THREE DIMENSIONAL KINETIC ANALYSIS OF ASYMMETRICAL LIFTING

BY

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ABSTRACT

Manual lifting is dynamic in nature and involves asymmetrical loading of the human body. This study investigated kinematic and kinetic characteristics of asymmetrical lifting in three dimensions, and then constructed a 3-D biomechanical force model of the lower back which is capable of quantifying torsional stress on the human spine.

Eleven healthy adult male manual workers were recruited as subjects and lifted a 10kg load placed at the sagittal plane (0^0) and at 30^0, 60^0 and 90^0 lateral planes to the right, from 150mm and 500mm initial lift heights, respectively, to an 800mm high bench in the sagittal plane. Subjects' spinal motions and the trajectorial movements of the load in three-dimensional space were monitored simultaneously by a Lumbar Motion Monitor and a V-scope Motion Analyzer.

Generally, the spinal motion factors increased as a function of increasing task asymmetry and differed ($p < 0.05$) between the lower (150mm) and higher (500mm) levels in the sagittal plane. In all asymmetrical conditions the motion factors showed a dramatic increase at the 500mm level compared to the increase at the 150mm level. The rates of increase in the horizontal and frontal planes were greater than those in the sagittal plane. Task asymmetry had a significant effect on the spinal kinematic parameters in the frontal plane at the two lift heights, and only at the high level (500mm) in the horizontal plane, with exception of average acceleration. Initial lift height exerted a significant effect on peak velocity and acceleration in both frontal and horizontal planes and on range of motion in the horizontal plane.

Kinetic characteristics of the object being lifted in three-dimensions increased with an increase in task asymmetry. The increase was more dramatic in the lateral direction in the horizontal plane. However, motion factors in the vertical direction dominated the full range of the lift, irrespective of task asymmetry and lift height. The kinetic measures differed ($p < 0.05$) between the lower (150mm) and the higher (500mm) levels in the vertical direction except for average force. Task asymmetry had a significant effect on dynamic measures in the anterior-posterior direction. Both task asymmetry and lift height had a significant effect on dynamic motion factors in the lateral direction.

From insights gained in the empirical study a three-dimensional biomechanical force model of the lower back was constructed based on a mechanism of muscle force re-orientation in the lumbar region. Acknowledging that the lower back is designed to be able to rotate around its longitudinal axis, the proposed model accounts for compression and shear forces and a torsional moment. The model has similar predictability to Schultz and Andersson's (1981) model when the human trunk exerts only a flexion-extension moment in the sagittal plane, but additionally predicts dramatic increases in shear forces, oblique muscle forces and torsional moment under asymmetrical lifting conditions which the Schultz-Andersson model does not. The increase rates in these forces and moment are not linearly related over task asymmetric angle.

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CHAPTER 1 : INTRODUCTION

With the rapid development of science and technology there has been a concomitant increase in the number of jobs in industry which are being undertaken by machines. Robots have been adapted to perform some highly structured and repetitive operations. Automatic production lines, controlled by electronic computers, greatly increase productivity and simultaneously reduce the labour intensity of workers. However, automation is always difficult in jobs which are unstructured, especially in the service industries, e.g. building construction, mechanical repair of equipment, baggage and package handling, police protection, and fire fighting (Chaffin, 1984).

Even in highly industrialized countries such as the United States, the United Kingdom, Germany and Japan, a large percentage of the workforce is presently required to exert significant physical strength as part of the job. It is estimated that over 30% of the total workforce in the U.S.A. is exposed to manual materials handling (MMH) situations, i.e. over 30 million workers are exposed daily, and reported work-related over-exertion injuries of all types occur to about 500,000 workers per year (NIOSH, 1981). Shahnavaz (1987) noted that in Sweden, where work conditions are recognised as being fairly acceptable, almost one in three of all blue-collar workers have some sort of muscular pain (physical load problem) mainly resulting from heavy lifting, vibration, repetitive short-cycle tasks, working posture, one-sided and static work. Even among white-collar workers, due to the very rapid expansion of VDT work, the problems of constrained posture as a result of prolonged positioning and repetitive strain are growing.

In most developing countries, the majority of the workforce is employed in agriculture, cottage industry or in small organisations (Asogwa, 1987; Mohan, 1987; Rebaza-Flores, 1987; Rainbird and O'Neill, 1995). Being less technologically advanced, under financial constraint, and supplied with a surplus of cheap labour, the manual labour component is very high and a great deal of man-machine interaction exists in industries of these countries. In South Africa, where the population of unskilled manpower vastly exceeds that of the skilled manpower, MMH is still a major factor in the majority of industries (Scott and Walraven, 1990).

For many years MMH has been recognized as a primary hazard of industrial workers (Chaffin and Park, 1973; NIOSH, 1981; Bishu **et al.**, 1984; Chaffin, 1988). In respect of the developed countries, Nordin (1987) and Matheson **et al.** (1995) reported that musculoskeletal disorders rank first among disease groups in frequency and effect. A large and diverse group of industries experience significant over-exertion injury and illness claims. In the U.S. about one quarter of all compensable work injuries are associated with the manual handling of objects (Snook, 1978; NIOSH, 1981). According to Buis (1990), manual-handling-related injuries account for 60-70% of all lost time due to work injuries in Australia, 60% of which are those involving the back. The injuries related to handling form a sizeable proportion (between 25 and 30%) of over-3-day injuries in most sectors of U.K. industry, and David (1985) has claimed that this proportion has remained constant, in spite of downward trends in reported accidents.

Such problems of manual handling injuries are more serious in technologically under-developed countries due to sub-optimal working conditions (Shahnavaz, 1987). Heavy lifting, harmful work postures because of poor machinery and workplace design, sub-optimal organization of work, static work, harmful materials handling, defective safeguards, and poor working environments (especially involving high levels of noise, vibration, heat and air pollution) are common in developing countries (Asogwa, 1987; Shahnavaz, 1987; Abeysekera, 1990; Shahnavaz **et al.**, 1991; Rainbird and O'Neill, 1995). Most workers in developing countries only achieve lower levels of education, and migrant-, child- and malnourished labour are common in the labour force (Asogwa, 1987; Mohan, 1987). These result in a disadvantaged workforce, more vulnerable to disease and job-related accidents. Unskilled workers are often employed largely or even solely in the jobs involving manual materials handling in warehouses, dockyards, commercial and recreational establishments and construction sites. Very few developing countries can provide accurate statistics on the nature and frequency of occupational injuries. Occupational accidents are usually under-reported and occupational diseases are very poorly reported in most developing countries (Shahnavaz, 1987). Official estimates are only available for large-scale industries. There are no detailed accident reports relative to agriculture and small-scale cottage industries (Mohan, 1987). For example, thousands of rural industries recently emerging in China are run privately and are thus beyond the purview of responsibility of the present Chinese industrial safety and health system. Since many of these industries are facing various problems, including low levels of technology installation, shortage of funds and lack of qualified managers and skilled workers, the working conditions of these small-scale industries are extremely poor and accident rates very high. It is very difficult to extrapolate from the limited information available, to the whole situation. Mohan (1987) estimated that the number of workplace fatalities would run into tens of thousands annually, and disabling injuries into several hundreds of thousands per **annum** in India alone.

NIOSH (1981) estimated that two-thirds of over-exertion injury claims involved lifting loads and about 20% involved pushing and pulling loads. From the extensive literature on the subject it is clear that most of manual handling injuries are to the lower back (Magora and Taustein, 1969; Rowe, 1969; Snook, 1978; NIOSH, 1981; Biering-Sorensen, 1985; Metzler, 1985; Charteris, 1991; Christensen **et al.**, 1995). Ayoub and Mital (1989) reiterate the point by reporting that nearly 50% of all back strains/sprains were precipitated by the manual lifting of loads.

Snook (1987) identified low back pain as the most common and costly musculoskeletal disorder in industry. Among work-related diseases and injuries, low back disabilities are, after lung diseases, the most frequent (Rowe, 1969; Kroemer, 1983). In the general population about one in two adults suffers at least once from low back pain (Valkenburg and Haanen, 1982). In certain working populations, such as plasterers, the lifetime prevalence of low back pain is no less than 80% (Dul and Hildebrandt, 1987). In the U.S.A., about 70 million people have suffered back injuries (Caillet, 1981; White, 1983). According to Haupt (1990), more than 20,000 cases of back-related injuries are reported annually to the Worker's Compensation Commission in South Africa. Shahnavaz **et al.** (1991) reported that 55.4% of workers in China and 74.8% of workers in Thailand have the problem of lower back pain. Klein **et al.** (1984) estimated that 19.0—25.5% of all workers' compensation claims are due to back pain. The unacceptably high incidence of back-related injuries is a world-wide problem. Estimates of working days lost to low-back pain are: 170 million working days per year in the U.S.A. (Loesser, 1979), 30 million days per year in Great Britain and 2 million days per year in Sweden (Hult, 1954). The frequent incidence of low back injuries costs U.S.A. employers billions of dollars annually, at a rate approximating $5100 per back injury (Kroemer, 1983; Kelsey **et al.**, 1978). This indeed constitutes an expensive industrial problem considering that, not only do industries pay worker's compensation, but also spend billions of dollars on tests, treatments, claims, lawsuit awards, settlements and surgeries. Cain and Pettry (1984) investigated the medical costs corresponding to various injuries in a coal mine in Virginia, U.S.A. and found that even though back sprains/strains comprised only about 17% of the injuries, they account for over 50% of the compensation costs and by far the majority of incidents requiring the longest time off from work.

Compared to other major health problems low back pain elicits relatively little attention at the workplace. One reason may be that musculoskeletal disorders are not causes of mortality (Chaffin and Andersson, 1984; Dul and Hildebrandt, 1987). It is important, however, to realize that the health and quality of life is greatly reduced for a large proportion of the population because of acute and chronic musculoskeletal disorders. It is therefore imperative that more effort be put into analysis and understanding of the problem, for as Nordin (1987) pointed out, workers constitute one of the most important sectors of the community, their health being an essential element in overall economic and social development. There thus is a need in each industry for a multidisciplinary approach for the prevention of musculoskeletal injuries.

The prevention of low back pain at the workplace has traditionally been attempted via three approaches: (i) selection of workers with sufficient capacity, and guidance of workers with reduced capacity; (ii) education and training of workers in work methods such as lifting techniques and (iii) ergonomic (re)design of the task or workplace. Many investigations have been carried out and guidelines provided for determining acceptable loads for two-handed, symmetrical lifting in the sagittal plane (Martin and Chaffin, 1972; Chaffin and Park, 1973; NIOSH, 1981; Ayoub and Selan, 1983; Ciriello and Snook, 1983; Liles and Deivanayagam, 1984; Mital, 1984b; ILO, 1988; Ayoub and Mital, 1989; Genaidy **et al.**, 1990).

Even though the vast majority of laboratory work on lifting has used sagittal symmetrical tasks, such tasks represent a minority of those encountered in the factory or warehouse. In a survey of over 2000 box movements in a variety of industries, Drury **et al.** (1982) found torso twisting in 80% of movements. Twisting of the spine while lifting is common in the workplace, where origin and destination are oriented at an angle to one another, where there is inadequate room to use a step turn, where lifting is done across the body as in swinging bags or boxes up from a low level, or where work is done in obstructed workplaces.

It is generally accepted that asymmetrical lifting is more hazardous to the musculoskeletal system than symmetrical lifting because of the combined effects of flexion, lateral flexion and accompanying axial rotation of the spine. Twisting while lifting can cause poor postural stability, increased EMG activity in the erector spinae and external oblique muscles (Kumar, 1980), increased intradiscal pressure and asymmetrical muscle activity (Andersson, 1985). Maximum acceptable weights and static strengths for lifting asymmetric loads asymmetrically are significantly lower than those for symmetrical lifting in the sagittal plane (Warwick **et al.**, 1980; Mital and Manivasagan, 1983; Garg and Badger, 1986; Mital and Fard, 1986; Garg and Banaag, 1988). Unfortunately, asymmetrical lifting has been studied in only a few cases, because of the experimental and biomechanical modelling complexities associated with three-dimensional force analysis. Further investigations are essential before practical and useful recommendations can be made.

In order to predict the risk and task performance capability of an individual, various kinds of biomechanical models of the lower back have been developed (Chaffin, 1984; Chaffin and Andersson, 1984; Chaffin, 1988; Ayoub and Mital, 1989). Current biomechanical spinal models of dynamic load-lifting activities are mostly restricted to sagittal plane motions. A limited number of three dimensional static models of the lumbar motion segments have been developed by Garg and Chaffin (1975), Schultz and Andersson (1981), Bean and Chaffin (1987) and Tracy (1990). The 3-D spinal motion segment models, however, are presently restricted to static analysis over a limited range of motion. Chaffin (1988) noted that the kinematic motion and inertial data acquired from asymmetric materials handling in industry will be highly complex in both spatial and temporal domains. To analyze such complex biomechanical data, and to be able to predict the stress imposed on the lumbar motion segments, requires development of a dynamic 3-D biomechanical model of a person when lifting a load under asymmetrical conditions.

1.1 STATEMENT OF THE PROBLEM

The majority of industrial manual lifting activities involve asymmetrical lifting (three-dimensional lifting). Uniplanar lifting activities in the strictly sagittal plane are extremely rare in the practical ambience of the working situation. The aim of the present study was to synthetically analyze the kinematic and kinetic characteristics of asymmetrical manual lifting activities under controlled laboratory conditions. For this purpose asymmetrical lifting is defined as cross-planar lifting. Two-handed symmetrical (uniplanar) lifting tasks were also studied to provide a basis for comparison. The specific objectives of the study were:

(1) To perform a comprehensive investigation of angular motions of the human spine during asymmetrical lifting in order to understand more rationally the complex kinematics and human limitations associated with asymmetrical lifting;

(2) To analyse comparatively the kinetic properties of asymmetrical lifting at various angles of asymmetry to the right of the sagittal plane;

(3) To provide, via this kinetic basis, a 3-D biomechanical model of the lower back under asymmetrical lifting conditions.

This study made use of the Lumbar Motion Monitor and the V-Scope Ultrasonic Motion Monitor to analyse comprehensively human body movements in three dimensional space during asymmetrical lifting. In an asymmetrical lifting act, the human spine moves in three planes in different ways and to different degrees. Even in the same plane the ranges of motion of vertebral segments differ from one another. These differences impose different stresses on body segments. However all movements of bodily segments are related to displacements of the load being lifted and the lifting techniques being used. In this study, relationships between lumbar spinal rotational angles and load displacements in the horizontal plane were investigated. A detailed evaluation of kinetic variations of asymmetrical lifting, in terms of increased angles from the sagittal plane, was carried out and the relationship between lumbar spinal motions and load movements in three dimensional space were analysed.

The present study was undertaken in two phases; an empirical (i.e. observational) phase during which detailed kinematic and kinetic data were gathered under symmetrical and asymmetrical lifting conditions; and a conceptual (i.e. creative) phase in which a 3-D biomechanically-based model was constructed to account for the kinematics of asymmetrical lifting. The intent was to have phase two facilitated by, but not necessarily restricted to, the findings derived from phase one.

1.2 RESEARCH HYPOTHESES RELATIVE TO PHASE ONE

Phase one was designed to evaluate the difference of angular movements and kinetic properties between symmetrical lifting and asymmetrical lifting at different angles. Generally, angular movement was surveyed within each specific plane involved, and kinetic measures were made according to the co-ordinate axes of a reference frame.

The following hypotheses were developed for this investigation:

Hypothesis 1:

Spinal kinematic measures during manual lifting are not affected by differences in task asymmetry or initial lift height.

Statistically:

$$H_o : \mu D(0)_{L;H} = \mu D(30)_{L;H} = \mu D(60)_{L;H} = \mu D(90)_{L;H}$$

$$H_a : \mu D(0)_{L;H} \neq \mu D(30)_{L;H} \neq \mu D(60)_{L;H} \neq \mu D(90)_{L;H}$$

where: D = the measured kinematic variables of angular movement of the human lumbar spine in three planes, including range of motion, angular velocity, and angular acceleration.

numerals = angular deviations from the sagittal plane.

L;H = low (150mm) and high (500mm) initial lift height levels.

Hypothesis 2:

The kinetic characteristics of manual lifting are not affected by differences in task asymmetry or initial lift height.

Statistically:

$$H_o : \mu E(0)_{L;H} = \mu E(30)_{L;H} = \mu E(60)_{L;H} = \mu E(90)_{L;H}$$

$$H_a : \mu E(0)_{L;H} \neq \mu E(30)_{L;H} \neq \mu E(60)_{L;H} \neq \mu E(90)_{L;H}$$

where: E = the kinetic measurements of manual lifting in three-dimensional space, including velocity, acceleration, force, work and power.

numerals = angular deviations from the sagittal plane.

L;H = low (150mm) and high (500mm) initial lift height levels.

1.3 DELIMITATIONS RELATIVE TO PHASE ONE

A working area of $13m^2$ was partitioned by a set of wooden screens and curtains in the Ergonomics Laboratory of the Department of Human Movement Studies, Rhodes University. Eleven African male volunteers, ranged in age between 27 to 56 years, were selected from the local working population as subjects for this study. Body mass, stature, hand grip strength, arm strength, back strength and leg strength were obtained. Each subject underwent a 45-min familiarization session relative to the experimental procedure, laboratory setting and equipment used in the study. Asymmetrical lifting was analysed in the sagittal plane (0^0), and at 30^0, 60^0 and 90^0 deviations ("task asymmetry") to the right of the sagittal plane, in each case from two initial grip heights (150mm and 500mm). The 10kg mass lifted in the present study was determined by applying the NIOSH guideline for manual lifting (NIOSH, 1981). A total of eight lifting tasks (4 task asymmetries x 2 vertical lift heights) were randomly assigned to each subject during data collection. The subjects were instructed to keep their feet in the sagittal plane and lift the load using both hands. Each subject was allowed to choose whichever lifting technique was found to be most comfortable.

The kinematics of the lumbar spine of the subjects were recorded using a Lumbar Motion Monitor and the movement of the load in three-dimensional space was monitored by a V-scope Ultrasonic Motion Analyser. When applying Newton's Laws of Motion for kinetic calculation, the load was recognized as a particle on which mass is concentrated. On completion of the dynamic analysis of lumbar motion and asymmetrical lifting, a three-dimensional biomechanical model of the lower back was developed.

1.4 LIMITATIONS

The following limitations must be borne in mind while examining the implications of the experimental results and subsequent conclusions:

1) The subjects were selected from a relatively small and ethnically homogeneous population, i.e. Xhosa-speaking manual workers in the Grounds and Gardens Section of the University.

2) Although the subjects were required to lift infrequently in order to minimize physiological fatigue, there was no control over the activities of participants immediately prior to experimental data collection.

3) The Lumbar Motion Monitor (LMM) was mounted on the subject's back throughout the experiment. The weight of the LMM backpack together with its harnesses (approximately 30N) was neglected in the lifting analysis.

4) The motion of the lumbar spine was defined as the movement of the first lumbar vertebra relative to the first sacral vertebra. That the

LMM validly reflects this was assumed, not measured, for reasons discussed in Chapter 3.

5) Since the subject's feet were restricted to the sagittal plane while executing asymmetrical lifting, the movements elicited were, as a consequence, not in a strict sense freely chosen.

6) The moment of inertia of the load was not considered in the 3-D force calculation due to the angular motion of the load being a small component compared to the linear movement.

CHAPTER 2 : REVIEW OF LITERATURE

2.1 INTRODUCTION

The act of manually lifting an object has been of on-going concern to professionals from a number of disciplines, including engineering, ergonomics, rehabilitation, biomechanics and management. Research in the area of manual materials handling has been conducted for over a quarter of a century and continues at an accelerated rate today. Basically, these researchers have sought to establish acceptable lifting limits, using several different approaches and have attempted to apply ergonomic principles to job design, employee placement and employee training.

In the early stages of this research a particular concern was shown for women and children performing manual lifting. Between 1930 and 1950, laws specifically limiting the weights that women and children could handle were enacted in the United States (NIOSH, 1981). These state regulations have been struck down as unconstitutional because they protectively discriminated against all women, without recognition of the large variation in capabilities among women. In 1962 the International Labour Organisation (ILO) suggested limits for occasional lifting. These limits were primarily based on inspection of injury and illness statistics. In consideration of the epidemiological, biomechanical, physiological and psychophysical criteria, a work practices guide for manual lifting was published in 1981 in the United States.

Lower back pain may result from one of a variety of pathological causes, or it may be triggered by one or more factors (Magora and Taustein, 1969). Rowe (1969) conducted a long-term study of low back disability in industry, involving over 1000 men, which indicated that degeneration of the intervertebral disc is the prime cause of symptoms in about 70% of working men with chronic or recurrent backache of sufficient degree to constitute a disability or placement problem. Disorders of the apophyseal joints are also considered to be important causes of back pain (NIOSH, 1981). Andersson (1981) reported six vocational factors as being important to lower back pain. They are: (i) heavy physical work; (ii) static work postures; (iii) frequent bending and twisting; (iv) lifting and forceful movements; (v) repetitive work and (vi) vibration. These factors are all similar in that they increase the load on the spine. They are often present at the same time, making the identification of the effect of each factor difficult.

It is believed that the stresses undergone in the spine are greatest at the levels of the fourth and fifth lumbar vertebrae. Also the sacral base is not normally oriented horizontally, producing a sharp curve in this region and giving a wedge shape to the intervertebral discs. This wedge shape may concentrate the stresses posteriorly on the disc thereby predisposing the lower lumbar discs which are particularly liable to damage.

Jayson (1981) pointed out that the disc itself is not innervated so that disc damage **per se** could not produce back pain. Pain felt in the back is often presumed to be due to the pressure on the nerve roots caused by a rupture or bulge of the annulus fibrosus of the disc. However, radiculopathy is but one of a dozen or more possible causes. It is often difficult to localize the source of symptoms felt in the back since the nerves around the spine are cross-innervated in an extremely complex fashion, and many deformable soft-tissue supportive structures also have sensory nerve supply.

Particularly because pathological models, by and large, fail to explain how back pain occurs it is imperative to act prophylactically to reduce the workplace risk factors associated with it. Risk can be greatly reduced if jobs are designed taking the human component into consideration. Ayoub and Mital (1989) defined three goals of job design: eliminating or reducing MMH, decreasing job demands and minimizing bodily movements. Snook (1978) concluded that proper design of manual materials tasks can eliminate up to 33% of back injuries. The average inter-individual variability in work capacity is much greater than the variability due to job characteristics (Garg and Saxena, 1980). Consequently it may not be feasible to design a job to fit a large population, which means that ergonomic methods of determining an individual worker's lifting capability are needed to match the worker's physical capabilities to the jobs' physical requirements. Back x-ray films, strength testing, medical examinations, psychological tests, job simulations and rating methods have all been used for employee placement purposes. Although these methods are not adequate, several such procedures have become common practice for pre-employment screening (Chaffin, 1974; Chaffin **et al.**, 1978; NIOSH, 1981; Pytel and Kamon, 1981; Bigos and Battie, 1987; Snook, 1987; Stewart, 1987).

2.2 MANUAL LIFTING ACTIVITY

2.2.1 Man-Machine-Environment System

A "system" is an aggregate of interactive components operating together to perform a function (Fraser, 1989). In the industrial situation man appears as a component of many systems. A man-machine system can be defined as an aggregate of one or more human beings and one or more physical components interacting to bring about, from given inputs, some desired output (McCormick and Sanders, 1982). The word "machine" here should be considered to consist of virtually any type of physical object, device, equipment, facility or thing that people use in carrying out some activity that is directed toward achieving some desired purpose or performing some function. A man-machine system can be as small and simple as one worker with a hammer, or as large and complicated as a petro-chemical complex with all its personnel. Man may play different roles in a man-machine system: (i) man as labourer; (ii) man as controller; (iii) man as monitor (Xie, 1987).

A man-machine system exists within an environmental context. The environment consist of physical space, e.g. workstation and ambient environment, e.g. illumination, atmospheric conditions and noise. McCormick and Sanders (1982) note that although the nature of people's involvement with their physical environment is essentially passive, the environment tends to impose certain constraints on their behaviour (such as limiting the range of their movements or restricting their view), or to predetermine certain aspects of behaviour (such as stooping down to look into a filing cabinet or wandering through a supermarket labyrinth in search of a single item).

Ayoub and Mital (1989) consider that manual materials handling (MMH) operations are systems which consist of three main components: (i) worker; (ii) task; (iii) environment. Any mismatch among these three components will lead to inefficient system behaviour which must be tolerated by its human component, often at great cost, suffering and pain, if the system is to remain operational. Since the components of a system are interactive, each properly functioning system has a certain equilibrium state. This equilibrium state is determined by the interactions that occur among the components; the effectiveness of the function is determined by the stability of the equilibrium. In a MMH system, the equilibrium state is mainly determined by matching relationships between the job demands and personal capabilities under a specific environmental conditions.

