|\left(\vec{P}_{M6(t)} - \vec{P}_{M8(t)}\right) \times \left(\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right)\right|}$$

$$\vec{u}_{3(t)} = \vec{v}_{3(t)} \times \vec{w}_{3(t)}$$



L. Shank (Vaughan *et al.*, 1992)

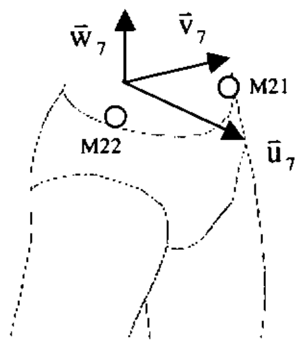


$$\vec{v}_{4(t)} = \frac{\vec{P}_{M1(t)} - \vec{P}_{M7(t)}}{\left|\vec{P}_{M1(t)} - \vec{P}_{M7(t)}\right|}$$

$$\vec{w}_{4(t)} = \frac{\left(\vec{P}_{M5(t)} - \vec{P}_{M7(t)}\right) \times \left(\vec{P}_{M1(t)} - \vec{P}_{M7(t)}\right)}{\left|\left(\vec{P}_{M5(t)} - \vec{P}_{M7(t)}\right) \times \left(\vec{P}_{M1(t)} - \vec{P}_{M7(t)}\right)\right|}$$

$$\vec{u}_{4(t)} = \vec{v}_{4(t)} \times \vec{w}_{4(t)}$$

Pelvis



$$\vec{v}_{7(t)} = \frac{\vec{P}_{M21(t)} - \vec{P}_{M22(t)}}{\left|\vec{P}_{M21(t)} - \vec{P}_{M22(t)}\right|}$$

$$\vec{w}_{7(t)} = \frac{\left(\vec{P}_{M22(t)} - \vec{P}_{Mpt(t)}\right) \times \left(\vec{P}_{M21(t)} - \vec{P}_{Mpt(t)}\right)}{\left|\left(\vec{P}_{M22(t)} - \vec{P}_{Mpt(t)}\right) \times \left(\vec{P}_{M21(t)} - \vec{P}_{Mpt(t)}\right)\right|}$$

$$\vec{u}_{7(t)} = \vec{w}_{7(t)} \times \vec{u}_{7(t)}$$

Locations of joint centres

R. Ankle joint (Vaughan *et al.*, 1992)

$$\vec{P}_{RA(t)} = \vec{P}_{M2(t)} - 0.008\,(A_{13})\,(\vec{u}_{1(t)}) + 0.393\,(A_{15})\,(\vec{v}_{1(t)}) + 0.706\,(A_{17})\,(\vec{w}_{1(t)})$$

L. Ankle joint (Vaughan *et al.*, 1992)

$$\vec{P}_{LA(t)} = \vec{P}_{M1(t)} - 0.008\,(A_{14})\,(\vec{u}_{2(t)}) + 0.393\,(A_{16})\,(\vec{v}_{2(t)}) - 0.706\,(A_{18})\,(\vec{w}_{2(t)})$$

R. Knee joint (Vaughan *et al.*, 1992)

$$\vec{P}_{RK(t)} = \vec{P}_{M8(t)} + 0.423\,(A_{11})\,(\vec{u}_{3(t)}) - 0.198\,(A_{11})\,(\vec{v}_{3(t)}) + 0.406\,(A_{11})\,(\vec{w}_{3(t)})$$

L. Knee joint (Vaughan *et al.*, 1992)

$$\vec{P}_{LK(t)} = \vec{P}_{M7(t)} + 0.423\,(A_{12})\,(\vec{u}_{4(t)}) - 0.198\,(A_{12})\,(\vec{v}_{4(t)}) - 0.406\,(A_{12})\,(\vec{w}_{4(t)})$$