Following systems principles, the author's feedback MMH system is depicted as in Figure 1.

The inputs of the system are:

(i) job factors such as load, frequency and distance moved;

(ii) personal factors such as physical status, age, sex, stature and strength, and;

(iii) environmental conditions such as temperature, humidity, illumination, noise and vibration.

There are two different types of system output. One involves production output such as productivity and quality of work, and the other involves human output such as physiological responses of operator, biomechanical stresses on the musculoskeletal system of humans, etc. These system outputs can be quantified by various technologies. The physiological responses, for example, can be measured in terms of heart rate or oxygen uptake, under experimental or industrial conditions.

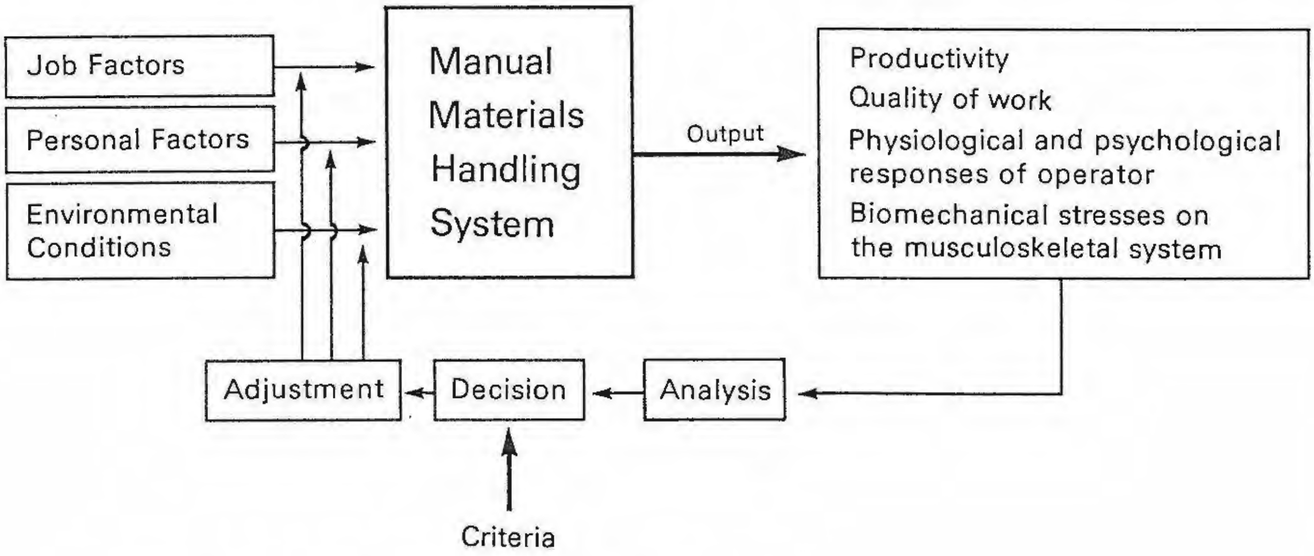

Figure 1 MMH feedback system

In the phase of decision making, comprehensive measurements modifying risk or improving productivity and quality are worked out depending upon the analysis of results and criteria such as safety and health regulations, benefit/cost and standards. Adjustments will be carried out to alter one or more specific job factors and/or personal factors as well as environmental conditions. A stable equilibrium of the system can be obtained through continuous adjustments.

In order to design an efficient MMH system, it is critical to understand the relationships between the elements of the three system inputs and the extent of human tolerance. Only when all cause and effect relationships become known is it possible to design an orderly MMH system structure.

2.2.2 Job Factors

Many aspects of the physical act of manually lifting a load have been identified as potentially hazardous to a person's musculoskeletal system. Herrin **et al.** (1974) and Chaffin and Ayoub (1975) concluded that the job factors which contribute to MMH hazards are as follows:

1) Load — measures of mass, force requirement, mass moment of inertia.

2) Dimension — measures of size of unit load (such as height, width, breadth of a box).

3) Location/site — position of load centre of gravity with respect to that of the worker.

4) Frequency/Duration/Pace — temporal aspects of the task in terms of repetitiveness of handling.

5) Stability — constancy of location of load centre of gravity when handling bulky or liquid materials.

6) Coupling — texture, handle size and location and shape.

7) Workplace Geometry — spatial aspects of the task in terms of movement distance, direction, obstacles and postural constraints.

Job factors include the elements which describe the MMH activity. Some of these factors are related to the object being handled, such as weight, shape, size and stability, while others describe the job itself such as frequency, distance moved and posture required. Ayoub and Mital (1989) further break down job factors listed above and summarize the major studies which have investigated designated job factors affecting material handling capabilities of workers. Three basic approaches, psychophysical, physiological and biomechanical, have been used to quantify the effects of job factors on worker's MMH capability. Table I summarises the conclusion of Ayoub and Mital (1989) regarding net effects of job factors on personal responses in manual lifting. The net effects indicate the material handling capability of workers.

Table I Net effects of job factors on personal responses
(Adapted from Ayoub and Mital, 1989)

Net Effects/Factor	MEER	HR	AW	WR	PE	SS	IAP
Frequency	+	+	–	+	+		
Task Duration	–	– or 0	–	–			
Object Size	+	+	–			+	
Object Shape			+				
Couplings	–		+			–	
Object Weight	+	+			+	+	+
Load Stability	0	0	–		+		
Vertical Lift Height	+	+	–		+		
Height of Start Point			–				
Asymmetrical Lifting	0	0	–		+	+	+

NOTE: MEER — Metabolic Energy Expenditure Rate
HR — Heart Rate
AW — Acceptable Weight
WR — Work Rate
PE — Perceived Exertion
SS — Spinal Stress
IAP — Intra-Abdominal Pressure
+ : Increase effect
– : Decrease effect
0 : No effect

Load lifted

The weight manually lifted is the most obvious and important factor which has been commonly accepted to be a composite index suggested as a limit of manual lifting (ILO, 1962; NIOSH, 1981). In 1988, the ILO published a document presenting the maximum weights in load lifting and carrying in various member states. The limits for manually lifted loads adopted by ILO member states ranged from 8kg to 90kg for adult male workers. Many countries provided more detailed limitations for women and young workers, and children. However, the majority of these limits take no consideration of the variability of individual capacity.

Rowe (1969) conducted a medical record survey of 2000 men, half of whom were sedentary and half routinely engaged in heavy handling work, and found that during the ten year survey period, 35% of sedentary workers and 47% of the heavy handlers made visits to medical departments for lower back pain. Magora and Taustein (1969) found that heavy industry workers had the highest incidence of lower back pain, followed by nurses, a group in which hard physical effort is required and back stress occurs. Chaffin and Park (1973) monitored 400 workers for a one-year period. It was concluded that the lifting of loads greater than about 16kg when held in close to the body, or equivalent conditions such as 9kg between 640mm and 890mm in front of the body, would be potentially hazardous for some people.

In a later study of 550 workers followed over a two-year period (Chaffin **et al.**, 1977), it was found that heavier jobs resulted in increased severity of injuries in terms of total lost workdays or medical work restriction days. In general, load handling of less than about 20kg resulted in relatively few incidents of severe strain or sprain diagnosis, but the heavier load handling jobs were associated with more severe sprains, joint dislocations and bone fractures.

Frequency and duration of lifting

Frequency is a critical job factor. As a factor in physical fatigue, task frequency is relatively more important than the characteristics of the load handled (Ayoub and Mital, 1989). Jobs were classified by NIOSH (1981) in three categories: "infrequent", "occasional high frequency" and "continuous high frequency". For infrequent lifting a person's musculoskeletal strength and potential high stress to the back are the primary limitations to ability. As such, biomechanical variables are predominant in determining hazard. Occasional high frequency lifting results in psychophysical stress and possible muscle fatigue as the primary limitation. For continuous high frequency lifting the limiting factor is "service organ" function and aerobic capacity.

Chaffin **et al.** (1977) studied the relationship between factors such as frequency, duration and pace of lifting, and back injury. They found that high frequency load lifting is related to increased injury rates. The more frequent the lifting of maximum loads on a job, the greater the frequency and severity rate of musculoskeletal problems and the greater the severity of traumatic injury. NIOSH (1981) has made a suggestion of a greater exposure to physical stresses during repetitive lifting which could accelerate "wear and tear" in connective tissues, a greater potential for muscle fatigue with repetitive lifting and a greater probability of an uncoordinated muscle action during a lift.

For a given task, the maximum weight of lift acceptable to an individual decreases non-linearly with an increase in rate (Asfour **et al.**, 1984 a,b; Garg and Saxena, 1982; Mital, 1984 a,b). Even though the maximum acceptable weight of lift decreases with pace, the total acceptable workload ($kg \cdot m.min^{-1}$) increases.

Ciriello and Snook (1983) investigated task frequency in the range from once every 5s to once every 8h and found that the maximum acceptable weight decreases, while heart rate and oxygen consumption increase with increases in lifting frequency. Karwowski and Ayoub (1984) reported that preferred loads at frequencies of 0.1 and 3.0 $lifts.min^{-1}$ were approximately three and two times greater, respectively, than those lifted at a frequency of 12.0 $lifts.min^{-1}$. For the spectrum of the frequencies(0.1, 3.0, 9.0 and 12.0 $lifts.min^{-1}$) used in their study, the relationship between the maximum acceptable weight of lift (MAWL) and frequency (FQ) was described by a power function:

$$MAWL(kg) = 30.504 \cdot FQ^{(-0.2208)}$$

Mital and Manivasagan (1983) reported that the decrease in mass lifted with an increase in frequency is an exponential function, expressed as follows:

$$Mass(kg) = 22.05 \cdot e^{(-0.033 \cdot FQ)}$$

The comparison between these two formulae is presented in Figure 2. Generally the power function produces a higher predicting value than the exponential function. MAWL computed via the power function decreases curvilinearly. Since the frequency has a very small weighing factor (-0.03) in the exponential function the mass decreases linearly when lifting frequency increases within the range 0 to 15 lifts.min^{-1}. In fact this exponential function can be replaced by a simple linear function as follows:

$$\text{Mass(kg)} = 22.05 - 0.533 \cdot FQ$$

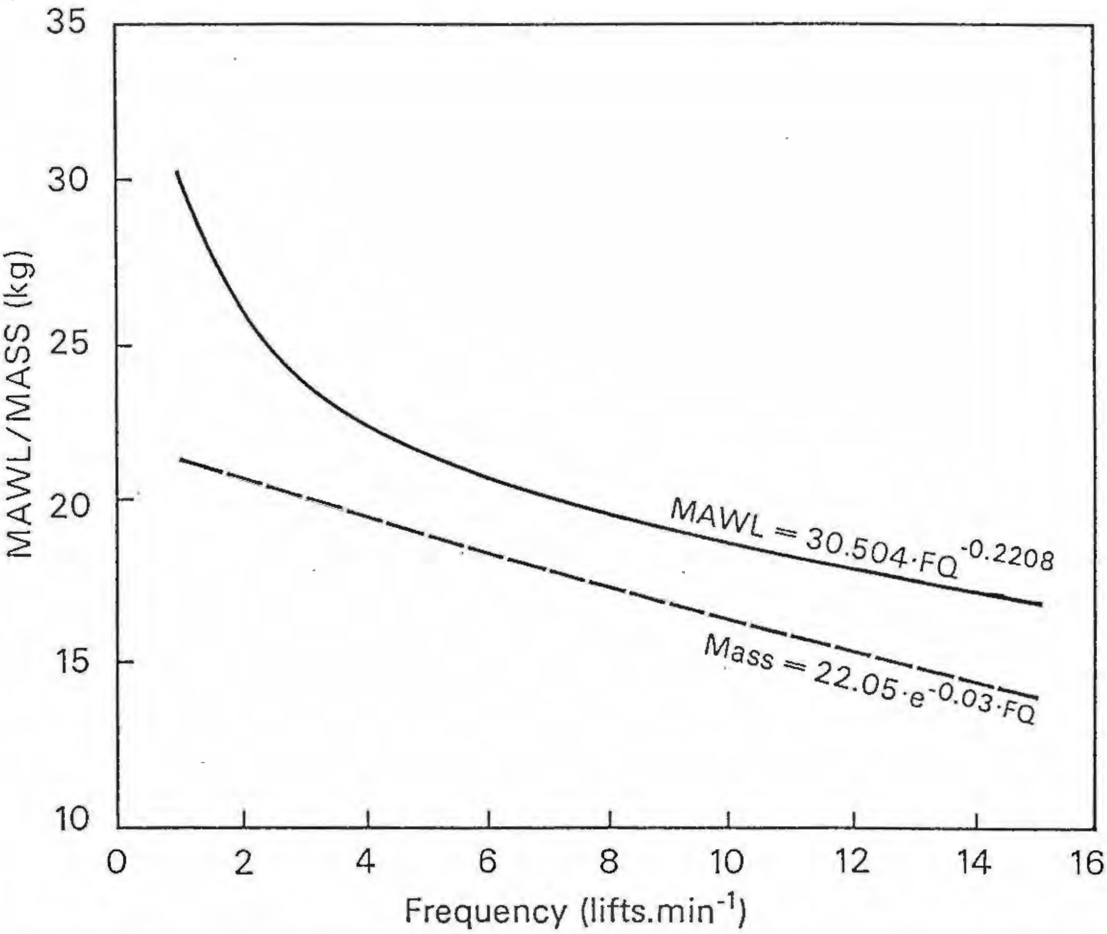

Figure 2 **Comparison between the predictions via power and exponential functions**

Garg and Saxena (1982) studied the maximum acceptable frequency of one-handed lifts. The maximum acceptable frequency for one-handed manual lifting tasks has been determined to be 50% of the maximum frequency that can be maintained for a period of 4min. For two-handed lifting tasks, the pace most acceptable to workers has been found to be 4 lifts.min^{-1} (Snook and Ciriello, 1974). Both the weight of the load and lifting distance had a significant effect on maximum acceptable frequency of manual lifting.

Endurance is the ability to sustain a specific activity continuously. Endurance is related to fatigue such that low endurance in prolonged physical activities is related to the aerobic work capacity of the individual (Deivanayagam and Ayoub, 1979). Thus in the case of a given healthy, well motivated individual, the endurance time on different tasks will depend on the energy output requirements of the task. The worker's energy expenditure level increases gradually with an increase in task duration. Such a gradual increase may be due to one or more of the following factors (Deivanayagam and Ayoub, 1979):

1) A progressive accumulative effect of the products of metabolism.

2) Changes in blood-flow distribution to various parts of the body other than the working muscles.

3) Deterioration of mechanical efficiency.

4) Changes in the constitution of metabolic substrate involved in the energy release process.

Mital (1983) reported that the maximum acceptable weight of lift (MAWL) decreases with time for all lifting frequencies (1, 4, 8 and 12 $lifts.min^{-1}$). He also found that the decrease in weight with time was non-linear. The average rate of decrease was 3.4% per hour for males and 2% per hour for females. Regression equations were developed for males and females respectively.

The results of a study conducted by Karwowski and Yates (1986) show that the time effect on the MAWL was not significant for low lift rates. However, there was a significant decline over time in the MAWL with a rate of 12 $lifts.min^{-1}$. The following equation was derived:

$$\text{Mass(kg)} = 9.304 - 0.00908 \cdot \text{Time(min)}$$

According to the above equation, the maximum amount lifted by females decreased at a rate of approximately 5.85% per every hour between 30min and 4h.

Ciriello **et al.** (1990) found that the effects of task duration on psychophysically-determined maximum acceptable weights and forces were not significant for task frequencies of 4.3 lift.min^{-1} or slower. Asfour **et al.** (1991) investigated the endurance time for prolonged arm lifting. The results indicated that endurance time decreased with an increase in frequency or load.

Fernandez **et al.** (1991) studied the effect of time on psychophysical lifting capacity over extended periods and found the average decrease in weight for an 8h period for the two frequencies (2 and 8 lifts.min^{-1}) was to 85.4% of the original maximum acceptable weight of lift (MAWL) determined in a 25min period. It was reported that all subjects lasted the 8h period for lifting the MAWL determined in a 25min period with a lower frequency (2 lifts.min^{-1}). On the other hand, most of subjects (9 out of 12) withdrew from the experiment involving a higher lifting frequency (8 lifts.min^{-1}), complaining of soreness in the lower back, upper legs and shoulders.

According to Astrand (1960), the maximum oxygen uptake of an individual of 25 years age and in good health would be on average 45ml.kg^{-1}.min^{-1}. A decline in the metabolic energy expenditure rate with time was found by Mital (1983). It is generally contended that 33% of maximum aerobic capacity is the appropriate limit for 8h of sustained performance. Under such conditions, during any 8h or 24h of sustained performance period, the worker should not exceed an energy expenditure rate of 0.3016 or 0.1652 kJ.kg^{-1}.min^{-1}, respectively (Deivanayagam and

Ayoub, 1979). A prediction function of the endurance time based upon any given equivalent energy rate (E) is as given below:

$$T_{(min)} = 10 \cdot \exp(1.09 - E) \div 0.29332$$

Object size/location

The physical dimensions of the load handled are important from a biomechanical, physiological and psychophysical point of view (NIOSH, 1981). Chaffin **et al.** (1977) observed that the more remote the load centre of gravity from the body (due to either the bulk of the object being handled or the workplace layout), the greater the frequency and severity of musculoskeletal problems and contact injuries.

Holding the object close to the body instead of away from it reduces the forward bending moment on the lumbar spine because the distance from the centre of gravity of the object to the centre of motion in the spine (the load arm) is minimized. The shorter the load arm is for the force produced by the weight of a given object, the lower the magnitude of this flexion moment, and thus the lower the load on the lumbar spine (Nordin and Frankel, 1989). If objects of the same weight, shape and density but of different sizes are held, the lever arm for the force produced by the weight of the object is longer for the larger object, and thus the flexion moment on the lumbar spine is greater.

As shown in Figure 3, the distance from the fulcrum in the disc to the front of the abdomen is 150mm in both cases and the objects lifted have a uniform density and each weighs 200N. In case A the width of this cubic object is 200mm;

in case B it is 400mm. Thus, in case A the forward-bending moment acting on the lowest lumbar disc is 50Nm, as the force of 200N produced by the weight of the object acts with a lever arm (La) of 250mm (200N x 0.25m). In case B the forward-bending moment is 70Nm, as the lever arm (La) is 350mm (200N x 0.35m).

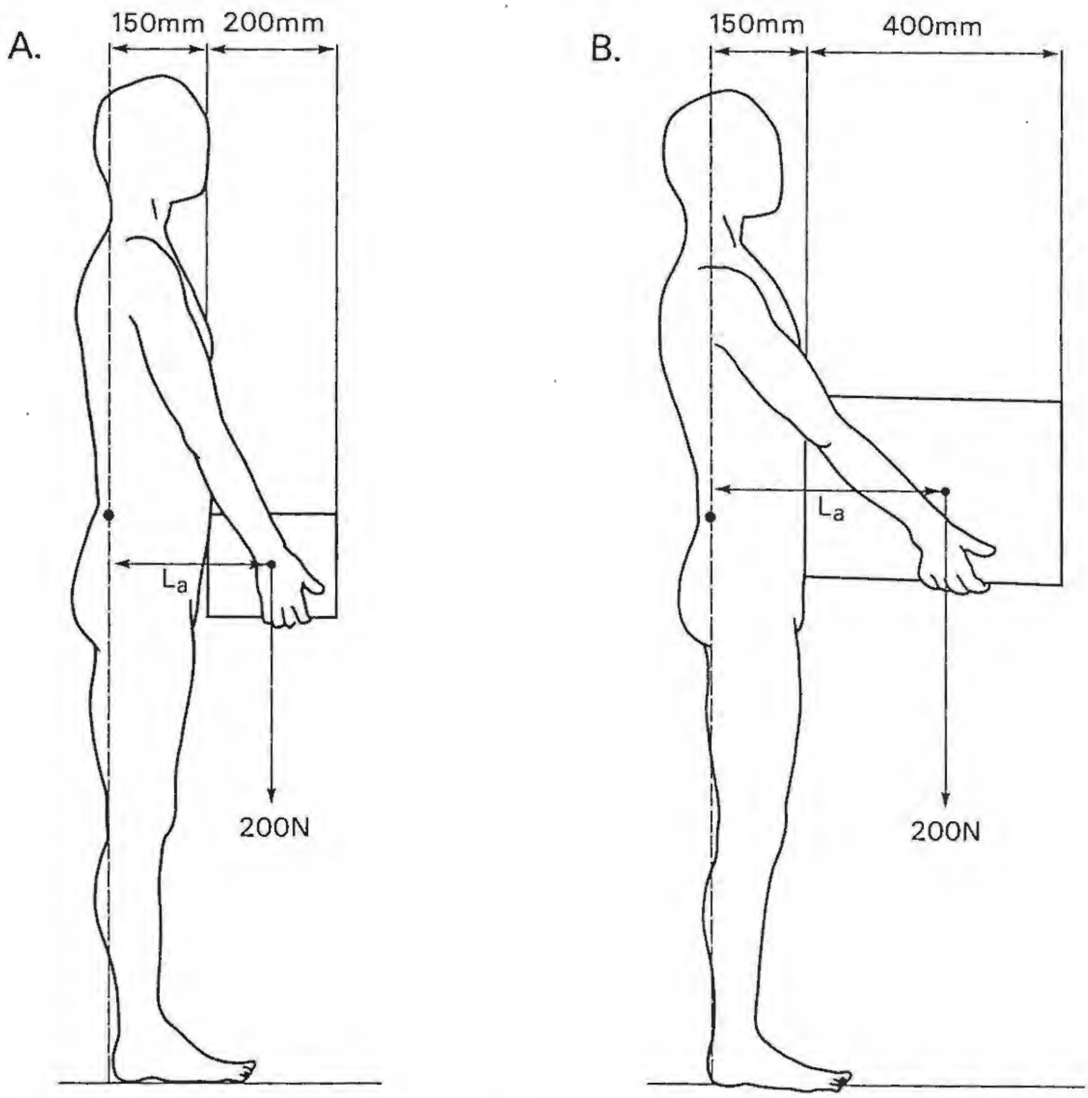

Figure 3 **The size of the object influences the loads on the lumbar spine**

Freivalds **et al.** (1984) used a biomechanical model for evaluation of job-related stresses imposed upon a worker during dynamic lifting. The results indicated that larger box sizes do not necessarily imply larger compressive forces, because of the decreased maximum loads selected by the subjects for the larger boxes. However, larger box sizes create larger moment arms and increase L_5/S_1 compressive forces as confirmed by two-way analysis of variance of F_{comp}/Load ratios. Similar results were observed by Mital and Kromodihardjo (1986) in three-dimensional biomechanical analyses of manual lifting. It was apparent that the acceptable weight of lift decreased significantly with box-size causing insignificant increases in average peak compressive force. The average peak shear forces also did not change significantly. It appears that individuals accepted less weight, as the box-size increased, in order to keep the stress level from becoming too severe. This indicated that, as the box-size increases in the sagittal plane, the lifting task becomes more stressful.

Mital and Ayoub (1981) reported that container dimensions in the sagittal and frontal plane influence the metabolic energy expenditure rate. Figure 4 shows the change in oxygen uptake for given container dimensions. According to this, container dimensions in the sagittal plane should not exceed 500mm. Any increase in container volume should be accomplished first by increasing its height, then its dimension in the sagittal and frontal planes.

Garg and Saxena (1980) reported that heavier weights can be lifted using bag containers (collapsable) than using boxes (non-collapsable). Smith and Jiang (1984) not only confirmed these findings but also found that these heavier loads are lifted at only a slightly higher physiological cost (Table II).