R. Hip joint (Vaughan *et al.*, 1992)

$$\vec{P}_{RH(t)} = \vec{P}_{Mpt(t)} + 0.598\,(A_2)\,(\vec{u}_{7(t)}) - 0.344\,(A_2)\,(\vec{v}_{7(t)}) - 0.290\,(A_2)\,(\vec{w}_{7(t)})$$

L. hip joint (Vaughan *et al.*, 1992)

$$\vec{P}_{LH(t)} = \vec{P}_{Mpt(t)} + 0.598\,(A_2)\,(\vec{u}_{7(t)}) + 0.344\,(A_2)\,(\vec{v}_{7(t)}) - 0.290\,(A_2)\,(\vec{w}_{7(t)})$$

Centre of Mass

The relative positions of the longest toes should be found before defined the centre of mass location of the body segments:

Longest toes

R Longest toe (Vaughan *et al.*, 1992)

$$\vec{P}_{RT(t)} = \vec{P}_{M2(t)} + 0.697\,(A_{13})\,(\vec{u}_{1(t)}) + 0.780\,(A_{15})\,(\vec{v}_{1(t)}) + 0.923\,(A_{19})\,(\vec{w}_{1(t)})$$

L Longest toe (Vaughan *et al.*, 1992)

$$\vec{P}_{LT(t)} = \vec{P}_{M1(t)} + 0.697\,(A_{14})\,(\vec{u}_{2(t)}) + 0.780\,(A_{16})\,(\vec{v}_{2(t)}) - 0.923\,(A_{20})\,(\vec{w}_{2(t)})$$

Centre of Mass Locations

R. Foot (Vaughan *et al.*, 1992)

$$\vec{P}_{1(t)} = \vec{P}_{M12(t)} + 0.44\left(\vec{P}_{RT(t)} - \vec{P}_{M12(t)}\right)$$

L. Foot (Vaughan *et al.*, 1992)

$$\vec{P}_{2(t)} = \vec{P}_{M11(t)} + 0.44\left(\vec{P}_{LT(t)} - \vec{P}_{M11(t)}\right)$$

R. Shank (Vaughan *et al.*, 1992)

$$\vec{P}_{3(t)} = \vec{P}_{RK(t)} + 0.42\left(\vec{P}_{RA(t)} - \vec{P}_{RK(t)}\right)$$

L. Shank (Vaughan *et al.*, 1992)

$$\vec{P}_{4(t)} = \vec{P}_{LK(t)} + 0.42\left(\vec{P}_{LA(t)} - \vec{P}_{LK(t)}\right)$$

R. Thigh (Vaughan *et al.*, 1992)

$$\vec{P}_{5(t)} = \vec{P}_{RH(t)} + 0.39\left(\vec{P}_{RK(t)} - \vec{P}_{RH(t)}\right)$$

L. Thigh (Vaughan *et al.*, 1992)

$$\vec{P}_{6(t)} = \vec{P}_{LH(t)} + 0.39\left(\vec{P}_{LK(t)} - \vec{P}_{LH(t)}\right)$$

Pelvis

$$\vec{P}_{7(t)} = 0.5\,(\vec{P}_{M14(t)} + P_{M15(t)}) + 0.5\,(A_{22})\,(\vec{u}_{7(t)}) - 0.368\,(A_{21})\,(\vec{w}_{7(t)})$$

Linear Acceleration

The linear accelerations of all the required body segments were calculated by the following equations:

Linear acceleration at CM along x-axis at instant time t

$$a_{ix(t)} = \frac{1}{12\,h^2}\left[16\left(x_{i(t-h)} + x_{i(t+h)}\right) - \left(x_{i(t-2h)} + x_{i(t+2h)}\right) - 30\,x_{i(t)}\right]$$

Linear acceleration at CM along y-axis at instant time t

$$a_{iy(t)} = \frac{1}{12\,h^2}\left[16\left(y_{i(t-h)} + y_{i(t+h)}\right) - \left(y_{i(t-2h)} + y_{i(t+2h)}\right) - 30\,y_{i(t)}\right]$$