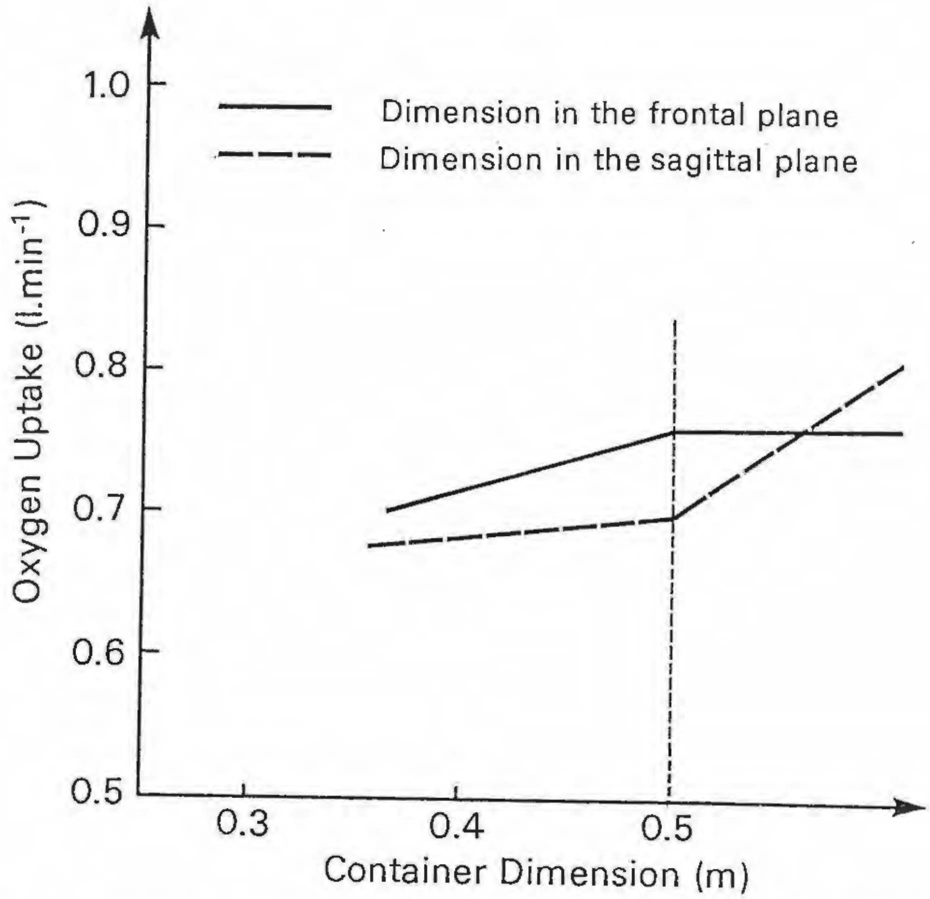

Figure 4 **Change in oxygen uptake with variations in box dimension in the frontal and sagittal plane** (Mital and Ayoub, 1981).

Table II **Comparison of the maximum acceptable weight of lift between collapsable (bag) and non-collapsable (box) containers** (Smith and Jiang, 1984).

Container	Load (kg)	Oxygen Uptake ($l \cdot min^{-1}$)	Heart Rate ($beats \cdot min^{-1}$)
Bag	24.26	1.50	129.7
Box	22.05	1.45	121.3

The human capability to lift decreases with an increase in horizontal displacement according to the biomechanical strength model developed by Martin and Chaffin (1972). The further the load distance from the body, the larger the compressive forces in the lower back (Figure 5).

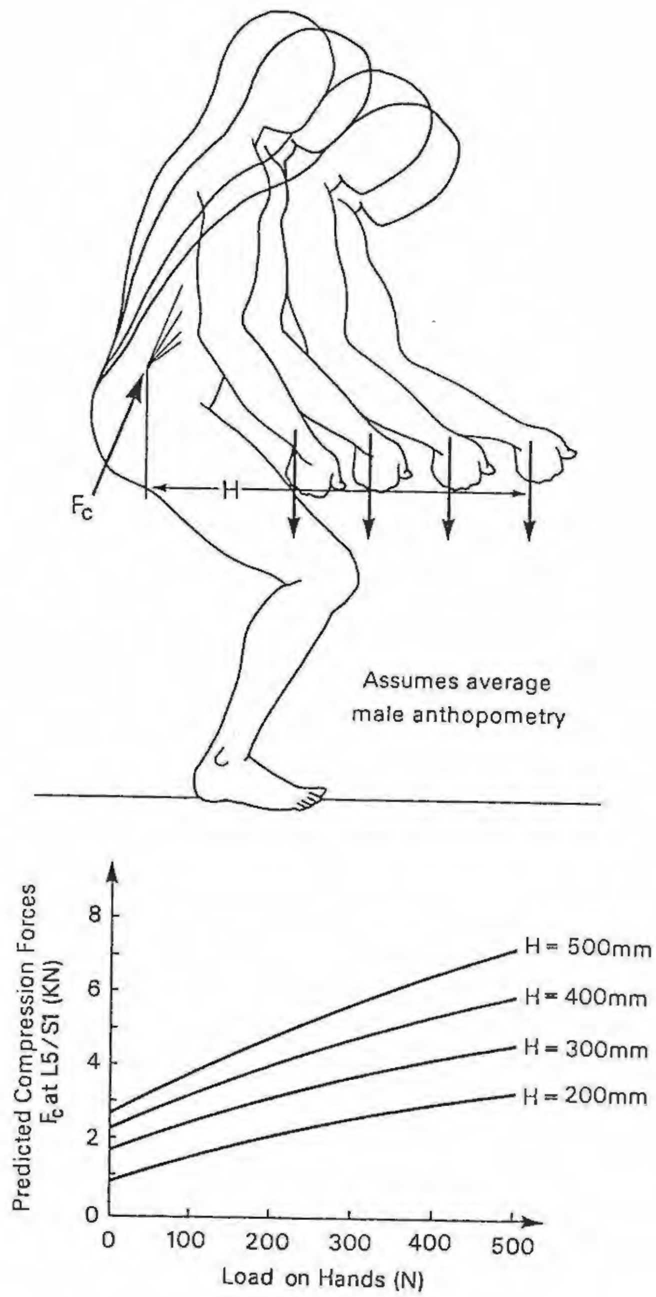

Figure 5 **Effect of moment-arm on disc Load** (Chaffin, 1984)

An individual's lifting capability decreases with the vertical height or vertical distance moved (Martin and Chaffin, 1972; Mital, 1984a.b; Snook, 1978; Ciriello and Snook, 1983). The starting point for lifting (or lowering) is important; for the same vertical distance, lifting capability can decrease with a change in the starting point of lift. This decrease in capability may be as large as 23% when the starting point of the lift is moved from knuckle to shoulder height (Ayoub and Mital, 1989). The height of the lift also affects the stress on the spine; lifting from the floor is more stressful than lifting from table height (Ayoub and Mital, 1989). The greatest compressive and shear forces occur during the first few fractions of a second of a lift from the floor (Freivalds **et al.**, 1984; Jager and Luttmann, 1989; Mital and Kromodihardjo, 1986). According to NIOSH (1981), load- lifting limits decrease with an increase in horizontal location and lift distance; the maximum limits are obtained while the load is located at 750mm above the floor.

Couplings

Box holding, handle positions and angles have been evaluated for physiological, psychophysical and biomechanical stresses. Using the definition of handle position and wrist angle as depicted in Figure 6, various conclusions have been reached (Deeb **et al.**, 1986):

1) Handle positions which are symmetrical (e.g. 2/2 and 8/8) produce the lowest forces at the handles and are associated in industry with heavy, compact boxes handled at floor level.

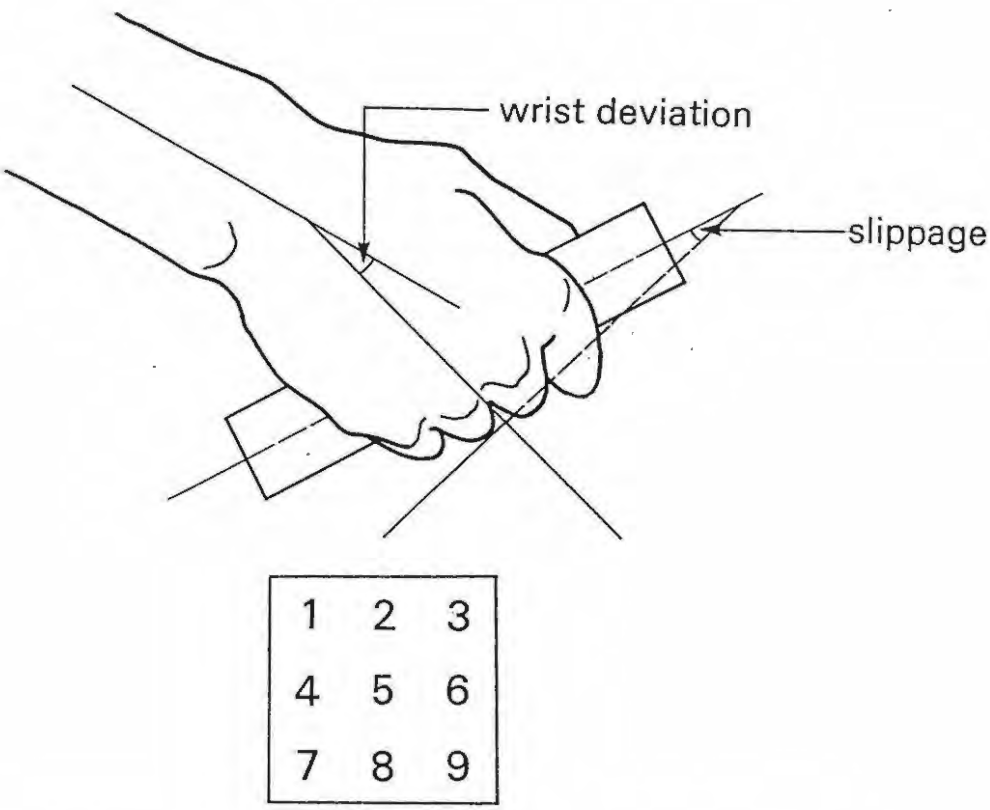

Figure 6 **Definition of handle position and angles measured**
(Adapted from Durry and Deeb, 1986a)

2) Handle positions which are asymmetric (e.g. 3/8, 6/8, 3/7) minimize perceived cost and heart rate and provide the horizontal and vertical stability required in handling large but lighter containers.

3) The hand accommodates to the handle angle both by deviating the wrist and by allowing slippage between hand and handle.

Handling loads with handles is safer and less stressful (Ayoub and Mital, 1989). Garg and Saxena (1980) observed a decline in the maximum weights acceptable to their subjects for lifting boxes without handles. Lifting boxes without handles requires additional metabolic energy (Mital and Ayoub, 1981).

Handles from 25 to 38mm in diameter, 115mm long, and with 30 to 50mm clearance all around are favoured in many different studies (Drury and Pizatella, 1983). In manual materials handling of boxes, the handle combinations of 6/8, 3/8 and 2/2 are optimum box handle positions for minimizing biomechanical, physiological and psychophysical stresses (Drury **et al.**, 1989 a,b; Drury and Deeb, 1986 a,b; Drury and Pizatella, 1983; Coury and Drury, 1982). Drury and Deeb (1986b) found position 2/2 best for these particular movements between floor and waist height, but found other positions such as 3/8, 6/8 and 3/7 better at other working heights.

There are different roles for each hand in maintaining box equilibrium. The higher hand appears to provide the horizontal stabilizing component while the lower hand contributes the vertical component (Coury and Drury, 1982). The handle positions 3/8, 6/8 and 3/7 have the greatest spatial separation, they make the best use of vertical and horizontal stabilizing components (Drury **et al.**, 1989 a,b; Drury and Pizatella, 1983). The position 8/8, where both hands are under the box, minimizes forces exerted and is used in special box movements, particularly with heavy boxes (Drury and Pizatella, 1983).

Deeb **et al.** (1986) studied the effect of straight and curved handles on box handling and found that a curved handle was no better than a straight handle in a lifting task. The handle angle required will vary with the height at which the box is lifted or held. At floor height handles should be nearer to the horizontal, while at waist and shoulder heights the handles should be nearly vertical.

The presence of handles on the boxes has an effect on compressive and shear forces on L_5/S_1 of the human spine. Freivalds **et al.** (1984) observed that boxes without handles resulted in slower rise times and lower peak compressive force, indicating more controlled motion. Mital and Kromodihardjo (1986) reported that the subjects accepted significantly heavier weights when lifting boxes with handles; a clear indication that lifting boxes without handles is more stressful. The ratio of compressive and shear forces to load when lifting without handles is larger than with handles.

2.2.3 Personal Factors

The worker's personal factors consist of all elements that describe a person's working capability. Many individual variables have been recognized that limit a worker's manual lifting capability. Ayoub and Mital (1989) noted that the effects of some of these variables have been found to be consistent, while at the same time there exists both conflicting and contradictory information on other variables. According to Chaffin and Ayoub (1975), the various personal factors which limit a person's lifting capability are as listed below:

1) Physical — general worker measures such as age; sex; physique.

2) Sensory — measures of worker's sensory processing capabilities.

3) Motor — measures of worker motor capabilities.

4) Psychomotor — measures of worker's capabilities in interfacing mental and motor processes.

5) Personality — measures of worker's value and job satisfaction.

6) Training/Experience — measures of worker's education level in terms of formal training or instruction in MMH skills, informal training and work experience.

7) Health status — measures of workers' general health profiles.

8) Leisure time activities — measures of the person's choosing to be involved in physical activities.

The physical factors such as age, sex and anthropometry are relatively more important in determining personal risk of injury in MMH.

Sex

Sex is perceived by some as being the most important personal characteristic that divides the working population because of differing anthropometric, biomechanical and physiological variables between males and females. Many lifting guidelines and norms recommend that women should not lift as much as men (ILO, 1962; Snook, 1978; NIOSH, 1981; Genaidy, **et al.**, 1990). On average female lifting strength is about 60—70% that of their male counterparts (Martin and Chaffin, 1972; Chaffin, 1974; Snook, 1978; Yates **et al.**, 1980; Mital, 1984a,b). Therefore, if asked to handle a given load, the average woman is more highly stressed than the average man (NIOSH, 1981). However, the range in strength of both males and females is very large. According to Yates **et al.** (1980), the maximal isometric lifting strength of women is approximately 50% of their male counterparts at lower lifting heights; in the higher lifting heights the relationship is reduced to 33%.

Stevenson **et al.** (1990) conducted a dynamic analysis of maximum strength on an incremental lifting machine (ILM) and reported that maximal ILM scores for females were 53% of those for males. The aerobic capacity of women is significantly different from that of men (NIOSH, 1981; Ayoub and Mital, 1989). Female aerobic capacity is, in general, 70—75% of male aerobic capacity (Astrand and Rodahl, 1977). At submaximal workloads, the metabolic cost of manual lifting is significantly smaller in women than in men (Mital, 1984a,b). Ciriello and Snook (1983) noted that although maximal acceptable weights lifted by males were significantly different from those of females, both sexes chose weights that produced similar cardiovascular strains.

Brown (1974), in a survey of industrial workers, reported that women appear to have larger relative numbers of complaints than men when required to perform heavy physical jobs. However, Chaffin and Park (1973) studied both men and women performing equally demanding light-to-moderate load handling jobs and reported an equal incidence of lower back pain cases.

Age

It is generally agreed that older workers have diminished capacity to withstand physical stresses (Charness, 1985). While physical capability declines after age 50—60yr., this does not appear to lead to a decrease in manual lifting capabilities (Ayoub and Mital, 1989). In a study of 74 male and female industrial workers, ranging in age from 18—61 years, Mital (1984a,b) observed no differences in the maximum acceptable weight of lift due to age. Indeed, Mital and Ayoub (1980) observed that isometric arm and leg strength increases with age.

A comprehensive clinical study of low back disability was conducted by Rowe (1969) which revealed that more than 70% of patients with lower back pain were in their thirties and forties; only 12% were in their twenties and 13% in their fifties. In a study of 3316 workers taken from 8 basic occupations, Magora and Taustein (1969) reported that 21.2% of lower back patients were in the age range up to 30yr, 32.4% in the range 31—40yr, 26.9% in the 41—50yr range and 19.5% in the age of 51yr and over. It is possible that the younger person may not have developed the requisite ability to recognise and control the hazards of MMH in the manner of the older worker. Job assignments may be based on age and a self-selection process may occur on many physically demanding jobs, which leaves the job for workers with capacities better suited to the job demands (Ayoub and Mital, 1989). Although older workers have perfected their skills in handling heavy loads they are likely to experience diminished physical capabilities, so age should be treated as a potential risk factor in manual lifting (NIOSH, 1981; Ayoub and Mital, 1989)

Body weight

Workers performing a load lifting task also lift their partial body weight, particularly at lower lifting heights. A heavier worker lifts more total weight and is physiologically more stressed than a lighter worker under the same task demands. It is generally accepted that body weight exerts a direct effect on the metabolic energy expenditure rate of a person while lifting (Garg **et al.**, 1978). This means that a heavier person tires more readily, leading to earlier onset of fatigue or cardiovascular problems. On the other hand, a heavier person is usually stronger than his lighter counterpart and usually has the mass necessary to counter-balance the handling of large objects. The body weight has extensively been used in predicting the maximum acceptable weight of lift (Mital and Ayoub, 1980, Yates **et al.**, 1980; Mital and Manivasagan, 1983; Karwowski and Ayoub, 1984).

Stature

Tauber (1970) indicated that taller people experience more lower back pain incidents than shorter people, but this contention was not supported by the studies conducted by Rowe (1971) and Chaffin and Park (1973). A taller worker has to stoop farther to pick up or set down a load, resulting in a relatively larger moment on the low back due to the weight of the load and body. On the other hand, it is stressful for shorter workers to handle loads away from the body or above their normal reach.

Li (1992) has employed a simple static biomechanical model for considering stature as a risk factor in manual lifting tasks. It has been found that the difference of truncal flexion angle between persons of different stature, while performing the same lifting task, dramatically increases with an increase in the initial vertical lift height. For the population at large, taller persons have larger segmental links and body masses than shorter individuals. This causes greater compressive and shear forces in the L_5/S_1 disc of a taller person. These differences between persons of different stature increase with increases in the initial vertical height of the lift.

Strength

Human strength has been widely applied for predicting manual lifting capability (Ayoub and Mital, 1989). Chaffin and Park (1973) and Chaffin (1974) studied human isometric strength to determine the potential relation to the incidence rate of lower back pain. The results disclosed that lower back pain episode rates sharply increases for those workers who do not have adequate isometric strength to perform lifting tasks. In a later study, Chaffin **et al.** (1978) not only confirmed these results, but revealed that the severity rates of lower back pain are much higher on jobs in which the demands exceed the strength of the workers. The need for utilization of some form of a strength testing program was recommended when placing people on jobs with a significant MMH component.

Well established isometric strength testing procedures have developed (Chaffin, 1974; NIOSH, 1981) and measurement of isometric strength has been successfully used in predicting the manual lifting capability of individuals. Mital and Ayoub (1980) used individual anthropometric parameters as independent variables to predict various isometric strengths. Computerized simulation for the prediction of human capabilities has been developed (Martin and Chaffin, 1972), but isometric muscle strength measurement is not a sufficiently complete and realistic assessment of the muscular abilities actually required in dynamic MMH activities (Kroemer, 1983). Most lifting in industry is performed dynamically. Therefore dynamic techniques, rather than static strength measurements, should be used to test a person's response to lifting. Diverse types of dynamic techniques have been developed including isokinetic strength and isoinertial strength measures (Kroemer, 1983; Ayoub and Mital, 1989). The reliability of the measurement of dynamic strength for determining a person's lifting capability has been confirmed by many researchers (Garg **et al.**, 1980; Pytel and Kamon, 1981; Kroemer, 1983). An experiment was conducted by Aghazadeh and Ayoub (1985) for comparison of models for prediction of lifting capability of individuals, incorporating static and dynamic strength. They concluded that with regard to accuracy, safety and speed of testing, the dynamic procedure may be superior to static testing for prediction of weight lifting capability.

Posture

Posture has been defined by some as the configuration the body assumes to initiate an activity (Ayoub and Mital, 1989). Traditionally, workers have been trained to "bend knees, and keep the back straight" when lifting. This recommendation is based on a simplistic mechanical logic that this posture minimises the spinal bending moment and therefore the compression forces on the back (NIOSH, 1981). If the load is small enough to be lifted between the knees, this "leg lifting" technique may be justified, if it reduces the stresses on the lower back due to shortening the moment arm of the load. Unfortunately, lifting this way from a squatting position with the back vertical often is not possible if the worker does not have the quadriceps strength necessary to extend the knees and raise the body from such a position. In other words, most people when lifting weights, lean the torso forward to reduce the moment on the knees.

In everyday situations most objects are so large that they cannot pass between the knees. When a large object is lifted around the front of the knees using the leg lifting technique, it necessarily causes the moment arm of the load about the lower back to be large. As a consequence high spinal muscle forces and lumbar compressive forces are produced. In contrast, the "back method" of lifting may allow a worker to bring the load closer to the body thus reducing the load moment and decreasing the compressive force on the lower back (Figure 7).

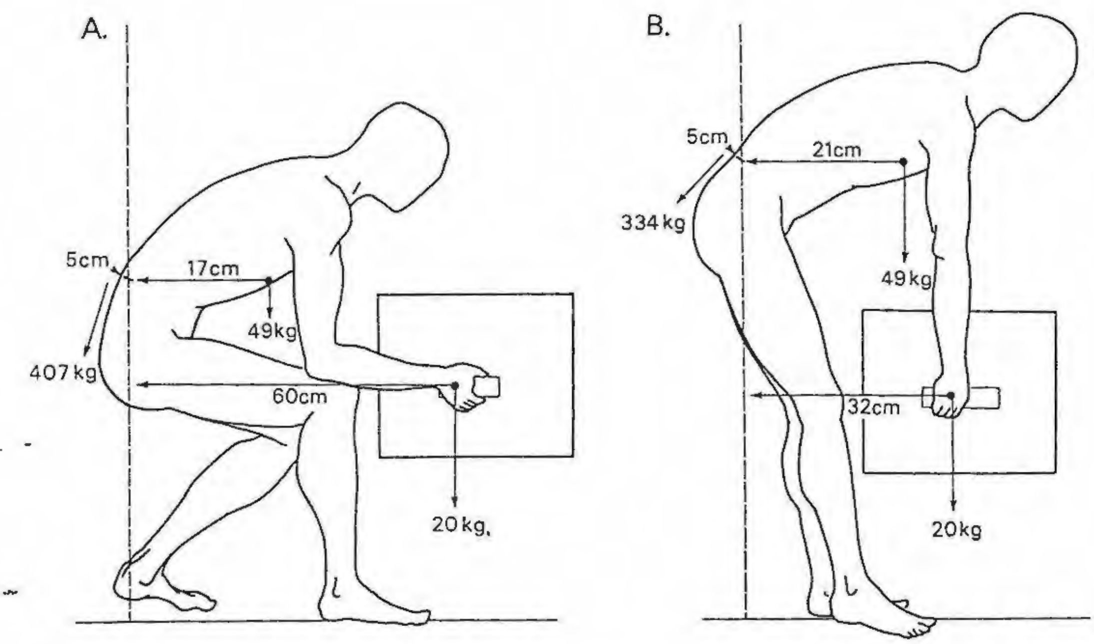

Figure 7 Lower back compression associated with torso lifting posture (Modified from Charteris, 1992)

A further limitation on lifting large objects with a squat lift arises from the fact that the arms must be extended farther forward in the horizontal (reach) direction than with the stoop posture (NIOSH, 1981). In this position, a high torque will be produced at the shoulders, and the shoulders may not have the strength to move the load upward. Andersson(1970) suggested that it is safer to allow workers to use their own common sense and muscle sense than to teach them new drills in performing certain jobs in which a series of predetermined positions must be consciously assumed.

Leskinen **et al.** (1983) analysed the dynamic compression on the spine under different lifting techniques. The leg-lift technique produced the lowest peak compressive force on the lower back. The compression × time integral over the accelerative phase of the lift, relating to the total stress of a lift, was smallest in the back-lift. Mittal and Malik (1991), studying Indian females, reported that the maximum weight was lifted with the back flexed and knee extended. Squatting (a posture used by Indians for performing household chores) produced the least biomechanical stress. The physiological cost of three different methods of lifting has also been studied by Kumar (1984). The stooped lifting technique was found to be physiologically least, and the squat method most demanding. The estimate was that the squat method will cost 3558kJ more for females and 3683kJ more for males than the stoop method, over an 8-hour working day. The inspiratory volume was consistently higher for the squat method of lifting when compared with the stoop or free-style methods. The squat method of lifting was most tiring and the free-style least tiring of the three techniques studied. Welbergen **et al.** (1991) reported that in power-lifters, mean power output (W) and oxygen uptake (VO_2) are significantly greater for squat lifting than for stooped lifting at the same lifting frequency. The difference is about 20—26% for mean power output and 23—34% for VO_2. They concluded that the stooped lift appears twice as effective as the squat lift for displacing loads. Maximum acceptable weight for the free style method are greater than those for the squat lifting method (Garg **et al.**, 1983)

Nemeth and Ekholm (1985) performed a biomechanical analysis of hip loads during extended and flexed knee lifting. The compression force was calculated to vary between 2.6 and 3.2 times body weight for lifting a 126N box in the sagittal plane with flexed and with straight knees. At a constant hip joint loading moment, they found that the degree of hip flexion greatly influenced the compressive force, the least force being obtaining at 35^0 hip flexion.