Linear acceleration at CM along z-axis at instant time t

$$a_{iz(t)} = \frac{1}{12\,h^2}\left[16\left(z_{i(t-h)} + z_{i(t+h)}\right) - \left(z_{i(t-2h)} + z_{i(t+2h)}\right) - 30\,z_{i(t)}\right]$$

where $i = 1, 2, 3\ldots,\text{or } 7$

$h = 1/\text{ frame rate}$

$h = 1/120$

APPENDIX 7

Body Segment Orientation

The following equations were used to calculate the orientation of the body segments:

R. Foot (Vaughan *et al.*, 1992)

$$\vec{v}_{1x(t)} = \frac{\left(\vec{P}_{M12(t)} - \vec{P}_{RT(t)}\right)}{\left|\vec{P}_{M12(t)} - \vec{P}_{RT(t)}\right|}$$

$$\vec{v}_{1z(t)} = \frac{\left(\vec{P}_{RA(t)} - \vec{P}_{M12(t)}\right) \times \left(\vec{P}_{RT(t)} - \vec{P}_{M12(t)}\right)}{\left|\left(\vec{P}_{RA(t)} - \vec{P}_{M12(t)}\right) \times \left(\vec{P}_{RT(t)} - \vec{P}_{M12(t)}\right)\right|}$$

$$\vec{v}_{1y(t)} = \vec{v}_{1z(t)} \times \vec{v}_{1x(t)}$$

L. Foot (Vaughan *et al.*, 1992)

$$\vec{v}_{2x(t)} = \frac{\left(\vec{P}_{M11(t)} - \vec{P}_{LT(t)}\right)}{\left|\vec{P}_{M11(t)} - \vec{P}_{LT(t)}\right|}$$

$$\vec{v}_{2z(t)} = \frac{\left(\vec{P}_{LA(t)} - \vec{P}_{M11(t)}\right) \times \left(\vec{P}_{LT(t)} - \vec{P}_{M11(t)}\right)}{\left|\left(\vec{P}_{LA(t)} - \vec{P}_{M11(t)}\right) \times \left(\vec{P}_{LT(t)} - \vec{P}_{M11t)}\right)\right|}$$

$$\vec{v}_{2y(t)} = \vec{v}_{2z(t)} \times \vec{v}_{2x(t)}$$

R. Shank (Vaughan *et al.*, 1992)

$$\vec{v}_{3x(t)} = \frac{\left(\vec{P}_{RK(t)} - \vec{P}_{RA(t)}\right)}{\left|\vec{P}_{RK(t)} - \vec{P}_{RA(t)}\right|}$$

$$\vec{v}_{3y(t)} = \frac{\left(\vec{P}_{M8(t)} - \vec{P}_{RK(t)}\right) \times \left(\vec{P}_{RA(t)} - \vec{P}_{RK(t)}\right)}{\left|\left(\vec{P}_{M8(t)} - \vec{P}_{RK(t)}\right) \times \left(\vec{P}_{RA(t)} - \vec{P}_{RK(t)}\right)\right|}$$

$$\vec{v}_{3z(t)} = \vec{v}_{3x(t)} \times \vec{v}_{3y(t)}$$

L. Shank (Vaughan *et al.*, 1992)

$$\vec{v}_{4x(t)} = \frac{\left(\vec{P}_{LK(t)} - \vec{P}_{LA(t)}\right)}{\left|\vec{P}_{LK(t)} - \vec{P}_{LA(t)}\right|}$$

$$\vec{v}_{4y(t)} = \frac{\left(\vec{P}_{LA(t)} - \vec{P}_{LK(t)}\right) \times \left(\vec{P}_{M7(t)} - \vec{P}_{LK(t)}\right)}{\left|\left(\vec{P}_{LA(t)} - \vec{P}_{LK(t)}\right) \times \left(\vec{P}_{M7(t)} - \vec{P}_{LK(t)}\right)\right|}$$