2.2.4 Environmental Factors

It is well known that an individual's performance is affected by the working environment. Heat load influences a person's physiological and psychological behaviour causing decreased work-rate, irritability, carelessness and fatigue (Ayoub and Mital, 1989). Sengupta **et al.** (1979) reported that metabolic energy expenditure decreases with increase of temperature. In MMH operations, higher ambient temperature and humidity results in elevated heart rates (Kamon, 1979). The physiological responses to humid and dry heat are different for men and women. In general, women should have greater tolerance limits and relatively lower heart rates and rectal temperatures as compared to men (Avellini **et al.**, 1980). Snook and Ciriello (1974b) observed that the lifting capability declined by 20% at a Wet Bulb Globe Temperature (WBGT) of 27^{o}C compared to a WBGT of 17.2^{o}C. Apparently no studies have been carried out in respect of the effects of noise, illumination and vibration on manual lifting activities (Ayoub and Mital, 1989).

2.3 ASYMMETRICAL LIFTING

There are several kinds of asymmetrical lifting: (i) lifting while trunk twisting; (ii) lifting with asymmetric hand position; (iii) handling boxes not symmetrically loaded; (iv) handling boxes located out of sagittal plane, even though there is no obvious trunk twisting. All of these can lead to asymmetric loading of the musculature and the skeletal structure. Asymmetric lifting causes not only a lateral bending moment on the lumbar column, but produces a rotation of each vertebra on its adjacent vertebrae (NIOSH, 1981). Farfan **et al.** (1970) indicated that disc degeneration most often involves the annulus fibrosus, which is the structure that provides 40–50% of the torsional resistance to twisting of the lumbar vertebrae. With disc degeneration, this torsional resistance can be reduced to less than one-half its normal strength, thus providing a significant injury potential. Asymmetric loading of the musculature of the back could produce a concentrated stress of sufficient magnitude to strain a specific muscle. NIOSH(1981) suggested that a person's arm and shoulder strength are not well enough developed to lift heavy weight in an asymmetric fashion.

Asymmetrical lifting of objects is very prevalent in industry (Drury **et al.**, 1982). It has been recognised that asymmetric lifting is more hazardous to the musculo-skeletal system than symmetrical lifting because of the combined effects of flexion and axial rotation of the spine. Very few studies have been undertaken in this area, largely because of the complexity of the problem.

Kumar (1980) has studied physiological responses to weight lifting in sagittal, lateral and oblique planes. He found peak intra-abdominal pressures were consistently lower in sagittal lifts and higher in lateral and oblique lifts; the EMG responses of the erector spinae and external oblique muscles being higher for the oblique and lateral lift. Boudrifa and Davies (1987) not only observed similar intra-abdominal pressures when lifting asymmetrically, but found that the EMG from the left side of the lumbar spine was lowest when lifting from the left, while the EMG from the right side of the lumbar column was lowest when lifting from the right side. This is of course to be expected, because when lifting from one side the trunk muscles of the other side counter-balance the effect of gravity. Based on the findings of Kumar (1980) and Boudrifa and Davies (1987), increased physiological cost in lateral plane activity was expected. However, this hypothesis was not supported by some later experimental results.

According to Kumar (1984), there was no significant difference in net VO_2 and energy expenditure for the same activity in different planes. The activities performed in the 30^0 lateral plane were subjectively assessed to be more tiring than those in sagittal plane. Work in a plane at 60^0 to the sagittal was rated most tiring. Kumar explained the discrepancy found between the subjective assessment and the physiological cost on the basis of asymmetric stress and movement, for he noted a partial redundancy of the ipsilateral extensor muscles and total redundancy of the ipsilateral external obliques and rotators and the contralateral internal obliques. The asymmetry of the situation heightened the biomechanical and physiological demands on some muscles, thereby increasing the overall stress.

Warwick **et al.** (1980) reported a 30% decrease in maximum static strength when the subject was rotated 90^0 or 135^0 from the sagittal plane at shoulder height and more than a 50% decrease in static strength when the subject was rotated by 90° at knee height. Garg and Badger (1986) studied the maximum acceptable weights (MAW) and maximum voluntary isometric strengths (MVIS) for asymmetric lifting. Both MAW and MVIS decreased with an increase in the angle of asymmetric lifting. Correction factors of 7, 15 and 22% for maximum acceptable weights and 12,21 and 31% for static strength at 30^0, 60^0 and 90^0 of asymmetric lifting are recommended. An 8.44% decrease in maximum acceptable weight was observed by Mital and Fard (1986) when subjects rotated 90^0 while lifting. No significant difference in oxygen uptake or heart rate was reported when lifting in the sagittal or lateral planes. Asymmetrical lifting or lifting asymmetrical objects was verbally rated by subjects as physically more difficult.

Garg and Banaag (1988) studied acceptable weights, heart rates and RPE during asymmetrical lifting and revealed that even with reduced weights for asymmetric lifting, there were significantly higher circulatory demands and perceived stresses compared to situations involving equivalent weights moved in the sagittal plane. An important finding of the study was that relative to sagittal lifts, the percentage decrease in maximum acceptable weight for a given angle of asymmetric lifting is independent of lifting frequency and initial height of the lift; similarly the percentage decrease in static strength was found to be independent of height. Correction factors of 9, 14 and 21% for maximum acceptable weight and 17, 31 and 42% for static strength at 30^0, 60^0 and 90^0 of asymmetric lifting, respectively, are also recommended.

Vink **et al.** (1992) investigated the maximum back extension strength of 12 subjects in 23 postures. Twenty of those postures were asymmetrical. The results affirmed the decrease in maximal extension force up to 40% in asymmetric trunk postures. The magnitude of force reduction due to asymmetry was strongly dependent on the plane of rotation and the symmetric reference position. It was found that this reduction in force can be explained by the influence of three factors: the length of the lumbar back muscles, the activation of the lumbar back muscle and the angular moment of the lumbar back muscles, acting as group.

Mital and Manivasagan (1983) reported that the maximum acceptable weight of lift (MAWL) was significantly higher when the centre of gravity (CG) of the object lifted was placed towards the right side for right-handed subjects or towards the left side for left-handed subjects. On average, 2.8% more weight was lifted when the CG offset was in the direction of the dominant side. However, MAWL decreased significantly when the CG was offset from the midsagittal plane. A linear equation of the MAWL with an increase in CG offset was developed.

From a kinetic analysis of manual lifting, Mital and Kromodihardjo (1986) found that lumbar compression and ground reaction force were lower when lifting asymmetrically, but shear forces acting on the lumbar spine were increased. Gallagher **et al.** (1994) also reported a decreased compression with asymmetrical lifting in stooping and kneeling postures. While lumbar compression is decreased in asymmetrical lifts, this type of exertion is associated with a large proportion of lower back pain cases (Snook, 1978). This would imply that the traditional biomechanical criteria (compression) used for ergonomic design of lifting tasks may not be related to the higher incidence of lower back pain associated with asymmetrical motion (Gallagher **et al.**, 1994).

2.4 BIOMECHANICAL ANALYSIS OF MANUAL LIFTING

2.4.1 Biomechanics of Lifting

Lifting is a whole body motion involving almost every joint in the body. The performance of manual lifting exposes the lifter to a variety of biomechanical hazards. It is well-established that the stresses induced in the lower back during manual materials handling are due to a combination of weight lifted and method of handling it. The external load is applied at the hands and the effects of external load and the weights of body segments are transmitted to the feet through those segments. Kroemer (1983) pointed out that in the body segment chain, the lumbar segment is one of the weakest links.

The spine is not a single joint but rather a series of small joints with flexible articulations between them (Aspden, 1988; Liu, 1990). With proper geometric and physiological data, the forces in each disc during a specific lifting activity can be predicted. Since clinical and biomechanical data indicate the problem to be centred in the lower lumbar spine, the L_5/S_1 disc has been extensively used to represent the spinal stresses of lifting (Chaffin and Park, 1973; NIOSH, 1981; Chaffin, 1984; Schultz and Andersson, 1981; Potvin **et al.**, 1992; Jager and Luttmann, 1989; Kromodihardjo and Mital, 1986).

Figure 8 is a representation of a subject lifting a load and shows the forces involved. Based on the static posture, the torques due to the load (**W_L**), the weights of trunk (**W_T**) and arms (**W_A**), about the lower back cause the human trunk to flex. In order to counteract this forward bending torque, the muscles in the lower back region must check correspondingly higher forces as they operate on small moment arms (within range of 38–50mm).

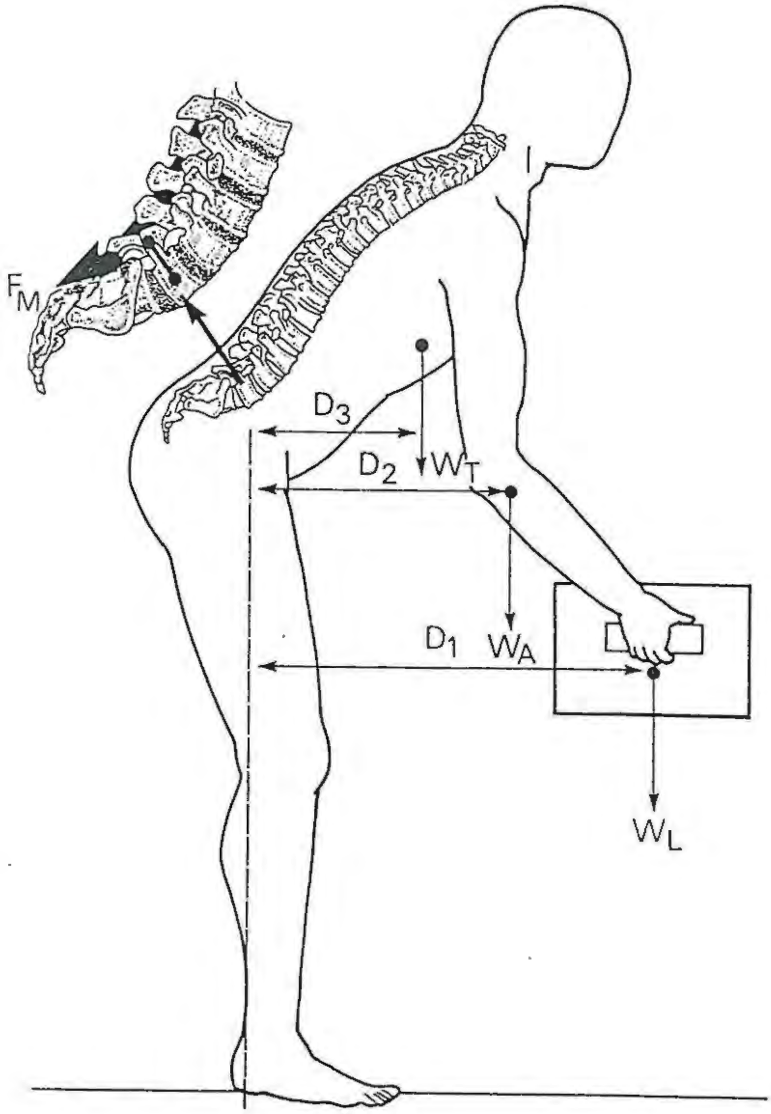

Figure 8 Forces Acting on the Lower Back in Lifting a Load

Thus, for equilibrium:

Bending moments = Extension moments

so:

$$W_L \cdot D_1 + W_A \cdot D_2 + W_T \cdot D_3 = F_M \cdot S$$

Where: W_L = weight of the load being lifted

D_1 = moment arm of the load

W_A = weight of the upper limbs

D_2 = moment arm of the upper limbs

W_T = weight of the trunk above L_5/S_1 disc

D_3 = moment arm of the trunk

F_M = back muscle force

S = moment arm of back muscle

The high forces generated by the lower back muscles are the primary source of compressive forces on the lumbar-sacral disc.

Assume the following:

$W_L = 98.1N$ $D_1 = 500mm$

$W_A = 78.5N$ $D_2 = 350mm$

$W_T = 412.0N$ $D_3 = 250mm$

If the distance of the action of the muscles from the base of the spine is taken to be 50mm, then the muscular force ($\mathbf{F_M}$) required to balance the forward bending moment can be found.

$$F_M \times 50 = 98.1 \times 500 + 78.5 \times 350 + 412.0 \times 250$$

$$F_M = 3590.5N$$

As can be seen, the force of the muscle must be very large to balance a weight of only 98.1N. The model presented above is a simplistic one which ignores the fact that lifting is a dynamic motion. If the dynamic aspect is considered the moments of inertia and angular accelerations would further increase the forward bending moment. Freivalds **et al.** (1984) reported that the dynamic effect tends to increase the L_5/S_1 compressive forces and amounts to increasing the static load by as much as 40% of its weight. The faster or more jerky the lifting motion, the greater the compressive forces and the greater the chance for injury.

The compressive force is of the more critical value in the back as this seems to be the source of many of the back problems associated with lifting (Chaffin and Park, 1973; Jayson, 1981; NIOSH, 1981). Combining the muscular force with the component of the weights acting on the base of the spine which are normal to the surface of the disc, will yield the estimate of the compressive force on the spine. The shear force is estimated from the component of the weights supported at the base of the spine which is parallel to the surface of the disc. According to Figure 9 the following equations can be obtained:

$$\text{Compressive force} = F_M + (W_L + W_T + W_A) \cdot \text{Cos} \propto$$

$$\text{Shear force} = (W_L + W_T + W_A) \cdot \text{Sin} \propto$$

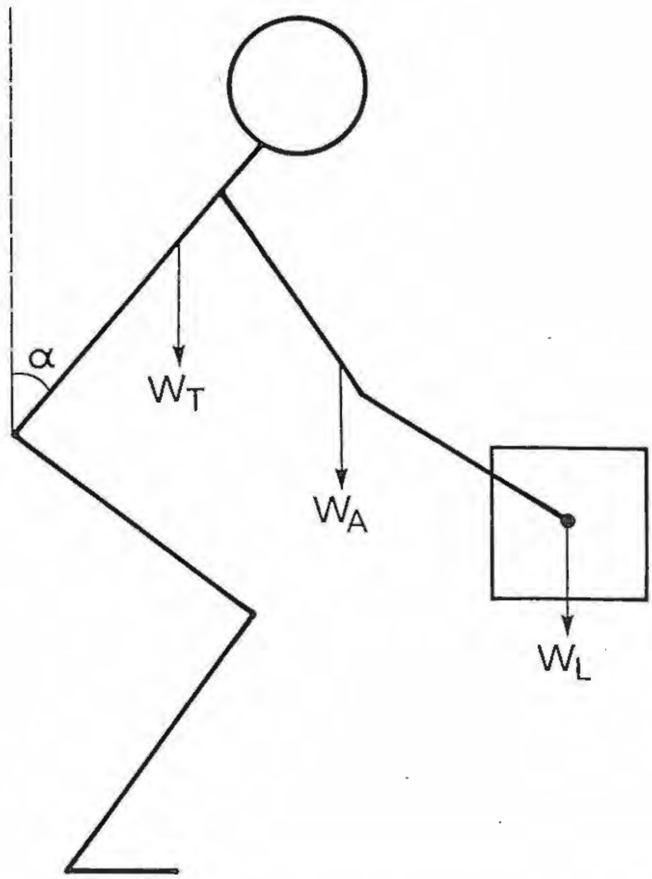

Figure 9 **Forces and angles involved in analysis of the stress on the lower back**

Chaffin and Park (1973) illustrated the observed incidence rates for lower back pain related to predicted back compressive forces on the L_5/S_1 disc as shown in Figure 10. NIOSH (1981) noted that it is apparent that jobs which place more than 650kg compressive load on the lower back are hazardous to all but the healthiest of workers and that a level of 350kg or lower should be viewed as an upper limit for job design.

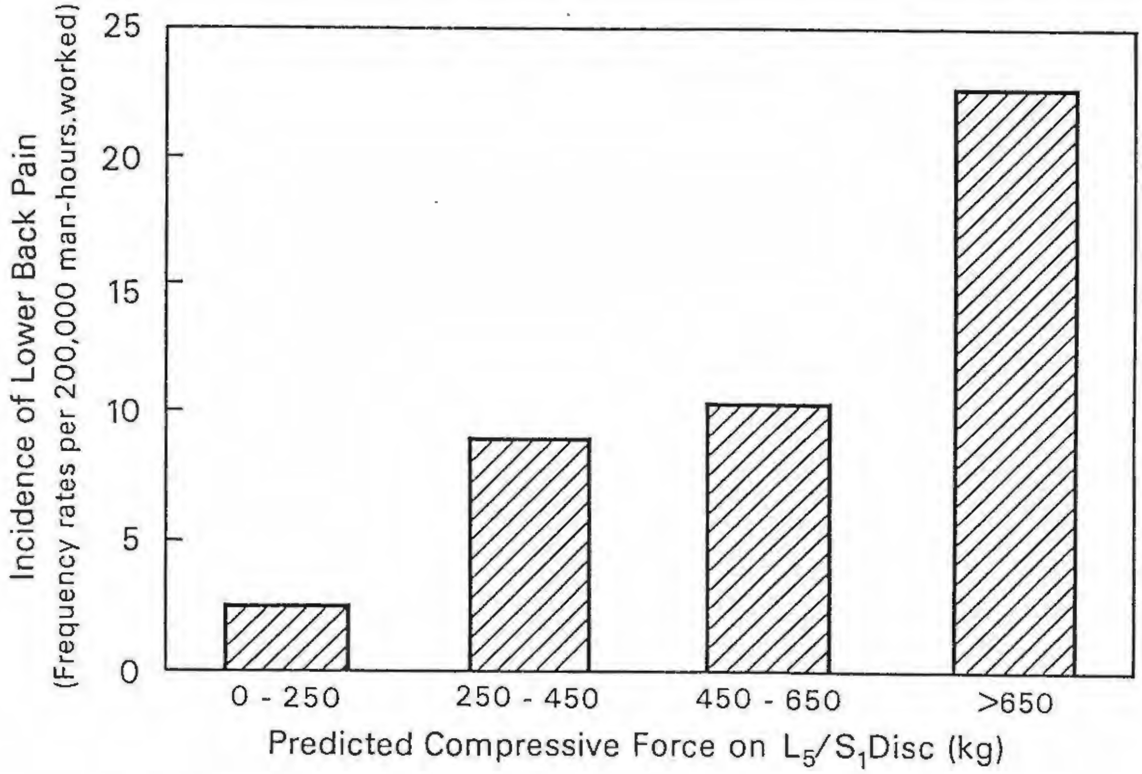

Figure 10 **Relation between lower back pain and compressive force** (Chaffin and Park, 1973)

2.4.2 Biomechanical Modelling of the Lower Back

Modelling of the human body, its segments and tissues, represents one of the methods currently utilized to study specific problems in human mechanics. Modelling is required when a physical test would be destructive or potentially injurious to an individual (Miller, 1979). Sufficient research has been performed to substantiate the view that single factors rarely cause musculoskeletal disorders in industry: it is far more likely to be the result of the interaction of several factors. Modelling technique permits consideration of interacting multiple risk factors either individually or collectively. Through modelling, despite necessarily simplifying assumptions, a manual materials handling task can be designed and evaluated without the need for comprehensive, perhaps risky experimentation, elaborate data collection and analysis. In studying a particular system through modelling, if it appears that the model does not simulate or predict a system's behaviour well enough, it is acceptable and simple to change some input parameters, or part of model, to study the complex nature of the real system. One of the most serious limitations of this method of research however, is the unavoidable trade-off between simplicity and validity when extrapolated to aid in understanding the complexities of the human body.

Almost a century ago, scientists had already started to build the foundation of modern biomechanical modelling (Chaffin, 1969). Since then many investigations have been conducted, resulting in better estimates of (1) the location of the mass centres of gravity, (2) the link lengths, and (3) the magnitudes of the moments of inertia of the various body segments (Chaffin and Andersson, 1984). However, it was not until access to high speed computers became widespread that anthropometric data could be used in developing biomechanical models to study the mechanics of the human body. Chaffin (1969) classified biomechanical models as a basis of two types. One type of model is formulated to determine whole body CG location in various postural configurations. The second model primarily estimates forces and torques at various articulations of the body during voluntary actions.

In 1988 Chaffin described three types of justification for biomechanical models of the lower back. There is the matter of correct interpretation of the complex data now available from diverse examples of sophisticated bio-instrumentation. Secondly, there is the obvious fact that ethically one cannot experimentally induce actual tissue injury in living humans. Yet via construction of biomechanical models of the lower back it is possible to use available in vitro tissue failure data to predict the risk of an injury to a person performing a specified task in industry. The third motivation for biomechanical modelling of the lower back is practicality: it is often not possible to measure some of the effects of manual work in industry. This is particularly true when a new work situation is being developed. Under these circumstances it becomes necessary to simulate the manual activity and predict whether the task can be expected to be safe.

In order to make biomechanical analyses possible, the human body is viewed as a system of links and connecting joints. The torso is often considered as a simple one-link or two-link system. In a biomechanical model each of the links is of the same length, mass and moment of inertia as that of the corresponding human segment. The mass is considered to be concentrated at a single point on the link, the centre of mass (CM).

Planar models

Detailed descriptions of planar biomechanical models for one or multiple linkages of the human body have been presented by Chaffin (1984) and Chaffin and Andersson (1984). Figure 11 shows a static sagittal plane lifting model.

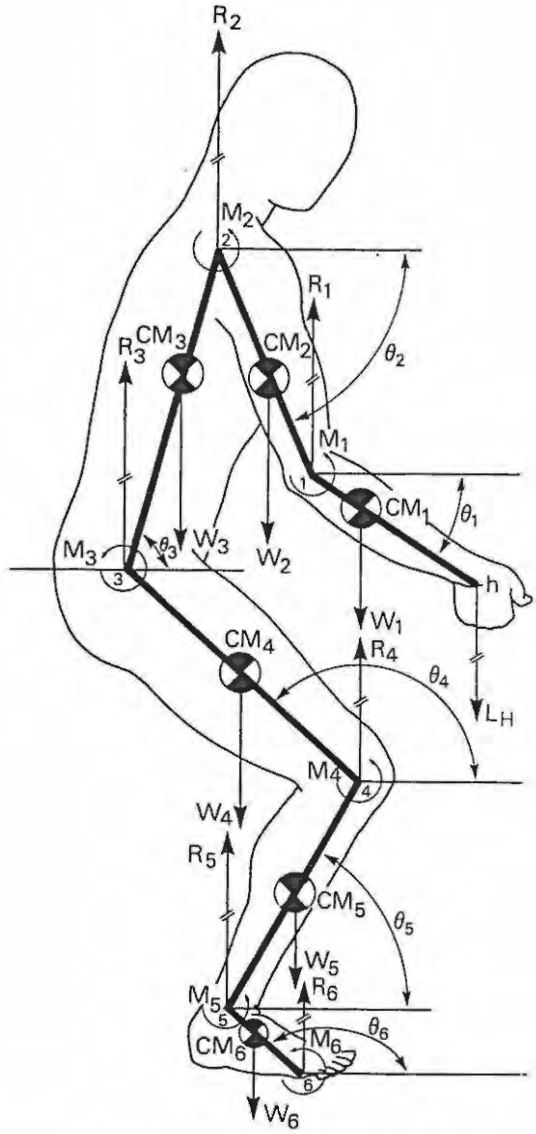

Figure 11 Six-link coplanar model (Chaffin and Andersson, 1984)

Chaffin (1969) introduced a computerized biomechanical model referred to as the Static Sagittal Plane (SSP) model which treats the human body as a series of seven links. Two types of data are needed for the task analysis by the SSP model. First, any external force that may be exerted against the hands is measured and entered into the program as a vector acting at CM of the hands. The second type of information required to describe the activity is the position of the body.

According to input data, the model computes the reactive forces and torques at each articulation during various simulated materials handling tasks.

Leskinen **et al.** (1983) developed a dynamic sagittal plane model for analysis of spinal compression under different lifting techniques. By comparison of static and dynamic biomechanical models, Leskinen (1985) found that the inertial factor increases the peak compressive force on L_5/S_1 by 33–60%, depending on different lifting methods.

Bejjani **et al.** (1984a) used a trigonometric anthropometric model to determine the relationship between knee and back forces during symmetrical sagittal plane lifting. A critical and severely limiting assumption of this model was that the entire upper extremity is held vertical with the forearm in contact with the patella. The model considered loads on the knees as well as on the back during various tasks. Knee and back joint reaction forces, along with their compressive and shear components, were computed as a function of knee, back and ankle angles. These authors reported a high inverse correlation between knee and back forces; as the knee takes an increased amount of load, spinal load is decreased, and **vice versa**. Based on this, a software package entitled "lift stress calculator", was developed to facilitate recommendation of the optimum method of lifting in a given situation for a subject of specific anthropometric, symptomological and physical characteristics of the load (Bejjani **et al.**, 1984b; Parnianpour **et al.**, 1986).

Freivalds **et al.** (1984) developed a biomechanical model for dynamic evaluation of MAL lifting. The basic assumption of this model is that the body is made up of rigid links joined at simple articulations. Seven such links are used: (1) hand-forearm, (2) upper arm; (3) thoracic-lumbar, (4) pelvis, (5) upper leg (sic), (6) lower leg and (7) foot. The trunk division, into two links, is at the L_5/S_1 level to allow the calculation of spinal compressive forces and moments at this disc which is the most often injured during lifting.

Chen and Peacock (1985) developed a six-segment sagittal plane model. The inputs of the model are segment anthropometry, starting joint-angle and load data. First a static equilibrium analysis around the foot support is carried out. The model has an interactive capability, allowing modification of all inputs in order to obtain equilibrium. Secondly, moments, joint-reactions and muscle force are computed as the load is moved to its destination. Finally the program calculates and displays a comparison with an acceptable load based on NIOSH (1981) guidelines for safe lifting practises.

Since it is very difficult to directly measure spinal loads, these are generally calculated from mathematical models of the spine. Traditionally the spine is assumed to be a cantilever, acting in a similar manner to that of a crane. Aspden (1988) has proposed a new model of the spine in which it is considered to function rather as an arch. This new model shows that spinal stresses are not as large as previously calculated using the cantilever model and that posture and curvature of the spine, as well as the magnitude of the weights being lifted, are important considerations.

Potvin **et al.** (1992) developed regression models for predicting L_4/L_5 compressive forces during dynamic lifting in the sagittal plane.

Lab model:

$$\text{PCOMP} = \text{HF}\cdot 6.87 + \text{BM}\cdot 58.80 + \text{MA45}\cdot 78.62 - \text{MAA}\cdot 45.90 - 3406.0$$

Field model:

$$PCOMP = LM \cdot 86.45 + BM \cdot 144.81 - AG \cdot 87.24 - MAA \cdot 32.42 + PA \cdot 13.01$$
$$- TA \cdot 8.77 + LIFT \cdot 296.42 - 578.5$$

Where: PCOMP: peak dynamic L_4/L_5 compression force (N);

HF: peak hand force (N);

BM: subject body mass (kg)

MA45: horizontal distance from load to L_4/L_5 (cm)

MAA: horizontal distance from load to ankle (cm);

LM: load mass (kg)

AG: subject abdominal circumference (cm);

PA: pelvic angle;

TA: trunk angle;

LIFT: type of lift (stoop = 0; squat = 1).

These two models explained at least 90% of the variance in the peak L_4/L_5 intervertebral disc compression force. It was concluded that both models were potentially useful to the practising ergonomist in assessing disc loading since they use relatively easily acquired independent variables.

Three-dimensional models

Two-dimensional coplanar models are very useful in the evaluation of many occupational tasks. In some cases an individual will lift a load from somewhere outside the sagittal plane, or twist while lifting, or will use one arm when lifting, in other words execute an asymmetrical action. In these situations, the external forces acting on the body must be treated in three dimensions, and these forces are considered to be non-coplanar. This results in six independent equilibrium equations in reference to three orthogonal axes at each joint:

$$\Sigma Fx = 0 \qquad \Sigma Fy = 0 \qquad \Sigma Fz = 0$$

$$\Sigma Mx = 0 \qquad \Sigma My = 0 \qquad \Sigma Mz = 0$$

A computerized model for 3-dimensional static strength evaluation of jobs was developed by Garg and Chaffin (1975). This model is based on a mechanical analog of the human body which treats its segments as a set of links. The model allows a user to specify anthropometry, body postures and hand loads. The outputs from the model are the reactive forces and moments at each of the joints of the linkage for the designated input values. The model also allows the user to compare the joint load moments with muscle strength moments to predict the strength exertion of specific populations. Garg **et al.** (1983) used this 3-D model for assessing the biomechanical stress on the lower back and found that pulling a load at an angle towards the body reduces the compressive force on the L_5/S_1 disc by 11% on average.

Schultz and Andersson (1981) have developed a 3-D static lower back model (Figure 12). The internal forces considered in this model are: anterior and right-lateral shear forces (Sa, Sr); right and left erector spinae muscle forces (Er, El); right and left latissimus dorsi forces (Lr, Li); external and internal oblique muscle forces on right and left side (Ir, Xr and Il, Xl); right and left rectus abdominous muscle force (Rr, Rl); disc compressive force (C) and intra-abdominal pressure force (P).

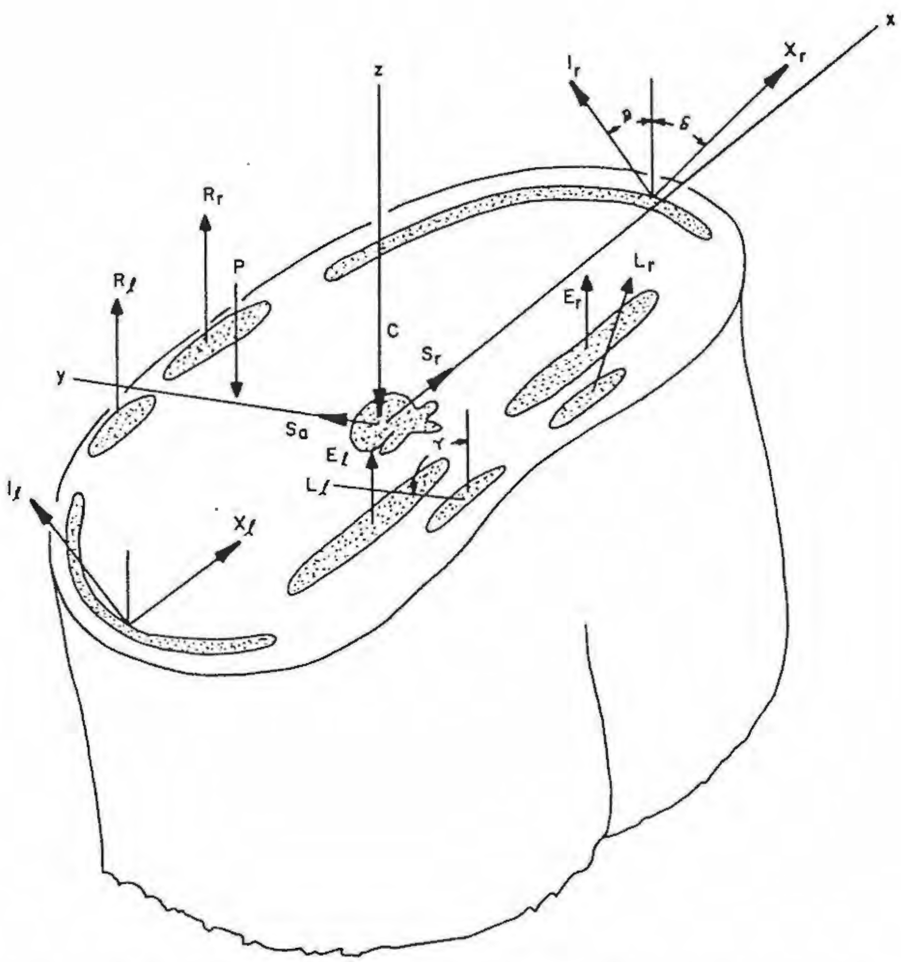

Figure 12 Ten-muscle model of the lower back (Schultz and Andersson, 1981)

Schultz **et al.** (1982) claimed this 3-D lower back model predicts the loads imposed on the lumbar trunk structures during performance of tasks involving trunk bending and twisting moderately well, but not as well as it can predict the loads imposed by sagitally symmetrical tasks that tend only to flex the trunk. It has been suggested that in fact trunk twisting by itself does not seem to load the spine or trunk muscles very much, and that trunk flexion and other activities that impose large flexion moments load these trunk structures heavily, whether or not accompanied by twisting.

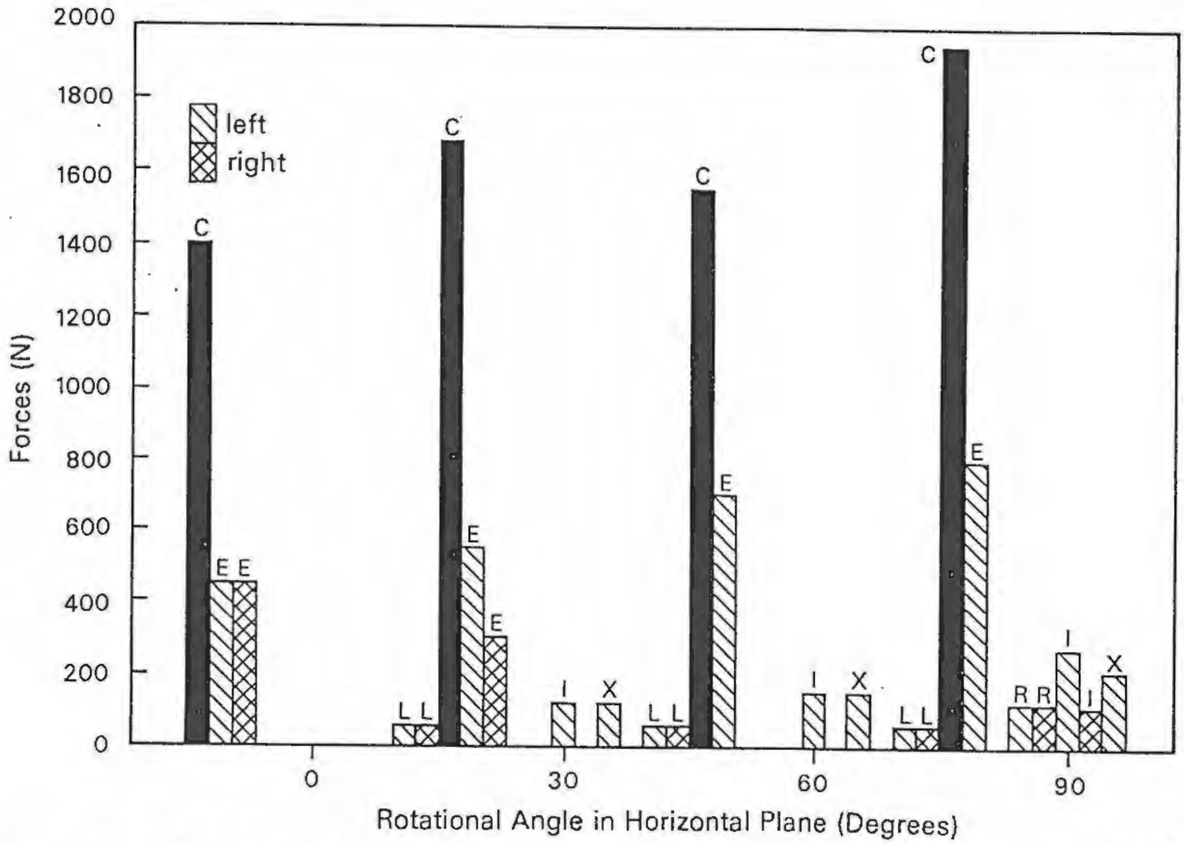

Figure 13 **Predicted forces on rotational angle** (Bean and Chaffin, 1988). (C:compression; E:erector spinae; I:internal obliques; X:external obliques; R:rectus abdominus; L:latissimus dorsi)

Bean and Chaffin (1988) introduced a double linear programming optimization technique for muscle contraction forces calculations which are statically indeterminate in Schultz and Andersson's 3-D lower back model. This technique simultaneously minimizes the torso muscle contraction intensities and motion segment compression forces. The complexity of predicted muscle responses to asymmetric lifting is depicted in Figure 13.

Kromodihardjo and Mital (1986) reported a three-dimensional dynamic biomechanical model which can accurately assess inertial forces and analyse symmetrical and asymmetrical manual lifting activities. This model utilized the

Internal Trunk Model (Schultz and Andersson, 1981) for calculating compressive and shear forces at the L_5/S_1 joint after finishing the analysis of kinematic and kinetic characteristics of manual lifting. The results of the study (Mital and Kromodihardjo, 1986) indicated that even though lower weights are accepted for lifting when lifting loads asymmetrically or in bigger boxes or handling boxes without handles, the spinal stresses generated are significantly higher than when lifting loads symmetrically or in compact boxes or when handling boxes with handles. At the maximum acceptable weight of lift, the compressive forces generated were observed to be at least 30% to 50% lower than the compressive failure limit of the spinal structure.

A simulation model of human trunk motion was presented by Reilly and Marras (1989) in which the equilibrium model of Schultz and Andersson (1981) was embedded. Given electromyographic and intra-abdominal pressure data collected under static or isokinetic dynamic conditions, the model can quantify the time-varying loading of the spine. Tracy (1990) developed a 3-D micro-model of the lower back. In this model the oblique muscles of the trunk are assumed not to be involved in flexion/extension moments and a series of simple rules is adopted to determine how to distribute forces among the trunk muscles. In most cases the difference between the force values predicted by the micro-model as against the optimization model (Bean and Chaffin, 1988), is less than 10%.

Jager and Luttmann (1989) used a dynamic 19-segment human model to assess the lumbar stress of load lifting. Various types of trunk flexion can be analysed due to the provision of 5 small link-segments in lumbar region. It was reported that the ranges of compression and shear forces on L_5/S_1 disc are 0.4 to

10kN and 0.2 to 0.9kN, respectively, depending upon postures and load weights. Thus the influences of lift velocity and jerkiness of movement on lumbar stress were quantified. Compressive forces at the fifth lumbar intervertebral disc were also compared. It was noted that this model is capable of calculating the lumbar stresses during asymmetric activity.

The above constitutes only a brief review of the state of the art of biomechanical modelling. Many of these efforts have been directly applied in industrial situations, for example, Gagnon **et al.** (1986) and Smith **et al.** (1982). To date, biomechanical modelling of occupational tasks has been largely static in orientation and confined to very simple uni-planar motions (Chaffin and Andersson, 1984). The need clearly exists to expand the scope of existing models to include more general manual tasks in industry by using three-dimensional, high-speed recording systems. The new more sophisticated models, aided by computers, promise to be useful in identifying problems assosiated with musculo-skeletal disorders in the workplace.

2.4.3 Dynamic Analysis of Manual Lifting

Almost all manual lifting activities are dynamic in nature. They involve body-segment movement. During the last decade several studies have been reported in which dynamic effects are discussed. There are large discrepancies in the reported difference between dynamic and static analyses of lifting tasks. Dynamic analysis may result in 1.2 to 3 times higher peak values of joint moments and compression force than static analysis. The large range is obviously explained by difference in biomechanical methods and studied tasks.

Wood and Hayes (1974) used a two-dimensional 6-segment model to determine force on the L_4/L_5 disc during a lift. One subject lifted a 5.67kg load from a 450mm bench to a standing position. The movement was performed at a slow speed with an average angular velocity of trunk of $60^0 \cdot s^{-1}$. Displacement data were obtained by means of motion film at the rate of 64fps. Moments of force about the L_4/L_5 vertebral joint were calculated dynamically and statically. An increase of 33% was reported for the moment of forces calculated by the dynamic model.

Smith **et al.** (1982) studied the effect of acceleration on data analysis and the effect of sampling rate with a two-dimensional 5-segment model. Resultant forces and torques were recorded as 11 female subjects lifted and lowered rolls of cotton fibre of 14kg mass in overreach positions. Statistically significant differences were found between the results of the static and dynamic analyses. It was reported that the maximum decrement in the resultant moment at the hip due to a change in the data sampling rate from 102 to 12.75fps was 43%.

Garg **et al.** (1982) compared results from a three-dimensional static model with those from a two-dimensional dynamic model. Six subjects lifted four different tote boxes from the floor to an 810mm high table using free style technique. Movements were recorded with a camera and stroboscopic light flashing at 20Hz. Peak dynamic moments and compressive forces at the L_5/S_1 disc were approximately two to three times greater than those based on the static model.

Leskinen (1985) used both dynamic and static biomechanical models to evaluate the lumbosacral compression when 20 subjects lifted a 15kg box from a

100mm high shelf to knuckle height via four lifting techniques. The movements of the body were recorded with a Selspot movement monitoring system. Acceleration of the load was recorded with two tri-axial accelerators, attached on the front and rear of the box, respectively. Data were sampled at the frequency of 100Hz. The forces exerted on the ground by the feet were recorded using a force platform. The model was a planar biomechanical model with only two body segments: the upper limbs and the trunk above L_5/S_1 level including the neck and head. The results revealed that the mean peak acceleration of the load was 4.9—6.3 $m.s^{-1}$, thus increasing the force at the hands by over 50%. The static peak compression was 3989—4650N and the dynamic 5866—6629N, the increase due to inertial factors being 33—60% depending on lifting technique.

Freivalds **et al.** (1984) used a planar 7-segment dynamic model to study a lifting task. Six male subjects lifted boxes from the floor to a table at waist height. The load in the boxes were selected psychologically by each subject. The average loads selected for four boxes were 271, 289, 321 and 334N, respectively. The lifting motion was recorded by means of stroboscope at 20Hz. Ground reactions were calculated. Average peak values for the different boxes varied from approximately 6000—7000N. They found that the dynamic effects for accelerating the load occurred around 150—200ms; the vertical ground reaction forces peaked at values as much as 40% greater than the static load.

McGill and Norman (1985) investigated the difference between statically an dynamically determined moments of force at the level of the L_4/L_5 intervertebral joint. A two-dimensional 6-segment model was used for determining the joint reaction forces and net moments. Four male subjects lifted a load of 18kg from the rear of a table and brought to rest against the subject's abdomen. The movements were filmed by a Locam 16mm camera at a rate of 80fps. The results showed that the dynamic model resulted in peak L_4/L_5 moments 19% higher on average, with a maximum difference of 52%, than those determined from the static model.

Mital and Kromodihardjo (1986) used a three-dimensional dynamic biomechanical model to analyse compressive and shear forces generated during symmetrical and asymmetrical lifting, lifting boxes with or without handles, and lifting loads in different size boxes. Four subjects performed the lifting tasks from the floor to shelf of 1m height. The weigh of load were determined psychologically and ranged from 244—310N. The lifting activity was filmed by using two Locam cameras at 125fps. The ground reaction forces were recorded by means of a Kistler force plate. The results of the study showed that the peaks of the ground reaction forces were on the average 30 to 50% higher than the force values when the acceleration ceased.

Frigo (1990) studied the dynamic loads on the spine using a three-dimensional model. Two subjects (one male and one female) lifted a box (18kg for the male and 14kg for the female) from floor to a table and lowered the box from the table down to the floor. The maximal compression force on the L_3/L_4 in lifting was approximately 2.4 times for the male and 1.9 times for the female with the load computed statically. During lowering, there were two compressive force peaks on the L_3/L_4. For the male, the first maximum was about 53% higher than the corresponding static load, while the second peak was approximately 66% higher. For the female, the first peak was about 50% higher than static load, and second one was 40% higher. However, the maximum values reached by the woman were considerable lower than the corresponding ones in the male.

From the above-cited studies, a conclusion can be drawn that inertial factors should not be neglected when analysing dynamic tasks. Therefore, the dynamic measures of velocity, acceleration and power should be taken into consideration for a complete analysis of the dynamics of lifting.

Bush-Joseph **et al.** (1988) tried to relate the peak extension moment at the level of the L_5/S_1 to the speed of lifting for three different lifting techniques. The lifting speeds ranged from slow ($0.80-0.90 m.s^{-1}$) to normal ($1.1-1.2 m.s^{-1}$) to fast ($1.7 m.s^{-1}$). It was reported that the peak flexion-extension moment at the L_5/S_1 level increased linearly with increasing lifting speed.

Stevenson **et al.** (1990) conducted a dynamic analysis of a 1.83m maximum strength test on an "incremental lifting machine" (ILM). One hundred and thirty-two military personnel (33 females and 99 males) completed a isoinertial lifting test from a starting height of 0.34m to a target height of 1.83m on the ILM. A force transducer attached to the back of the armature provided continuous velocity and displacement data from which the displacement, velocity, acceleration/force, and power profiles were determined. The mean maximum acceleration/force ($7.3 m.s^{-2}$/ 455.8N for females and $8.3 m.s^{-2}$/901.1N for males) was recorded at mid-thigh location (46% of stature). The maximum velocity ($2.01 m.s^{-1}$ for females and $2.19 m.s^{-1}$ for males) was produced at chest height (77% of stature). The maximum power (654.85W and 1339.95w for females and males, respectively), i.e., the product of force and velocity, was recorded at waist height (64% of stature), mid-way between the sites for the maximum force and velocity measurements.

Lindbeck and Arborelius (1991) studied the inertial effects from a single body segment in dynamic analysis of lifting. Ten male subjects lifted a 12.8kg box using two different techniques and two speeds (slow and fast). They reported that the inertial contributions from individual segments to the dynamic effects of the whole body were relatively small except from the large head-neck-trunk segment. The results from semidynamic analysis agreed well with those from complete dynamic analysis for the ankle, hip, and L_5/S_1 joints.

Gagnon and Smyth (1992) investigated effects of the different velocity-acceleration patterns on joint loading in manual lifting. Five experienced workers lifted two different loads (6.4 and 11.6kg) from an initial height of 150mm to a height of 1850mm. There were large kinematic differences between the accelerated task and the slow task. For the 6.4kg load, the average and the maximum velocities doubled with the accelerated lift, whereas the average and the maximal accelerations increased three-fold. For heavier loads, the average and the maximal velocity were only about 50% higher for the accelerated lift whereas the average and maximal accelerations doubled. The maximal compression force at the L_5/S_1 was higher for the accelerated lift by 28% and 35% for the 6.4 and 11.6kg loads, respectively. The accelerated lift increased the shoulder flexion moment by 69% and 88% for the 6.4 and 11.6kg loads, respectively. The elbow flexion moment was increased by 71% for the 6.4kg load and 80% for the 11.6kg load. The increase in the muscular moments of the lower limbs was smaller than for the upper limbs. For the accelerated lift , the hip extension moments were respectively 24% and 35% larger for the 6.4kg and 11.6kg loads than for the slow lift. The moments of the other joints of the lower limbs were generally not different. It was reported that the maximum values of kinematic and kinetic factors were not affected by the pause during the accelerated lift.

Marras **et al.** (1993) investigated the effect of trunk motion characteristics on risk of injury under industrial conditions. More than 400 repetitive manual lifting jobs in 48 diverse industries were surveyed. These jobs were categorized into two groups: low- and high-risk of occupation-related low back disorders. An exoskeletal Lumbar Motion Monitor was employed for documenting the three-dimensional components of trunk motion in the work environment. The results indicated that motion variables of the trunk discriminated well between high- and low-risk groups. Occupation-related low back disorder risk was associated with a combination of five measures representing both the workplace and trunk motion factors. Load moments and lifting frequency were the workplace factors, while lateral trunk velocity, twisting trunk velocity and sagittal flexion range were the trunk motion factors. A comprehensive risk assessment model was developed for defining the probability of low back disorder.

2.5 EVALUATION OF MANUAL LIFTING

2.5.1 Lifting Capability Estimation

In order to design a job to match the physical capabilities of the majority of the workers and be able to place workers on jobs that do not exceed their physical capacity, there is a need for methods and procedures to estimate the lifting capacity of individuals. The lifting capacity of a person implies that such person can repeatedly lift a certain weight over a long period of time without over-exertion or excessive fatigue. Over the past decade, biomechanical and psychophysical approaches have been used extensively to study lifting and to develop mathematical models for prediction of lifting capacity.

Mital and Ayoub (1980) developed a set of models for predicting lifting capacity of male and female industrial workers by utilizing their individual performance and anthropometric characteristics. These models predict the maximum acceptable weight of lift (MAWL) including individual body-weight and explain between 85% and 90% of the variance. According to Mital and Manivasagan (1983), the density of the material handled, location of the system CG, and task frequency all exert a significant effect, and age and body weight also appear to be important predictors of MAWL. Shoulder strength was suggested to be a limiting factor in lifting tasks. These authors developed a stepwise regression equation which underpredicted the MAWL by 3kg on average.