$$\vec{v}_{4z(t)} = \vec{v}_{4x(t)} \times \vec{v}_{4y(t)}$$

R. Thigh

$$\vec{v}_{5x(t)} = \frac{\left(\vec{P}_{RH(t)} - \vec{P}_{RK(t)}\right)}{\left|\vec{P}_{RH(t)} - \vec{P}_{RK(t)}\right|}$$

$$\vec{v}_{5y(t)} = \frac{\left(\vec{P}_{RH(t)} - \vec{P}_{RK(t)}\right) \times \left(\vec{P}_{M8(t)} - \vec{P}_{RK(t)}\right)}{\left|\left(\vec{P}_{RH(t)} - \vec{P}_{RK(t)}\right) \times \left(\vec{P}_{M8(t)} - \vec{P}_{RK(t)}\right)\right|}$$

$$\vec{v}_{5z(t)} = \vec{v}_{5x(t)} \times \vec{v}_{5y(t)}$$

L. Thigh

$$\vec{v}_{6x(t)} = \frac{\left(\vec{P}_{LH(t)} - \vec{P}_{LK(t)}\right)}{\left|\vec{P}_{LH(t)} - \vec{P}_{LK(t)}\right|}$$

$$\vec{v}_{6y(t)} = \frac{\left(\vec{P}_{M7(t)} - \vec{P}_{LK(t)}\right) \times \left(\vec{P}_{LH(t)} - \vec{P}_{LK(t)}\right)}{\left|\left(\vec{P}_{M7(t)} - \vec{P}_{LK(t)}\right) \times \left(\vec{P}_{LH(t)} - \vec{P}_{LK(t)}\right)\right|}$$

$$\vec{v}_{6z(t)} = \vec{v}_{6x(t)} \times \vec{v}_{6y(t)}$$

Pelvis

$$\vec{v}_{7z(t)} = \frac{\left(\vec{P}_{M21(t)} - \vec{P}_{M22(t)}\right)}{\left|\vec{P}_{M21(t)} - \vec{P}_{M22(t)}\right|}$$

$$\vec{v}_{7x(t)} = \frac{\left(\vec{P}_{M22(t)} - \vec{P}_{Mpt(t)}\right) \times \left(\vec{P}_{M21(t)} - \vec{P}_{Mpt(t)}\right)}{\left|\left(\vec{P}_{M22(t)} - \vec{P}_{Mpt(t)}\right) \times \left(\vec{P}_{M21(t)} - \vec{P}_{Mpt(t)}\right)\right|}$$

$$\vec{v}_{7y(t)} = \vec{v}_{7z(t)} \times \vec{v}_{7x(t)}$$

Carden Angles

In order to calculate the angular accelerations, Carden angles need to be found first. Carden angles are defined as three successive angles of rotation that transform a Cartesian coordinate system to another by an ordered sequence of rotations. The sequence adopted in this study was done by rotating the initial system of axes (x, y, z) with an angle θ_3 counter-clockwise about the flexion-extension axis (z-axis) and the resultant coordinate system was named as the (u, v, w) axes. Then, the system of axes (u, v, w) were rotated with an angle θ_2 counter-clockwise about the abduction-adduction axis (v-axis) to produce (u', v', w') axes. Finally, the system of axes (u', v', w') were rotated with an angle θ_1 counter-clock-wise about the rotation axis (u'-axis) to get the desired (x', y', z') system. Angles θ_3, θ_2 and θ_1 were used to describe the orientation of the (x', y', z') system relative to the (x, y, z) system.

Let $[R_z]$ $[R_y]$ $[R_x]$ be the three successive steps of rotation and their forms were defined as follows:

$$[R_z] = \begin{bmatrix} \cos\theta_3 & \sin\theta_3 & 0 \\ -\sin\theta_3 & \cos\theta_3 & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

$$[R_y] = \begin{bmatrix} \cos\theta_2 & 0 & -\sin\theta_2 \\ 0 & 1 & 0 \\ \sin\theta_2 & 0 & \cos\theta_2 \end{bmatrix}$$

$$[R_x] = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\theta_1 & \sin\theta_1 \\ 0 & -\sin\theta_1 & \cos\theta_1 \end{bmatrix}$$

The resulting rotation matrix $[R] = [R_x]\,[R_y]\,[R_z]$ then was formed as follow:

$$[R] = \begin{bmatrix} c\theta_2\, c\theta_3 & c\theta_2\, s\theta_3 & -s\theta_2 \\ s\theta_1\, s\theta_2\, c\theta_3 - c\theta_1\, s\theta_3 & s\theta_1\, s\theta_2\, s\theta_3 + c\theta_1\, c\theta_3 & s\theta_1\, c\theta_2 \\ c\theta_1\, s\theta_2\, c\theta_3 + s\theta_1\, s\theta_3 & c\theta_1\, s\theta_2\, s\theta_3 - s\theta_1\, c\theta_3 & c\theta_1\, c\theta_2 \end{bmatrix}$$

where c = cosine
s = sine

The three angles of each body segment were calculated by equating this resulting rotation matrix [R] to the numerical values of the corresponding rotation matrix found as follows:

$$\begin{bmatrix} \vec{v}_{ix(t)} \\ \vec{v}_{iy(t)} \\ \vec{v}_{iz(t)} \end{bmatrix}_{LCS} = [R_i] \begin{bmatrix} \vec{v}_{ix} \\ \vec{v}_{iy} \\ \vec{v}_{iz} \end{bmatrix}_{TV\ system}$$

$$[R_i] = \begin{bmatrix} \vec{v}_{ix(t)} \\ \vec{v}_{iy(t)} \\ \vec{v}_{iz(t)} \end{bmatrix}_{LCS} \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

$$[R_i] = \begin{bmatrix} r_{ixx(t)} & r_{ixy(t)} & r_{ixz(t)} \\ r_{iyx(t)} & r_{iyy(t)} & r_{iyz(t)} \\ r_{izx(t)} & r_{izy(t)} & r_{izz(t)} \end{bmatrix}$$

The Carden angles for each body segment at instant time t were calculated as follows:

$$-\sin\theta_{i2(t)} = r_{ixz(t)}$$

$$\theta_{i2(t)} = \sin^{-1}\left(-r_{ixz(t)}\right)$$

$$\sin\theta_{i1(t)}\cos\theta_{i2(t)} = r_{iyz(t)}$$

$$\theta_{i1(t)} = \sin^{-1}\left(\frac{r_{iyz(t)}}{\cos\left[\sin^{-1}\left(-r_{ixz(t)}\right)\right]}\right)$$

$$\sin\theta_{i3(t)}\cos\theta_{i2(t)} = r_{ixy(t)}$$

$$\theta_{i3(t)} = \sin^{-1}\left(\frac{r_{ixy(t)}}{\cos\left[\sin^{-1}\left(-r_{ixz(t)}\right)\right]}\right)$$

where i = 1, 2, …,or 7

Angular Acceleration

The angular accelerations of all the required body segments were calculated by the following equations:

Angular acceleration, CM rotating about x-axis at time t

$$\alpha_{ix(t)} = \frac{1}{12\,h^2}\left[16\left(\theta_{i1(t-h)} + \theta_{i1(t+h)}\right) - \left(\theta_{i1(t-2h)} + \theta_{i1(t+2h)}\right) - 30\,\theta_{i1(t)}\right]$$

Angular acceleration, CM rotating about y-axis at time t

$$\alpha_{iy(t)} = \frac{1}{12\,h^2}\left[16\left(\theta_{i2(t-h)} + \theta_{i2(t+h)}\right) - \left(\theta_{i2(t-2h)} + \theta_{i2(t+2h)}\right) - 30\,\theta_{i2(t)}\right]$$