Karwowski and Ayoub (1984) developed prediction models for the MAWL based on subject characteristics and frequency of lift. The model is given in the form of:

$$\mathrm{MAWL(kg)} = 38.269 - 2.546(\mathrm{FQ}) + 0.049(\mathrm{BW}) \times 0.304(\mathrm{AGE}) + 0.095(\mathrm{SS})$$

Where : BW = body weight (kg)

SS = shoulder strength (kg)

AGE = age of subject in years

FQ = frequency of lifting ($\mathrm{lift.min^{-1}}$)

Aghazadeh and Ayoub (1985) conducted a study for the comparison of models for prediction of weight lifting capacity of individuals incorporating static and dynamic strengths. Both dynamic-strength and static-strength models tended to underpredict the capacity of individuals. Their dynamic model yielded a correlation coefficient between the actual and predicted load of $r=0.86$ and average absolute error of 2.2kg. On the other hand, the corresponding figures for their static model were $r=0.86$ and 4.2kg, respectively. These researchers pointed out that lifting is a complex dynamic activity; several factors, including strength, "fitness", flexibility, reaching ability, motor skill and experience collectively contribute to individual lifting capacity and performance. Apparently a dynamic test can predict a dynamic task better than a static measure. Not only does use of dynamic strength result in a better model with a greater prediction capacity, but dynamic-strength testing can be safer and faster to administer.

Genaidy **et al.**(1988) reviewed the recent literature and found that most psychophysical capacity models do not account for the effects of many important parameters such as task duration, asymmetrical lifting activities, percentage of the working population, and container couplings. It is important that these parameters be considered if capacity models are to apply to realistic jobs. Two MAWL models for males and females were proposed by Genaidy **et al.**(1990). The input to the models are: vertical range of lift, frequency of lift, box dimension in the sagittal plane, population percentage, task duration, presence or absence of handles and sagittal or asymmetrical condition. Compared to NIOSH limits, their two models yield lower load values in the frequency range 1-6 lifts.min^{-1} and higher load values in the frequency range 6—12 lifts.min^{-1}. A procedure for evaluation of lifting tasks was also given.

2.5.2 Lifting Task Analysis

A great deal of effort has been expended to develop a method for assessing lifting task risk. Chaffin and Park (1973) and Chaffin (1974) combined the weight, and horizontal and vertical locations of the object handled, into an index of job-stress referred to as a "lift strength rating (LSR)". The value of this rating ranged from zero, when little or no lifting is involved, to 1.0 where the lifting was such that only a very strong person could perform the job due to excessive weight or awkward posture. It was concluded from a longitudinal study that the lifting of loads in positions which create an LSR equal to or greater than 0.2 be considered potentially hazardous to some people.

Charteris and Scott (1990) announced a computer-aided system designed to identify task-operator mismatch. This system identified four task-related risk factors (mass, frequency, reach and stoop/stretch) and four operator-related factors (age, arm strength, back strength and aerobic capacity).

A Job Severity Index (JSI) has been used for control of lifting injury (Ayoub and Selan, 1983; Liles **et al.**,1984; Liles, 1986). This method is based upon the assumption that a direct link exists between the lifting stress encountered by a worker, or working population, and the number, severity and cost of lifting injuries sustained. The JSI can be expressed as a simple ratio of job demand to worker capacity for a given job condition. Since a job consists of several tasks, the JSI is the time- and frequency-weighted average of the maximum weight required by each task, divided by the capacity associated with lifting ranges required by each task. A detailed JSI calculation procedure was described by Liles (1986).

Ayoub and Selan (1983) and Liles and Deivanaygan (1984) have reported the results of two large field studies involving 385 male and 68 female industrial workers involved in 101 different jobs. For frequency of back injuries, frequency of disabling back injuries and cost of injury, substantial increases occurred at a JSI level equal to or greater than 1.5. For the "severity of disabling back injuries" measure a large increase occurred when JSI levels exceeded 2.25. A lifting guideline was recommended that asserts as a general rule that lifting tasks with a JSI above 2.25 fall into the unacceptable range; tasks with JSI less than 1.5 represent a nominal safety risk to most workers, and tasks having JSI values falling between these criteria limits (1.5—2.25) require administrative control.

Waikar **et al.** (1991) studied five different lifting tasks based on biomechanical and subjective estimations of stresses at the lower back. The results revealed that some of the tasks which were evaluated as less severe from the biomechanical point of view, were considered very severe in terms of subjective estimations of stress on the lower back. The task "reach to overreach" was subjectively rated as most difficult and the "knuckle to elbow" task was the least, approximately 60% less difficult than the "reach to overreach" task. According to the biomechanical stresses on L_5/S_1, the "floor-to-knuckle" task was associated with the highest compressive force and the task "reach to overreach" had the lowest compressive force associated with it. The compressive forces developed were approximately 75% higher for the "floor to knuckle" lifting task than the "reach to overreach" task. It was concluded that subjective preference and subjective criterion must be given strong consideration in designing or evaluating industrial tasks.

Most investigations which have been done in the area of manual materials handling unfortunately have not considered the interaction between biomechanical, physiological and psychophysical approaches for measuring the combined effect of physical stresses on industrial workers. Jung and Freivalds (1991) pointed out that applying only one ergonomic approach is insufficient to reveal the combined effect, and furthermore, may lead to different and conflicting conclusions. They have suggested a methodology utilizing the multiple criteria decision-making process for resolving conflicting biomechanical, physiological and psychophysical criteria typically found in the analysis of manual materials handling jobs. This approach determined the degree of contribution of each stressor to an overall stress level with regard to workplace or task attributes as determined by the unacceptability of this stress by the operator.

2.5.3 NIOSH

In 1981, the National Institute for Occupational Safety and Health (NIOSH) in the United States, published a "Work Practices Guide for Manual Lifting" which subsequently became internationally recognised as the most comprehensive lifting guideline. It was based on the following four principles:

1. epidemiology of musculoskeletal injury;
2. biomechanical concepts;
3. physiological considerations;

4. psychophysical lifting limits.

The NIOSH (1981) guidelines were designed to identify hazardous lifting jobs and provide recommendations to reduce the risk elements associated with lift-related jobs. The principle task variables considered by NIOSH included:

1) Weight of object lifted (L).

2) Horizontal location of the hands at origin of lift measured from the mid-point between ankles (H).

3) Vertical location of the hands at origin of lift measured from floor level (V).

4) Vertical travel distance of hands from origin to destination of object (D).

5) Frequency of lifting average over period of lifting (F).

6) Duration of the period during which lifting takes place (less than one hour or on an eight-hour basis).

Since a large individual variability in risk of injury and lifting performance capability exists in the population, the NIOSH recommendations are based on two levels of hazard. The first level establishes an Action Limit (AL) based on the following assumptions:

1) Musculoskeletal injury incidence and severity rates increase moderately in populations exposed to lifting conditions described by the AL.

2) A 3500N compression force on the L_5/S_1 disc can be tolerated by most young, healthy workers. Such force would be created by conditions described by the AL.

3) Metabolic rates would exceed 2.5kcal.min^{-1} for most individuals working above AL.

4) Over 75% of women and over 99% of men could lift loads described by the AL.

The second level of hazard in the guidelines establishes a Maximal Permissible Limit (MPL). This limit is defined to meet the following criteria:

1) Musculoskeletal injury rates and severity rates have been shown to increase significantly in the population when work is performed about the MPL.

2) Biomechanical compression forces on the L_5/S_1 disc are not tolerated over 6500N in most workers. This would result from conditions above the MPL.

3) Metabolic rates would exceed 5.0kcal.min^{-1} for most individuals working above the MPL.

4) Only about 25% of men and less than 1% of women workers are capable of safely performing work above the MPL.

Based on the AL and MPL, three operational zones are defined. Lifting tasks below the AL represent a nominal risk to workers, tasks above the MPL are considered unacceptable and require engineering controls, while tasks falling between the AL and MPL require administrative or engineering controls.

To allow consideration of the collective effect of the task variable, a prediction equation was defined by the NIOSH guidelines. An ideal load is given and adjusted by factors that can be used to improve job design. To determine the maximum load lifting value for a job at the AL, the equation is:

$$AL(kg) = 40 \cdot \left[\frac{15}{H}\right] \cdot (1 - 0.004 \cdot |V - 75|) \cdot \left[0.7 + \frac{7.5}{D}\right] \cdot \left[1 - \frac{F}{F_{max}}\right]$$

$$MPL(kg) = 3 \cdot (AL)$$

Where: H = horizontal location (15—80cm)

V = vertical location (0—175cm)

D = vertical travel distance [25—(200 – V)cm]

F = average frequency of lift (0.25—15.0 lifts.min^{-1})

F_{max} = maximum frequency which can be sustained.

The application of these guidelines, hereinafter in dictated as "NIOSH'81" is limited to two-handed, symmetrical, and smooth lifting directly in front of the body (no twisting or turning) using handles. The data on which the NIOSH guide was based come primarily from studies undertaken in the U.S.A. The calculations required to determine specific NIOSH limits require no information relating to the workforce. Therefore, the predicted risk may be over- or underestimated when applying the limits to some other countries. Evans (1990) compared isometric strength of Cantonese males with NIOSH'81. The results showed the mean Maximal Voluntary Isometric Strength (MVIS) lay somewhere between the MPL and the AL values proposed in the Guide when the loads were at 200mm and 400mm from the lifter's ankles, but was significantly below the AL with loads at 600mm. Further, at 800mm most the subjects were unable to apply any measurable positive vertical force. Evans suggested that required inputs of some workforce information (at least, in terms of stature and body weight) would make the guideline more useful.

The load limit prediction equation defined by the NIOSH'81 guideline was revised and expanded in 1991 (Waters **et al.**, 1993). The revised lifting equation still chooses the compressive force at the L_5/S_1 interface as the critical stress vector and retains the original biomechanical criterion of 3.4kN compression. The 1991 lifting equation provides methods for evaluating asymmetrical lifting tasks, for objects with less than optimal hand-container couplings, and offers new procedures for evaluating a larger range of work durations and lifting frequencies than the 1981 equation.

The ideal mass of 40kg in the earlier equation has been reduced to 23kg in the 1991 equation. The 1991 equation continues to use a vertical height of 750mm for the standard reference location. However, the horizontal displacement factor was increased from 150mm to 250mm for the 1991 equation. Waters **et al.** (1993) concluded that the revised 1991 equation is applicable to a wide variety of lifting jobs and therefore, more likely to protect more workers than the 1981 equation.

The revised lifting equation, which includes six multipliers, is presented in the following format:

$$RWL = LC \times HM \times VM \times DM \times AM \times FM \times CM$$

Components:

RWL: recommended weight limit;

LC: load constant = 23kg;

HM: horizontal multiplier = (25/H);

VM: vertical multiplier = $[1-(0.003|V-75|)]$;

DM: distance multiplier = $[0.82 + (45/D)]$;

AM: asymmetric multiplier = $[1 - (0.0032 \cdot A)]$;

FM: frequency multiplier, depending on the vertical height of the lift and lift frequency;

CM: coupling multiplier, depending on the vertical height of the lift and the quality of the couplings.

where:

H = horizontal distance(cm) of hands from midpoint between the ankles, measured at the original and the destination of the lift.

V = vertical distance(cm) of the hands from the floor, measured at the origin and destination of the lift.

D = vertical travel distance(cm) between the origin and the destination of the lift.

A = angle of asymmetry — angular displacement of the load from the sagittal plane, measured at the origin and destination of the lift.

F = average frequency rate of lifting measured in lifts.min^{-1}.

2.6 ULTRASONIC TECHNOLOGY FOR HUMAN MOVEMENT MEASUREMENTS

Ultrasound has been used to measure distance on a variety of products and applications, such as automatically focusing cameras and sonar (Hsiao and Keyserling, 1990). In these applications distance is measured in only one dimension.

Fleischer and Lange (1983) used an ultrasonic method with a single piezocrystal and three microphone receivers for the analysis of hand-movements in space. The spatial co-ordinates of a moving point are determined by its distances from three fixed reference points, which are obtained by measuring the transmission time of an ultrasonic pulse from the small moving piezocrystal source to three microphones arranged above the workstation. In order to record the angle between the hand and forearm, three ultrasonic sources were used later by Fleischer and Becker (1986). The application of these systems was limited to a single joint. Also the transmitters needed to face in specific directions, which limited their ability to measure certain types of motion.

A three-dimensional ultrasonic system for posture measurement was developed by Hsaio and Keyserling (1990). In order to overcome shadow problems, the ultrasonic system uses more than three receivers to detect signals to reduce the likelihood that a transmitter is obscured from the necessary receivers. The system can be connected to up to eight receivers. Any combination of three unobscured receivers can be used to generate a set of co-ordinates for a transmitter position. The system also allows the user to relocate receiver positions to eliminate the shadowing problem. The ultrasonic system is able to utilize 14 transmitters at various joints. The system sequentially processes the ultrasonic signals in 0.03s per transmitter-receiver pair. The sampling time depends upon the number of transmitters and receivers used, and range from 0.03s (only use one transmitter and one receiver) to 3.0s (use 14 transmitters and 8 receivers). Three experiments: sensitivity and calibration, accuracy/precision and human postural behaviour, were performed to evaluate the measurement system. The results showed that the three dimensional ultrasonic measurement system can be used for accurate measurements of static posture in a laboratory setting. The system still had a few shadowing problems. It was said that further development of the system using parallel processing and 20 transmitters is underway.

This literature review has identified and acknowledged the complexity of the problem in the area of manual materials handling. It serves as a primary basis for the empirical study and modelling, which follows. However, it should be noted that a number of studies have recently attempted to predict trunk muscle forces using surface eletromyography (McGill and Hoodless, 1990; Marras and Sommerich, 1991; Granata and Marras, 1993; Mirka and Marras, 1993; Cholewick and McGill, 1994 and Hughes **et al.**, 1994). These EMG-based models account for the effects of coactivation of the trunk muscles which influence the loading of the lumbar spine. EMG studies have shown that significant coactivity among trunk muscles occurs during dynamic motion and torsion (Marras **et al.**, 1984; Pope **et al.**, 1986 and McGill, 1991). While the existence of a qualitative relationship between the EMG signal and the corresponding muscle forces is well known, the quantitative nature of this relationship is hotly debated within the scientific community (Nigg and Herzog, 1995). The problems associated with quantifying EMG signals together with the invasive methods of using EMG, deem the use of EMG highly impractical in field testing. The objective in developing the present dynamic 3-D model was to acknowledge the complexity of the coactivation of the trunk musculature, but to put focus on the kinematic and kinetic characteristics of asymmetrical lifting.

CHAPTER 3 : PHASE I: THE EMPIRICAL STUDY

3.1 METHODS AND PROCEDURES

3.1.1 Introduction

Most ergonomic information is based on observation and experimentation. According to McCormick and Sanders (1982), ergonomic research can be classified into three types: descriptive studies, experimental research, and evaluative research. No matter which category of research is undertaken, however, there are several fundamental decisions which must be addressed in order to plan and execute the work properly. These include picking a research setting, selecting variables, choosing a sample of subjects, and deciding how the data will be collected.

The choice of research setting involves complex trade-offs. Research carried out in the field usually has the advantage of realism in terms of relevant task variables, environmental constraints, and subject characteristics including motivation. The results obtained from such studies can be generalized to the real world operational environment. The disadvantages, however, include cost, safety hazards for subjects, and lack of experimental control. In field studies there is often no opportunity to replicate the experiment, and many variables cannot be held constant. Research undertaken in the laboratory has the principal advantage of experimental control and the experiment can be replicated as desired. Data collection can be made more precise. For this, however, the researcher may sacrifice some realism and generalizability.

Manual materials handling (MMH) constitutes a complex system which consists of three main components: the worker, the task and the environment. Each of these components has several elements which interact with each other in a complex manner. Field studies can provide a good overall view of the interactions among the elements within MMH systems. However, it is very difficult to examine the role of any single factor separately. In the laboratory, extraneous variables can be controlled according to the experimental requirements. Therefore, researchers are able to test relevant variables as often as possible until satisfactory results are achieved.

A great deal of research in the area of MMH has been carried out in the laboratory setting. Variables such as object weight, frequency of lifting, vertical lift height and horizontal reach have been thoroughly investigated under specified conditions and widely used as risk control factors in industrial lifting activities. However, the majority of these factors have been tested only in symmetrical lifting situations. As mentioned in the previous chapter, asymmetrical lifting activities are prevalent in industrial situations and in daily life. Further investigations are needed in order to understand the interactions between these well studied variables in conjunction with asymmetric lifting factors.

Although the field study method may seem ideal for investigation of natural, largely asymmetrical lifting tasks, the inherent lack of experimental controls in the field constitutes an insurmountable obstacle for the researcher who wishes to investigate the important role of asymmetrical factors in lifting activities. In addition, precise measurements are needed for kinematic and kinetic analyses of asymmetrical lifting. Unavoidably, therefore, in the present study, the kinematic and kinetic properties of asymmetrical manual lifting were investigated under controlled laboratory conditions.

3.1.2 Asymmetrical Lifting Task

In industrial working situations there are several kinds of asymmetrical lifting and the extent of asymmetry will vary considerably. In the present investigation, asymmetrical lifting was studied in the sagittal plane (0^0), and in planes at 30^0, 60^0 and 90^0 to the right. In successive trials a box was placed in position perpendicular to one of those planes. The subject was instructed to keep his feet in the sagittal plane and required to lift the box, using both hands, from two initial vertical grip heights of 150mm (the box handle height) and 500mm (average knee height of subjects) to a level of 800mm without moving the feet. The subject then took one step forward as needed, and finally placed the box on an 800mm high bench in the starting sagittal plane. Each subject lifted the box using whatever working posture was found to be most comfortable. The horizontal distance between the mid-point of the link between ankles and the centre of the box was controlled at 350mm. The dimensions of the box being lifted in the study were 430x300x270mm. A total of eight lifting tasks (2 vertical heights x 4 planar angles) was investigated. The eight lifting tasks were randomly assigned to the subject during data collection. In order to minimize physiological fatigue, each subject was required to perform all tasks infrequently, i.e. less than once per five minutes. The total work duration was 45 min. In this study, the mass moved was fixed at 10kg for all eight lifting tasks.

The weight of load lifted (98.1N) was determined by applying the "Work Practice Guide for Manual Lift" (NIOSH, 1981). These "NIOSH'81" guidelines are based on four criteria: (i) epidemiological; (ii) biomechanical; (iii) physiological; (iv) psychophysical. These data were utilized to define two lifting criteria known as the Action Limit (AL) and the Maximal Permissible Limit (MPL).

The AL is calculated by the following equation:

$$AL(kg) = 40 \cdot \left(\frac{15}{H}\right) \cdot (1 - 0.004 \cdot |V - 75|) \cdot \left(0.7 + \frac{7.5}{D}\right) \cdot \left(1 - \frac{F}{F_{max}}\right)$$

where: H = the horizontal location of the hands forward of the mid-point between the ankles at the origin of lift (cm);

V = the vertical location of the hands at the origin of lift (cm);

D = the vertical travel distance from the origin to destination of lift (cm);

F = the frequency of lift in lift.min^{-1};

F_{max} = the maximum frequency that may be sustained over the specified duration of task performance.

The MPL is three times the AL.

According to NIOSH, the AL represents the lift capabilities of 99% of males and 75% of females, while the MPL reflects the capability of only 25% of males and fewer than 1% of females.

In the present study, the task variables and the AL's for the two selected initial vertical heights of lift were as follows:

—150mm initial grip height lift:

H=350mm; V=150mm; D=800mm; F=0; AL=10.3kg.

—500mm initial grip height lift:

H=350mm; V=500mm; D=450mm; F=0; AL=13.4kg.

The application of NIOSH'81 guidelines is limited to two handed, symmetrical, and smooth lifting directly in front of the body (no twisting or turning) using handles. As described in Chapters 1 and 2, asymmetrical lifting is more hazardous to the musculoskeletal system than symmetrical lifting. The drop-off in maximum acceptable weight and static strength (from the sagittal plane values) increases with an increase in the angle of asymmetry.

Because NIOSH'81 was only applicable to sagittal lifting tasks, the 1981 lifting equation was revised in 1991, and first made public two years later (Waters **et al.**, 1993). The revised equation reflects new findings and particularly considers additional asymmetric and coupling risk factors. Since the present study was conducted before publication of the revised lifting equation in 1993, the NIOSH'81 guidelines were observed throughout the empirical phase, although the 1993 revisions were considered in the later modelling phase of the present study.

Table III shows AL as predicted by the 1981 equation versus the weight limits recommended in the 1991 revision, for the lifting tasks investigated in the present study. It is clear that the weights recommended by the 1991 equation decrease gradually as the asymmetrical angle increases from 0^0 to 90^0. The average rate of decrease with other values held constant is 10.7% per every 30^0.

By contrast the AL predicted by the 1981 equation remains constant, twisting not being a factor incorporated into the original lifting sequence.

Table III Comparison between acceptable loads via NIOSH'81 and NIOSH'91 (starting height as used in the present study)

Lifting Task		NIOSH'81	NIOSH'91
Height	Asymmetric Angle	(kg)	(kg)
150mm	0^0	10.3	11.8
	30^0	10.3	10.7
	60^0	10.3	9.5
	90^0	10.3	8.4
500mm	0^0	13.4	14.7
	30^0	13.4	13.3
	60^0	13.4	11.9
	90^0	13.4	10.5

3.1.3 Equipment

The V-scope ultrasonic motion monitor

The V-scope ultrasonic motion monitor, which provides full three-dimensional multibody tracking capacity at a high sampling rate, was originally designed for laboratory measurement, recording and demonstration of physical motion. The device consists of three major components: a microcomputer, three transmitter-receiver "towers" (180x80x40mm) and four colour-coded transmitter-receiver "buttons" (25x25x15mm) see Figure 14.

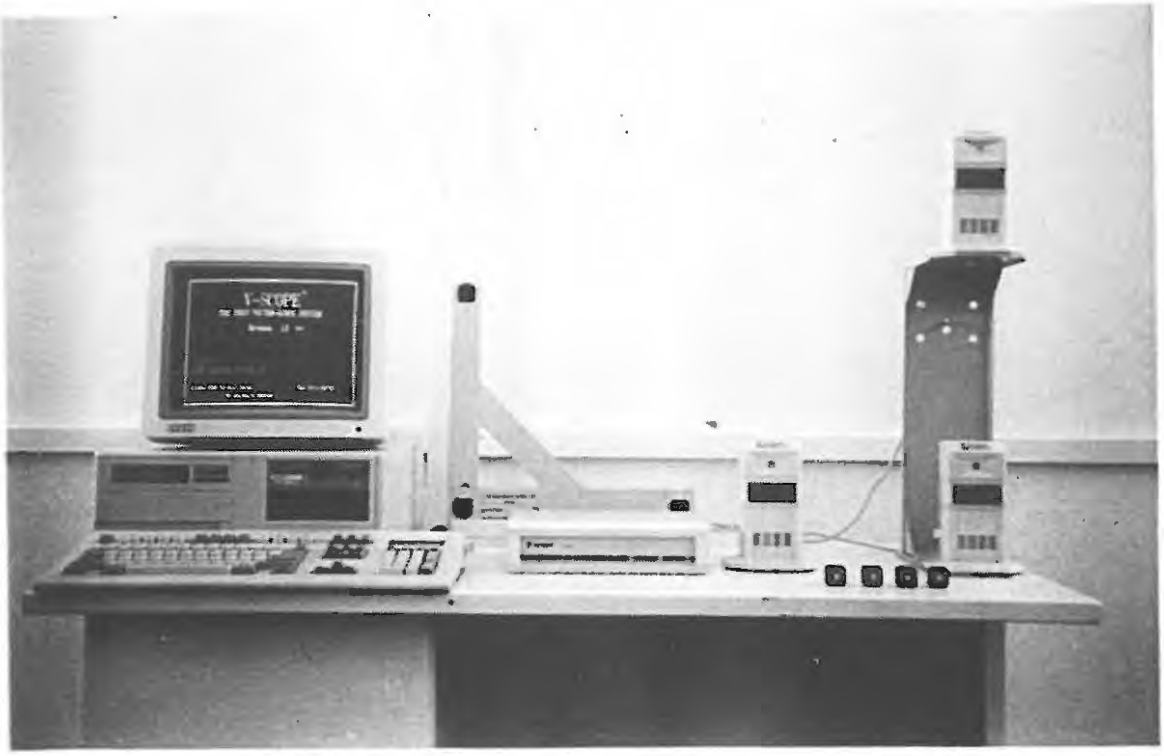

Figure 14 The V-scope system

The V-scope microcomputer is the "brain" of the system. It controls the operation of the towers, generates and processes signals, determines tower-button distances and calculates three-and two-dimensional positioning and interfaces with the personal computer.

The three towers are equipped with an ultrasonic transducer, an infrared transmitter and a thermistor which measures the ambient temperature for calculation of the velocity of sound in air. The minimal number of properly-positioned towers necessary to determine the spatial co-ordinates of the buttons is equal to the number of dimensions of the actual experiment in space, e.g. either two or three towers can be used for two-dimensional experiments. The three towers are designated tower A, tower B and tower C, according to the respective port at the rear of the V-scope microcomputer to which they are connected by cable.

The button is a very small, light (12.9g) and complicated device, incorporating an infrared receiver and synchronized ultrasonic transmitter. It is easily attached to an object, animate or inanimate, for subsequent motion analysis. The centre of the button's transducer is the exact point whose position is measured by the V-scope. The buttons are energized by an internal battery. To conserve battery power, the button is normally inactive. A strong infrared signal is necessary to activate the button's power. The button remains on as long as it communicates with the towers and shuts off automatically after about a second from the last communication.

The operation of the V-scope is manipulated by initiating the V-scope software on a personal computer. After the V-scope microcomputer activates the towers by sending a burst of electrical pulses to them, the towers transmit a narrow pulsed infrared triggering signal. The button's body is composed of plastic material which is transparent to infrared, thus allowing the infrared trigger from the towers to reach the infrared sensor inside the button. The triggering pulse is transformed by the electronics of the button into a synchronized ultrasonic wave transmitted by the button's ultrasonic transducer. Each tower in turn receives the ultrasonic signal from the button at a time determined by its distance from the button. The received ultrasonic signal is transformed by the same frequency and phase. This signal is amplified and fed into the V-scope microcomputer. The microcomputer compares the incoming signal to the original outgoing signal to accurately determine the time elapsed between transmission and reception. This time interval, multiplied by the velocity of sound in air, provides the tower-button distance. This distance measurement process is performed simultaneously for all active towers. Knowing the exact relative position of the different towers, the V-scope transforms the tower-button distances into the spatial co-ordinates of the button by the mathematical process of triangulation. This position process is repeated many times per second at predetermined intervals, thus providing the desired continuous tracking. The sampling rate of the V-scope can be varied from 10 to 100 ms.

The V-scope system can be used for recording the motions in one-, two- or three-dimensional space. The following five tower configurations can be selected to best suit any experiment:

1) 3-D horizontal configuration;

2) 3-D vertical configuration;

3) 2-D horizontal configuration;

4) 2-D vertical configuration;

5) 1-D configuration.

During an experiment, tower-button communication between all active towers and buttons has to be maintained continuously. There must be a direct line of sight at each moment between at least one tower and each button. As all towers transmit the same triggering signal, one tower is sufficient for the infrared communication. A sonic path must be maintained continuously between each active button and all active towers. In most cases small objects in the line of sight will not interfere with the sonic path. The angle of sight of both button and tower is 160°. The tower-button communication range is 0.1m—5m. The maximum range is reduced when the gain is set at low level and a very high sampling rate is chosen.

A 3-D vertical tower configuration was chosen for the present study. In a 3-D vertical configuration where three towers are active, the three distances $\{d_A, d_B, d_C\}$ from a button to towers A, B and C respectively, are measured simultaneously by the V-scope (Figure 15).

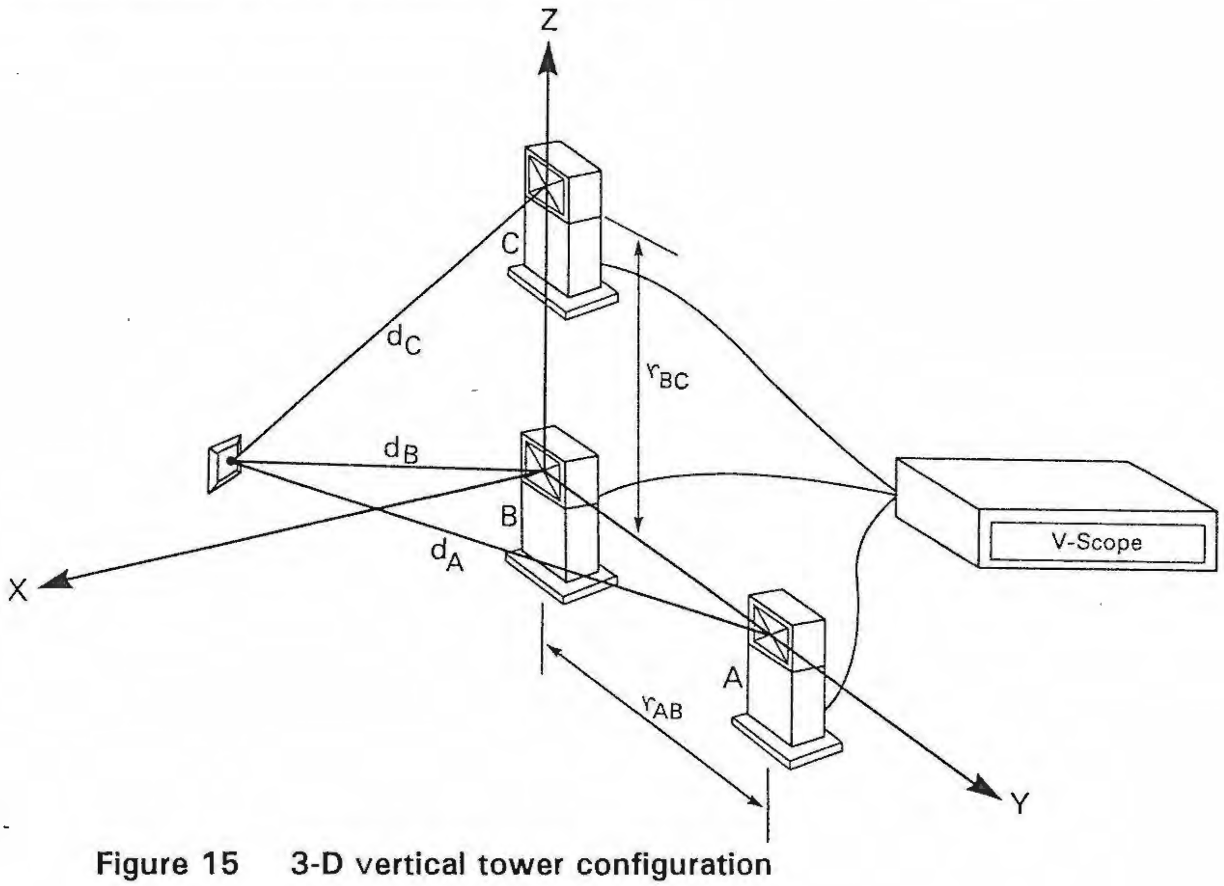

Figure 15 3-D vertical tower configuration

Using trigonometric formulae, the V-scope microcomputer performs a triangulation calculation, which may be expressed symbolically as:

{the co-ordinates of Tower A,B,C} $\oplus \{d_A, d_B, d_C\} \rightarrow$ **[X,Y,Z] of the Button**

The inter-tower distances A-B and B-C are programmed into the computer, so it is essential that they be measured precisely. These distances may be measured either manually or automatically by using a manufacturer-supplied metal set-up template. The lines connecting towers A-C and B-C must be exactly perpendicular. Inaccuracy in the measurement of the 90^0 angle between the lines connecting towers A-B and B-C results in inaccurate data when the experiment is run. The exact positions of the towers serve as reference points for the calculation of the cartesian co-ordinates of the buttons. Thus, an error in the measurement of inter-tower distances or orthogonality affects the accuracy of the button's recorded location.

The V-scope is programmed with an automatic tower set-up procedure, which facilitates tower positioning and verifies tower orthogonality. When the automatic tower procedure is used, tower and button exchange roles: the tower's position is measured using a known button configuration and known inter-button distance. Three buttons placed at predetermined locations on the supplied V-scope metal set-up template define the three-dimensional space in which each tower's co-ordinates are measured. From these measurements, inter-tower distances and angles are calculated and displayed on the PC screen in real time, enabling the user to position the tower accurately.

The V-scope tracks the motion of one or more bodies in space. The time-function of the position vectors $\mathbf{R}(t) = [\mathbf{X}(t), \mathbf{Y}(t), \mathbf{Z}(t)]$ is continuously measured and recorded by the system for each movable body. All other physical parameters, such as velocity and acceleration, are obtained from $\mathbf{R}(t)$ by mathematical processes executed in real time.

The velocity vector **V**(t) is calculated as:

$$\boldsymbol{V}(t) = \frac{\boldsymbol{R}(t)_{n+1} - \boldsymbol{R}(t)_n}{t_{n+1} - t_n}$$

The acceleration vector **A**(t) is obtained as:

$$\boldsymbol{A}(t) = \frac{\boldsymbol{V}(t)_{n+1} - \boldsymbol{V}(t)_n}{t_{n+1} - t_n}$$

For 3-D vertical tower configuration as shown in Figure 15, the following formulae have been derived for [X, Y, Z] co-ordinate calculation (Litek, 1990).

$$X = \sqrt{\left(d_B^2 - Y^2 - Z^2\right)}$$

$$Y = \frac{d_B^2 + r_{AB}^2 - d_A^2}{2 \cdot r_{AB}}$$

$$Z = \frac{d_B^2 + r_{BC}^2 - d_C^2}{2 \cdot r_{BC}}$$

Y and Z co-ordinates are calculated from measured data, while the X co-ordinate is derived from the calculated values of Y and Z. The accuracy of X is always lower than that of Y and Z, since it includes the summed imprecisions incorporated in the calculation of the other two co-ordinates.

The accuracy of the calculated co-ordinate is a non-linear function of its components: i.e., the accuracy in the measurement of the tower-button distance and of the inter-tower distance. The former depends on background noise and V-scope resolution capability, while accuracy in the determination of inter-tower distance and tower orthogonality depends on the user's measurement, or on V-scope resolution capability when using the automatic tower setup. The expected error in the calculation of the co-ordinates, caused by the inaccurate measurement of independent variables, is given by the formulae below (Litek, 1990):

$$\delta X = \pm \sqrt{\left(\frac{d_B}{X}\right)^2 \cdot \delta d_B^2 + \left(\frac{Y}{X}\right)^2 \cdot \delta Y^2 + \left(\frac{Z}{X}\right)^2 \cdot \delta Z^2}$$

$$\delta Y = \pm \sqrt{\left(\frac{d_B}{r_{AB}}\right)^2 \cdot \delta d_B^2 + \left(\frac{d_A}{r_{AB}}\right)^2 \cdot \delta d_A^2 + \left(\frac{(Y - r_{AB})}{r_{AB}}\right)^2 \cdot \delta r_{AB}^2}$$

$$\delta Z = \pm \sqrt{\left(\frac{d_B}{r_{BC}}\right)^2 \cdot \delta d_B^2 + \left(\frac{d_C}{r_{BC}}\right)^2 \cdot \delta d_C^2 + \left(\frac{(Z - r_{BC})}{r_{BC}}\right)^2 \cdot \delta r_{BC}^2}$$

In a 3-D vertical tower configuration (Figure 15), for example, assume the inter-tower distance r_{BC} is 0.5m and r_{AB} is 1m, measured with an accuracy of 1mm ($\delta r_{BC} = \delta r_{AB} = 1$mm), and the button-tower distance accuracy is 0.5mm ($\delta d_A = \delta d_B = \delta d_C = 0.5$mm). The button is placed such that X=Y=Z=1m, thus $d_A = 1.414$m, $d_B = 1.732$m and $d_C = 1.5$m. In this case,

$$\begin{aligned}\delta X &= \pm \sqrt{\left(\frac{1.732}{1.0}\right)^2 \cdot 0.5^2 + \left(\frac{1.0}{1.0}\right)^2 \cdot 0.9^2 + \left(\frac{1.0}{1.0}\right)^2 \cdot 2.5^2} \\ &= \pm\ 2.79mm\end{aligned}$$

$$\delta Y = \pm\sqrt{\left(\frac{1.732}{1.0}\right)^2 \cdot 0.5^2 + \left(\frac{0.5}{1.0}\right)^2 \cdot 0.5^2 + \left(\frac{1.0 - 1.0}{1.0}\right)^2 \cdot 1.0^2}$$
$$= \pm\ 0.90mm$$

$$\delta Z = \pm\sqrt{\left(\frac{1.732}{0.5}\right)^2 \cdot 0.5^2 + \left(\frac{1.5}{0.5}\right)^2 \cdot 0.5^2 + \left(\frac{1.0 - 0.5}{0.5}\right)^2 \cdot 1.0^2}$$
$$= \pm\ 2.50mm$$

For the non-orthogonal setup the X co-ordinate is given by:

$$X = \sqrt{d_B^2 - (Y^2 + Z^2 - 2 \cdot Y \cdot Z \cdot Cos\alpha)}$$

Where α is the angle between the Y and Z axes. When α differs from 90^0 by $\delta\alpha$, the additional error in X due to non-orthogonality is given by:

$$(\delta X)_\alpha = \pm\sqrt{\left(\frac{Y \cdot Z}{X}\right)^2 \cdot Sin^2\alpha \cdot (\delta\alpha)^2}$$

If α is accurate up to 1^0 (0.00176 rad) then at X=Y=Z=1000mm.

$$(\delta X)_\alpha = \pm\sqrt{\left(\frac{1000 \cdot 1000}{1000}\right)^2 \cdot Sin^2\left(\frac{\pi}{2}\right) \cdot (0.00176)^2}$$
$$= \pm\ 17.6mm$$

Comparing the errors introduced by inaccurate measurement of inter-tower distances and button-tower distances, the non-orthogonality of the tower set-up causes a much greater error on the X co-ordinate. But this large error can be easily reduced to several millimetres by careful placement of the towers and the buttons.

For example, if the button is placed at $X = 1500$mm and $Y = Z = 500$mm, the error in α is still 1^{0}, but the error $(\delta X)_{\alpha}$ will be reduced to 2.9mm.

A comprehensive assessment of the V-scope was conducted by Daw (1991). A turntable and the right triangle template used in the auto-setup for the V-scope towers, were used for calibrating the performance of the V-scope. The experiment showed that the V-scope yielded very good results with the positioning error within 0.5mm. Daw further tested the movement accuracy of the V-scope using a free fall, a pendulum and a gyroscope device. He concluded that the V-scope is easy to use and gives accurate movement measurement. Charteris **et al.** (1994) followed Daw's calibrating procedures and retested the validity and repeatability of the V-scope using a free-fall test and template test. The average error for positioning was 0.85mm. For movement measurement, the difference between the V-scope values and expected values was within $\pm 0.8\%$. The applications to which Charteris and his colleagues put the V-scope, originally designed for experimentation in the physics laboratory, represent the first reported use of this device in human motion analysis. A dynamic analysis of manual lifting and shoulder extension exercise was conducted separately. Li **et al.** (1993) and Li (1993) successfully applied the V-scope in 3-D dynamic analysis of asymmetrical lifting.

Lumbar motion monitor

The Lumbar Motion Monitor (LMM) is modelled after the trunk motion control system of the human back. The spinous processes and transverse processes form a "T" section about the neural arch of each spinal vertebra. This T-section anchors a 3-dimensional tension element that mediates motions of the spine in various phases. These sections are connected to each other via ascending and descending processes by ligaments and muscles that envelop the facet joints of the spinal segments. The LMM is an exoskeleton of the spine that replicates the motion of these T sections in the lumbar region. Detailed description of the LMM may be found in Marras et al. (1992; 1993).

The Lumbar Motion Monitor industrial assessment package comprises two exoskeletal instruments (large and small), a calibration unit, a laptop computer, an umbilical cable and three harnesses (large, medium and small) (Figure 16).

Figure 16 The LMM industrial assessment package

The Lumbar Motion Monitor was designed to identify, monitor and document the three-dimensional components of motion experienced by the dynamic human spine during MMH. The LMM is attached via the harness system to the thorax and the pelvis, acting as stable anchors between which predominantly lumbar spinal motion may be measured relative to the pelvis as a base. The unit is positioned on the subject's back in such a way as to create a parallel motion in the mechanical exoskeletal T-sections, thus duplicating the movement of the subject's spine.

There are 20 and 26 mechanical T-sections in the small and large exoskeletal units respectively. The ends of each edge of the exoskeletal T-section are connected via wires to potentiometers in the base of the LMM. These three wires differentially change the voltage readings in the potentiometers as the exoskeleton moves forwards, backwards, or to the sides. A cable is also placed through the junction in each T-section and is connected to a fourth potentiometer. This potentiometer changes as the exoskeleton is twisted.

LMM signals are sampled at 60Hz via an analog-to-digital converter. Voltage readings from the potentiometers are converted into angular position using a regression model. Angular velocity and acceleration are obtained through numerical differentiation. Accuracy of measures of velocity and acceleration are dependent upon the ability of the system to validly measure position. Four channels are required for the three axes of movement. The sagittal and rotational motions use one channel each, while the lateral movements use the sum of two channels to develop the resultant movement vector.

The mechanical ranges of motion of the LMM in operation are as follows, relative to anatomical standing:

Sagittal: −35 to +65 degrees;

Lateral: −45 to +45 degrees;

Rotational: −45 to +45 degrees.

The resolution of the LMM is 0.2343 degrees.

Marras et al. (1992) assessed the accuracy and repeatability of the Lumbar Motion Monitor to measure the instantaneous changes in trunk position, velocity and acceleration in three-dimensional space. A three-dimensional reference frame was developed to calibrate the LMM with respect to the device's position in three-dimensional space. A two-dimensional Motion Analysis system was used to determine how the LMM compared to this system in two-dimensional space. The results indicated that the LMM was about twice as accurate as the video-based motion evaluation system. It was also pointed out that the LMM provides an inexpensive means of monitoring trunk motion in a working environment. In the past few years the LMM has been extensively used for identifying and quantifying back motion factors associated with the risk of developing low back disorders (Marras et al. 1990; Ferguson, et al. 1992; Marras et al. 1993a, 1993b). This attributes of the LMM made it a measurement device of choice for use in the present study.

3.1.4 Experimental Lay-out and Data Collection

An area of 13m^2 was partitioned by a set of wooden screens and curtains in the Ergonomics Laboratory of the Department of Human Movement Studies, Rhodes University, in order to eliminate other infrared sources, such as sunlight, from the

V-scope working area. There was no extreme of ambient temperature which could influence the performance of the subjects during data collection.

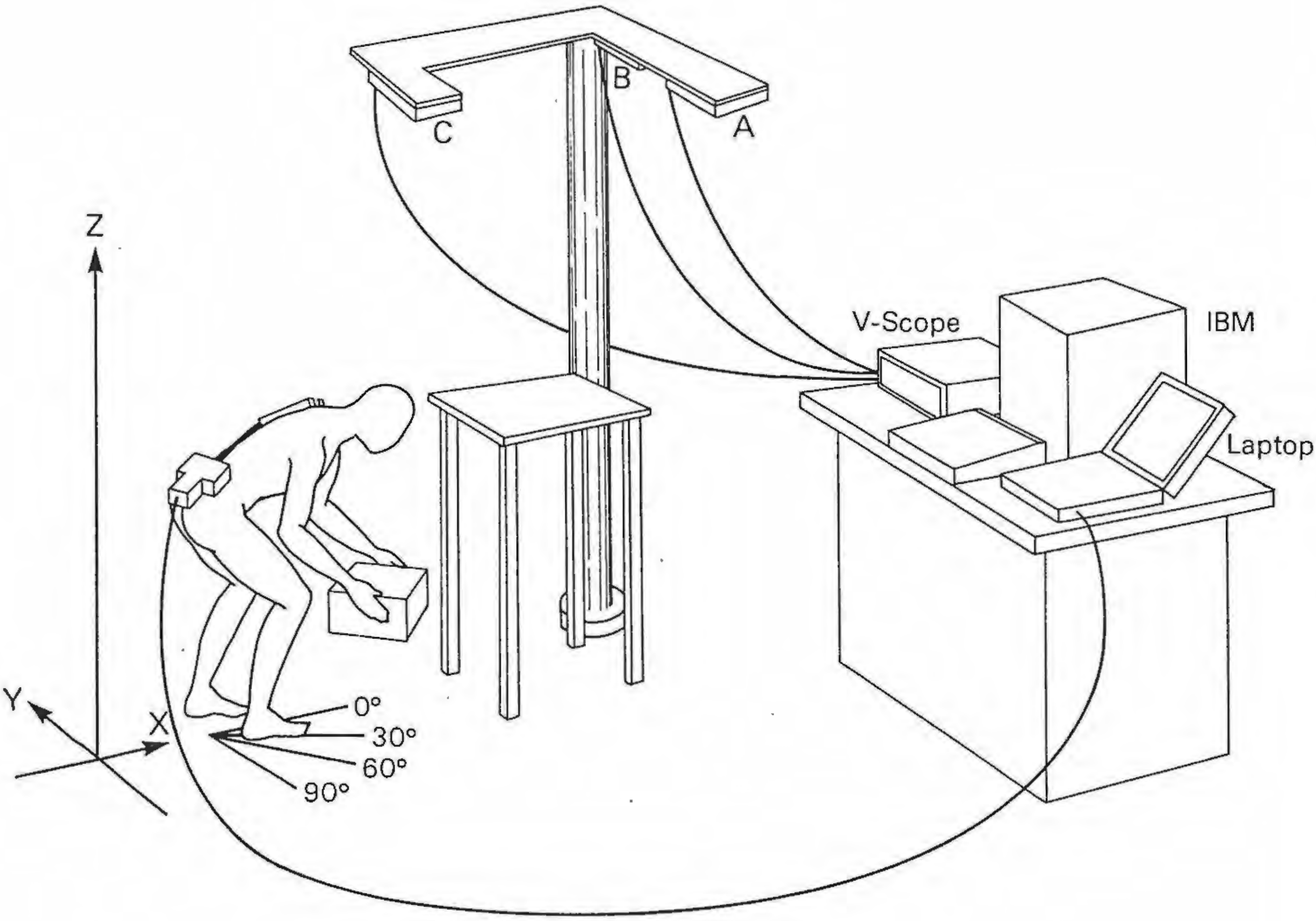

Figure 17 Experimental set-up

Figure 17 shows the experimental set-up. The V-scope ultrasonic motion analyser was used for measuring the movements of the box being lifted in the study. Three V-scope towers were installed above the working area and a V-scope button was attached to the centre of the box. The sampling time was set at 10ms. The V-scope microcomputer was connected to three towers through three supplied cables. During the experiment, all data were sent to an IBM-compatible personal computer to which the V-scope microcomputer was connected for further data process and display.

During the manual lifting activity, the kinematics of the lumbar spine of each subject were recorded using the exoskeleton. The data capture rate of the LMM is 60Hz. The LMM was carried on the back of the subject throughout the experiment. The output of LMM is transmitted through the umblical cable to a laptop, where spinal kinematics are calculated.

Each subject was required to participate in two sessions in the Ergonomics Laboratory. In the first session, subjects were thoroughly briefed as to the purpose of the experiments and asked to sign informed consent forms (see Appendix 1). Thereafter relevant anthropometric data were obtained, isometric strength was measured and finally each subject performed the eight lifting tasks to ensure familiarisation with the experimental procedures, laboratory setting and equipment. During the second session the kinematic data reflecting lumbar spinal motions and those of the box being lifted were simultaneously recorded by Lumbar Motion Monitor and V-scope, respectively, while the subject was engaged in the lifting tasks.

3.1.5 Experimental Design

Two independent variables, positional asymmetry and lifting grip height, were manipulated in the study. Asymmetry had four levels: 0^0 (sagittal symmetric); 30^0; 60^0 and 90^0. The experimental positions were marked on the floor as shown in Figures 17 and 18. A pair of rubberised footprint cut-outs was affixed to the floor to standardize the required standing position. The lifting grip height had two levels: 150mm and 500mm. The combination of these two independent variables created a 4x2 factorial design. The eight lifting tasks were randomized in respect of order of presentation. This randomization of tasks created a situation similar to that of a sorting operation in industry. In this manner an experimental layout resembling a realistic industrial task was used to quantify trunk motion characteristics.