Angular acceleration, CM rotating about z-axis at time t

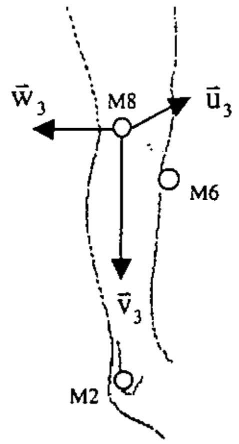
$$\alpha_{iz(t)} = \frac{1}{12\,h^2}\left[16\left(\theta_{i3(t-h)} + \theta_{i3(t+h)}\right) - \left(\theta_{i3(t-2h)} + \theta_{i3(t+2h)}\right) - 30\,\theta_{i3(t)}\right]$$

where i = 1, 2, 3…,or 7

APPENDIX 8

Static trial for the right shank:

According to Vaughan et al., (1992), right knee joint centre during dynamic lifting trials was found by using M2 (right lateral mallelous), M6 (right tibialis anterior) and M8 (right lateral femoral epicondyle) to form a reference coordinate system together with equation 3.1 as follow:



$$\vec{v}_{3(t)} = \frac{\vec{P}_{M2(t)} - \vec{P}_{M8(t)}}{\left|\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right|}$$

$$\vec{w}_{3(t)} = \frac{\left(\vec{P}_{M6(t)} - \vec{P}_{M8(t)}\right) \times \left(\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right)}{\left|\left(\vec{P}_{M6(t)} - \vec{P}_{M8(t)}\right) \times \left(\vec{P}_{M2(t)} - \vec{P}_{M8(t)}\right)\right|}$$

$$\vec{u}_{3(t)} = \vec{v}_{3(t)} \times \vec{w}_{3(t)}$$

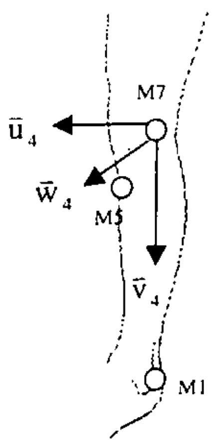
$$\vec{P}_{RK(t)} = \vec{P}_{M8(t)} + 0.423\,(A_{11})\,(\vec{u}_{3(t)}) - 0.198\,(A_{11})\,(\vec{v}_{3(t)}) + 0.406\,(A_{11})\,(\vec{w}_{3(t)})$$

Equation 3.1

where A_{11} is right knee diameter

Static trial for the left shank:

According to Vaughan et al., (1992), left knee joint centre during dynamic lifting trials was found by using M1 (right lateral mallelous), M5 (right tibialis anterior) and M7 (right lateral femoral epicondyle) to form a reference coordinate system together with equation 3.2 as follow:



$$\vec{v}_{4(t)} = \frac{\vec{P}_{M1(t)} - \vec{P}_{M7(t)}}{\left|\vec{P}_{M1(t)} - \vec{P}_{M7(t)}\right|}$$

$$\vec{w}_{4(t)} = \frac{\left(\vec{P}_{M5(t)} - \vec{P}_{M7(t)}\right) \times \left(\vec{P}_{M1(t)} - \vec{P}_{M7(t)}\right)}{\left|\left(\vec{P}_{M5(t)} - \vec{P}_{M7(t)}\right) \times \left(\vec{P}_{M1(t)} - \vec{P}_{M7(t)}\right)\right|}$$

$$\vec{u}_{4(t)} = \vec{v}_{4(t)} \times \vec{w}_{4(t)}$$

$$\vec{P}_{LK(t)} = \vec{P}_{M7(t)} + 0.423\,(A_{12})\,(\vec{u}_{4(t)}) - 0.198\,(A_{12})\,(\vec{v}_{4(t)}) - 0.406\,(A_{12})\,(\vec{w}_{4(t)})$$

Equation 3.2

where A_{12} is left knee diameter