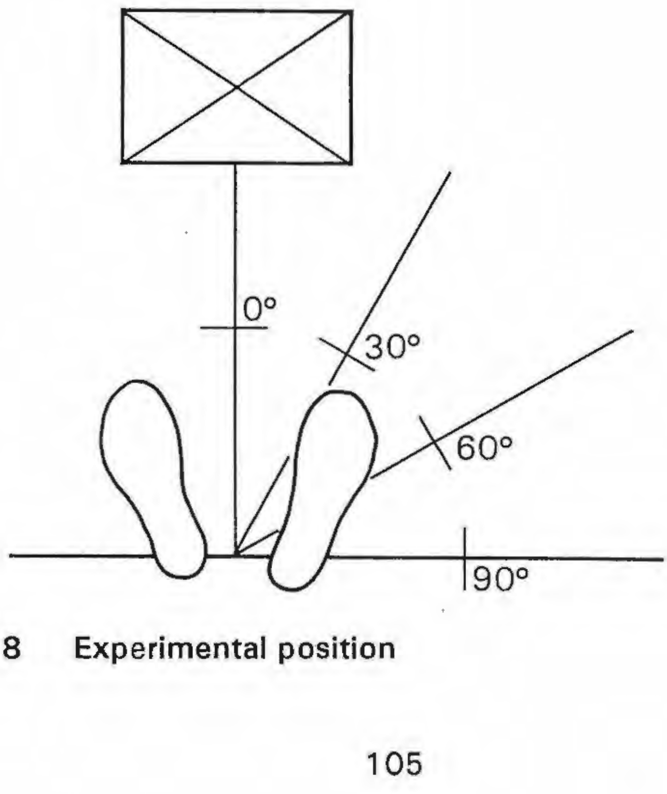

Figure 18 Experimental position

A total of 62 dependent variables, 41 involving kinetic measurements of manual lifting in the study and 21 involving kinematic characteristics of the dynamic lumbar spine, were measured. These dependent variables are listed in Tables IV and V. Generally, the maximum and average measurements of the various variables were recorded during the investigation. For the lifting, 13 temporal factors, 8 linear velocity, 8 linear acceleration measurements and 8 force measurements were measured on three axes in three dimensional space. In addition, 1 work and 2 power measurements were calculated for the lifter performing the external lifting task. For the human spine, 6 temporal and 3 spatial factors, 6 angular velocity and 6 angular acceleration measurements were recorded in each of the sagittal, frontal and horizontal planes.

Table IV Kinetic Measurements of Manual Lifting

	Temporal Factors (s)	
1	T_{max}	Time at target height.
2	T_{Vmax}	Time at maximum resultant velocity.
3	$T_{V(X)}$	Time at maximum component velocity on X-axis.
4	$T_{V(Y)}$	Time at maximum component velocity on Y-axis.
5	$T_{V(Z)}$	Time at maximum component velocity on Z-axis.
6	T_{Amax}	Time at maximum resultant acceleration.
7	$T_{A(X)}$	Time at maximum component acceleration on X-axis.
8	$T_{A(Y)}$	Time at maximum component acceleration on Y-axis.
9	$T_{A(Z)}$	Time at maximum component acceleration on Z-axis.
10	T_{Fmax}	Time at maximum resultant force.
11	$T_{F(X)}$	Time at maximum component force on X-axis.
12	$T_{F(Y)}$	Time at maximum component force on Y-axis.
13	$T_{F(Z)}$	Time at maximum component force on Z-axis.
14	T_{Pmax}	Time at maximum power.
	Velocity ($m.s^{-1}$)	
15	V_{max}	Maximum resultant velocity.
16	$V_{(X)}$	Maximum component velocity on X-axis.
17	$V_{(Y)}$	Maximum component velocity on Y-axis.
18	$V_{(Z)}$	Maximum component velocity on Z-axis.
19	V_{mean}	Average resultant velocity.
20	$V_{(X)mean}$	Average component velocity on X-axis.
21	$V_{(Y)mean}$	Average component velocity on Y-axis.
22	$V_{(Z)mean}$	Average component velocity on Z-axis.
	Acceleration ($m.s^{-2}$)	
23	A_{max}	Maximum resultant acceleration.
24	$A_{(X)}$	Maximum component acceleration on X-axis.
25	$A_{(Y)}$	Maximum component acceleration on Y-axis.
26	$A_{(Z)}$	Maximum component acceleration on Z-axis.
27	A_{mean}	Average resultant acceleration.
28	$A_{(X)mean}$	Average component acceleration on X-axis.
29	$A_{(Y)mean}$	Average component acceleration on Y-axis.
30	$A_{(Z)mean}$	Average component acceleration on Z-axis.
	Force (N)	
31	F_{max}	Maximum resultant force.
32	$F_{(X)}$	Maximum component force on X-axis.
33	$F_{(Y)}$	Maximum component force on Y-axis.
34	$F_{(Z)}$	Maximum component force on Z-axis.
35	F_{mean}	Average resultant force.
36	$F_{(X)mean}$	Average component force on X-axis.
37	$F_{(Y)mean}$	Average component force on Y-axis.
38	$F_{(Z)mean}$	Average component force on Z-axis.
	Power (W)	
39	P_{max}	Maximum power.
40	P_{mean}	Average power.
	Work (J)	
41	W_K	Total mechanical work.

Table V **Kinematic characteristics of lumbar spine**

	Temporal Factors (s)	
1	$T_{V(S)}$	Time at maximum angular velocity in the sagittal plane.
2	$T_{V(F)}$	Time at maximum angular velocity in the frontal plane.
3	$T_{V(H)}$	Time at maximum angular velocity in the horizontal plane.
4	$T_{A(S)}$	Time at maximum angular acceleration in the sagittal plane.
5	$T_{A(F)}$	Time at maximum angular acceleration in the frontal plane.
6	$T_{A(H)}$	Time at maximum angular acceleration in the horizontal plane.
	Range of Motion (degree)	
7	$ROM_{(S)}$	Sagittal range of motion.
8	$ROM_{(F)}$	Frontal range of motion.
9	$ROM_{(H)}$	Horizontal range of motion.
	Angular Velocity (degrees.s^{-1})	
10	$V_{(S)}$	Maximum angular velocity in the sagittal plane.
11	$V_{(F)}$	Maximum angular velocity in the frontal plane.
12	$V_{(H)}$	Maximum angular velocity in the horizontal plane.
13	$V_{(S)mean}$	Average angular velocity in the sagittal plane.
14	$V_{(F)mean}$	Average angular velocity in the frontal plane.
15	$V_{(H)mean}$	Average angular velocity in the horizontal plane.
	Angular Acceleration (degrees.s^{-2})	
16	$A_{(S)}$	Maximum angular acceleration in the sagittal plane.
17	$A_{(F)}$	Maximum angular acceleration in the frontal plane.
18	$A_{(H)}$	Maximum angular acceleration in the horizontal plane.
19	$A_{(S)mean}$	Average angular acceleration in the sagittal plane.
20	$A_{(F)mean}$	Average angular acceleration in the frontal plane.
21	$A_{(H)mean}$	Average angular acceleration in the horizontal plane.

3.1.6 Analytical Protocol

In the present experimental set-up (Figure 17), the 3-D V- scope tower configuration provided a right handed orthogonal co-ordinate {x', y', z'}. The origin of the co-ordinate was placed at the centre of the ultrasonic receiver of tower B. The x' axis was vertical pointing downwards. The y' and z' axes were in the horizontal plane and orthogonal to each other (Figure 19).

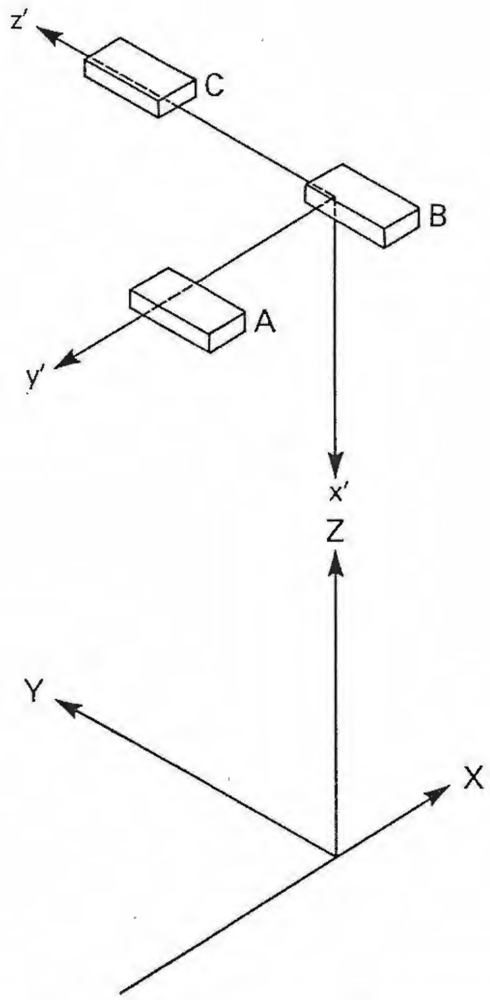

Figure 19 Three-dimensional co-ordinate set-up for the experiment

For the convenience of succeeding analyses, a new co-ordinate {X, Y, Z} was derived from the V-scope co-ordinate {x', y', z'}. The X and Y axes in the {X, Y, Z} co-ordinate were in the horizontal plane and Z axis was vertical pointing upwards (Figure 19).

During investigation, the V-scope system directly provided three dimensional co-ordinate values of displacement, velocity and acceleration. The equations for the relationship among the V-scope co-ordinates {x', y', z'} and the derived co-ordinates {X, Y, Z} are as follows:

1) Displacement,

$$X = -y';$$

$$Y = z';$$

$$Z = -x';$$

2) Velocity,

$$V_X = -v_{y'};$$

$$V_Y = v_{z'};$$

$$V_Z = -v_{x'};$$

3) Acceleration,

$$A_X = -a_{y'};$$

$$A_Y = a_{z'};$$

$$A_Z = -a_{x'}.$$

Trajectorial velocity (V) and acceleration (A) were obtained from the following equations:

$$V = \sqrt{V_X^2 + V_Y^2 + V_Z^2}$$

$$A = \sqrt{A_X^2 + A_Y^2 + A_Z^2}$$

A change in an object's motion is related to an unbalanced external force acting on it. Newton's second law of motion as expressed by Blanchard **et al.** (1960), states:

> *The rate of change of momentum of an object is directly proportional to the net force applied to it and is in the direction of the applied force.(pp.8)*

Momentum, as the product of mass and velocity, relative to the Second Law, indicates that:

$$\frac{d}{dt}m \cdot V \propto F$$

For constant **m**,

$$m \cdot \frac{dV}{dt} = m \cdot a \propto F$$

Where **a** is acceleration, **v** is velocity, **t** is time, **F** is force, and **m** is mass. We can choose the unit of force in such a way that the constant of proportionality in the above equation is 1.

Then,

$$F = m \cdot a$$

In the present study the box was moved from one rest state to another rest state in three dimensional space due to the force which the subject applied to it. This force can be resolved into three components $\mathbf{F_X}$, $\mathbf{F_Y}$ and $\mathbf{F_Z}$ (Figure 20). In addition, there is a gravitational force acting on the box. This gravitational force is equal to the weight of the box directed vertically downwards. In accordance with Newton's second law of motion, three equations were derived for force calculation.

$$F_X = m \cdot A_X$$

$$F_Y = m \cdot A_Y$$

$$F_Z = m \cdot A_Z + m \cdot g$$

where: $F_{X,Y,Z}$ = component forces on X, Y and Z-axes;

$A_{X,Y,Z}$ = component accelerations on X, Y and Z-axes;

m = mass of the box;

g = acceleration of gravity.

The trajectorial force (**F**) which the subject applied to the box was obtained from the equation below.

$$F = \sqrt{F_X^2 + F_Y^2 + F_Z^2}$$

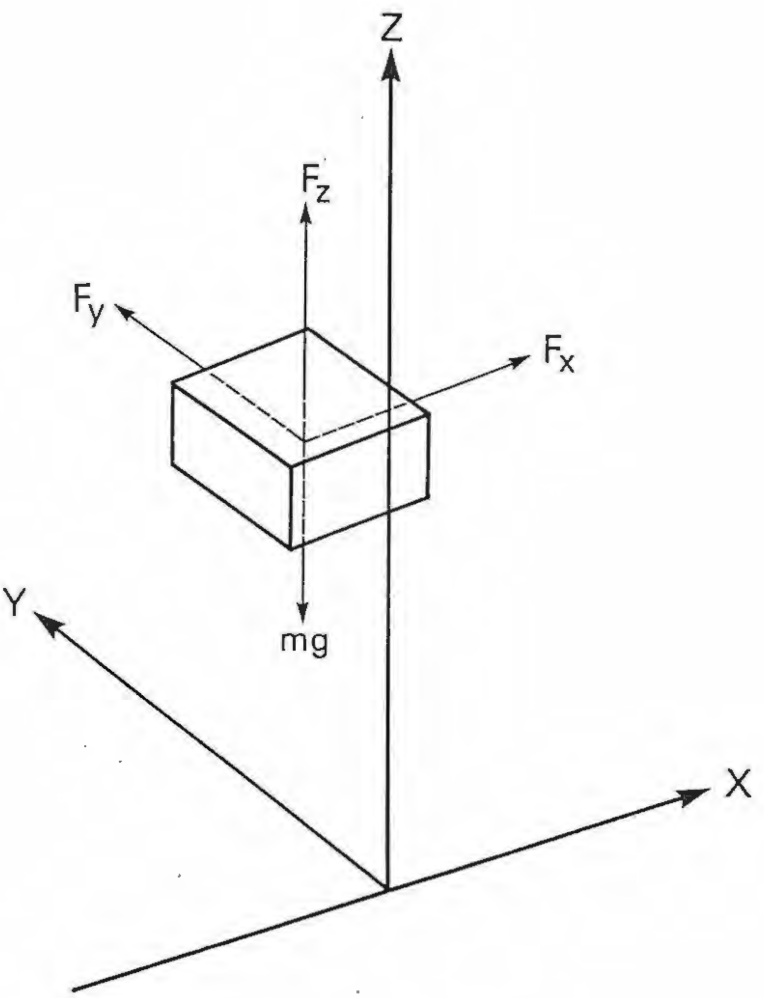

Figure 20 Forces acting on the box

The work (**W_k**) done by the subject to lift the box was defined to be,

$$W_k = \int_{S1}^{S2} \boldsymbol{F \cdot dS}$$

Where **F** is the trajectorial force exerted by the subject and **ds** is an increment of distance through which the force is applied.

The power (**P**) output of the subject doing the external work was calculated by following equation:

$$P = F \cdot V$$

where: F = trajectorial force;

V = trajectorial velocity.

3.1.7 Statistical Analysis

A number of dependent measures, relative to the kinematics of the human lumbar spine and kinetics of the act of lifting, were analysed statistically to evaluate the effects of the independent variables: task asymmetry and lift grip height. The test hypothesis was that no significant kinematic and kinetic differences exist in lifts as a function of task asymmetry and initial height.

The 0.05 level of probability was employed throughout the statistical treatment of the data to test significance of differences, variability and relationship. The present study utilized the STATGRAPHICS Statistical Graphics System (version:6.0). Descriptive statistics were run to provide a basis for testing normality and homogeneity of variance of data. Two-way ANOVA's were run to test whether there was a significant difference between the dependent variables under the various experimental conditions. One-way ANOVA's were conducted when the interaction between independent variables significantly affected dependent variables. Multiple regression was used to indicate trends of the dependent variables against independent variables.

3.1.8 Subjects

Most ergonomics research involves the use of human beings as subjects about whom relevant data are obtained. Male Caucasians have been widely used as subjects in ergonomic studies undertaken in Europe and North America. Data bases and guidelines have been developed for job design and assessment purposes in most developed countries (Snook, 1978; NIOSH, 1981; Mital, 1984a; Ayoub and Mital, 1989; Buis, 1990; Snook and Ciriello, 1991). In the case of developing countries, unfortunately, there are relatively few available data. Developing countries need and can benefit from modern technology developed in more highly industrialized countries and at present a great deal of technology transfer is undertaken from First to Third World countries. However, direct use of First World advanced modern technology in the Third World, without consideration of geophysical, sociocultural, and labour force differences, may cause immense losses in financial resources and manpower (Shahnavaz, 1987; Abeysekera, 1990; Scott, 1993). In some cases, the technology developed in industrialized countries is totally inapplicable to developing countries (Abeysekera, 1990; Evans, 1990)

In South Africa, fewer than 16% of the potential workforce is comprised of people of European descent (Walraven, 1988). Most are Black Africans. In the present study, eleven Black males, ranging in age between 27 to 56 years, were selected from the employees working in the Grounds and Gardens Section of Rhodes University. Selected subjects were all in good health with no contra-indicating medical history relative to lifting capability. Manual materials handling is a major component of their daily work. Participation as a subject was entirely voluntary and the subjects were at liberty to withdraw from the study at any time and for any reason. During the experiment subjects were dressed in their normal work clothing, i.e. overalls and protective work boots.

Anthropometric measurements and isometric strengths were assessed before participation in the lifting tasks. A Holtain Stadiometer was used to measure the subject's stature. This measurement was made with the subject standing barefoot, head erect, looking straight forward. The subject's body mass was measured by using a Toledo Scale with an accuracy of ± 0.02kg. After the setting was zeroed, the subject stood quietly on the scale, barefoot in minimum clothing. An investigator recorded the body mass from the scale monitor.

The procedure to assess isometric strengths has been summarized by experts in considerable detail (Chaffin, 1975; Ayoub and Mital, 1989). The description of the characteristics of the strength measuring devices used can be found in Chaffin and Andersson (1984). The subject's isometric arm, back and leg strengths were measured using the procedure employed by Chaffin **et al.** (1977). The subject's grip strength was measured for both hands following the procedure outlined in Kamon and Goldfuss (1978).

Basic data reflecting mean age, body weight, stature and isometric strengths are summarized in Table VI. Back and leg strengths were higher by 69% and 11%, respectively, and arm strength was lower by 45% compared to values reported by Chaffin **et al.** (1977) for male American industrial subjects. No difference was found between the grip strength measured in the present study and that reported by Kamon and Goldfuss (1978).

Table VI **Anthropometric and isometric strength measurements**

		MEAN	S.D.	RANGE
AGE (Yr)		44	10.3	27 -- 56
BODY WEIGHT (N)		701.7	188.8	524.6 -- 1139.1
STATURE (mm)		1752	62	1626 -- 1866
GRIP STRENGTH (N)	Left	464.1	77.7	305.1 -- 567.0
	Right	481.8	85.6	259.0 -- 564.1
ARM STRENGTH (N)		210.5	126.1	49.1 -- 461.1
BACK STRENGTH (N)		920.9	229.1	539.6 -- 1275.3
LEG STRENGTH (N)		1047.9	215.3	627.8 -- 1344.0

3.2 RESULTS AND DISCUSSION

3.2.1 Three-dimensional Kinematics of the Human Spine

The kinematics of the human spine in three-dimensional space were recorded in sagittal, frontal and horizontal planes simultaneously, when the subjects performed lifting tasks from two vertical lift heights (150mm and 500mm) and under four task asymmetry conditions (0^0, 30^0, 60^0 and 90^0). Tables VII and VIII show the overall mean, standard deviation and significant effects in respect of each variable.

Table VII Temporal factors of lumbar spine

Variables		(1) 150mm Vertical Height				(2) 500mm Vertical Height				Significance (p<0.05)
		A 0^0	B 30^0	C 60^0	D 90^0	E 0^0	F 30^0	G 60^0	H 90^0	
$T_{V(S)}$	MEAN	0.39	0.37	0.35	0.38	0.3	0.34	0.35	0.36	
	S.D.	0.15	0.17	0.14	0.1	0.1	0.16	0.12	0.12	
$T_{V(F)}$	MEAN	1.24	0.88	0.73	0.72	0.46	0.65	0.76	0.61	1-2; A-D
	S.D.	0.32	0.57	0.48	0.49	0.47	0.5	0.51	0.47	
$T_{V(H)}$	MEAN	0.89	0.9	0.85	1.05	0.75	0.84	0.84	0.89	1-2;A-B;A-C;A-D
	S.D.	0.3	0.38	0.34	0.2	0.52	0.37	0.17	0.3	
$T_{A(S)}$	MEAN	0.89	0.19	0.18	0.21	0.14	0.13	0.15	0.21	A-B;A-D
	S.D.	0.3	0.08	0.08	0.1	0.06	0.06	0.06	0.09	E-F;E-H
$T_{A(F)}$	MEAN	0.18	0.42	0.41	0.7	0.34	0.74	0.6	0.45	1-2
	S.D.	0.08	0.41	0.37	0.4	0.26	0.36	0.48	0.39	
$T_{A(H)}$	MEAN	0.8	0.66	0.72	0.7	0.66	0.4	0.66	0.56	1-2
	S.D.	0.21	0.3	0.33	0.32	0.55	0.21	0.24	0.36	

The temporal measures differed in random fashion (Table VII). In the sagittal plane, the time taken to attain maximum velocity ($T_{V(S)}$) showed no significant difference between lifts at various conditions; and the time taken to achieve maximal acceleration ($T_{A(S)}$) in the sagittal lift was significantly different from that during twisted lifts at 30^0 and 90^0 deviations at the two vertical heights.

A two-way ANOVA displayed a significant interaction on the time measures in the frontal plane between the lower (150mm) and the higher (500mm) levels. Therefore a one-way ANOVA was run, which showed that the time taken to attain maximum velocity in the sagittal lift was significantly longer than that in 90^0 at the lower level. In the horizontal plane a significant difference was found in the temporal factors between the two initial lift levels. At the lower lift height the time taken to achieve maximum velocity in the sagittal lift was significantly different from that at 30^0, 60^0 and 90^0 deviations.

Table VIII Kinematics of lumbar spine

Variables		(1) 150mm Vertical Height				(2) 500mm Vertical Height				Significance ($p<0.05$)
		A 0°	B 30°	C 60°	D 90°	E 0°	F 30°	G 60°	H 90°	
$ROM_{(S)}$	MEAN	73.5	76.4	76.4	76.5	55.9	59.5	61.5	64.5	1-2
	S.D.	4.58	3.71	2.36	2.85	6.22	5.9	5.97	4.77	
$ROM_{(F)}$	MEAN	8.5	9.9	12.7	16.5	5.1	8.2	11.6	16.0	A-C;A-D;B-D;C-D
	S.D.	1.87	1.99	2.75	3.57	1.57	1.64	2.43	3.78	E-G;E-H;F-H;G-H
$ROM_{(H)}$	MEAN	3.8	4.3	4.7	5.0	1.5	3.5	4.6	5.5	1-2
	S.D.	0.62	0.55	0.59	0.95	0.41	0.98	0.46	0.68	E-F;E-G;E-H;F-H
$V_{(S)}$	MEAN	104.2	112.8	114.5	120.9	82.1	83.5	97.7	113.7	1-2;A-D
	S.D.	6.67	9.8	9.62	10.42	15.45	17.4	16.3	11.33	E-H
$V_{(F)}$	MEAN	18.7	18.4	19.5	23.8	8.8	14.5	16.8	23.4	1-2;A-D;B-D
	S.D.	1.84	1.76	2.41	3.67	1.42	3.19	3.02	4.57	E-H;F-H
$V_{(H)}$	MEAN	7.2	8.5	9.1	9.5	1.4	6.9	8.7	9.1	1-2
	S.D.	1.22	1.45	0.98	1.52	0.81	1.45	1.62	1.32	E-F;E-G;E-H
$V_{(S)mean}$	MEAN	33.4	32.8	33.0	32.3	21.8	24.2	27.0	31.5	1-2
	S.D.	5.62	6.7	5.67	5.91	9.31	7.26	6.53	6.14	E-H
$V_{(F)mean}$	MEAN	2.5	2.6	3.3	4.6	0.9	1.3	3.1	6.3	A-D;B-D
	S.D.	0.65	0.79	0.85	1.06	0.53	0.52	0.95	1.37	E-H;F-H
$V_{(H)mean}$	MEAN	1.7	2.0	2.4	2.7	0.2	1.7	2.0	2.2	
	S.D.	0.76	1.23	0.97	1.42	0.24	1.0	1.2	1.35	E-F;E-G;E-H
$A_{(S)}$	MEAN	439.0	495.4	504.5	513.5	241.6	376.2	408.9	500.2	1-2
	S.D.	44.4	50.2	41.7	78.5	55.5	38.9	30.6	67.8	
$A_{(F)}$	MEAN	77.6	76.2	76.8	94.0	36.8	63.2	69.3	97.5	1-2;A-D;B-D
	S.D.	4.21	5.82	9.72	10.16	6.84	5.7	26.74	42.72	E-H;F-H
$A_{(H)}$	MEAN	28.8	34.5	34.9	35.2	8.8	21.8	35.4	36.2	1-2
	S.D.	5.08	5.65	6.22	6.43	3.63	5.04	3.44	3.57	E-F;E-G;E-H;F-G;F-H
$A_{(S)mean}$	MEAN	4.1	4.5	3.7	2.2	5.5	9.8	7.9	4.2	1-2
	S.D.	2.54	4.56	5.06	1.99	6.12	11.08	7.15	5.43	
$A_{(F)mean}$	MEAN	1.4	1.5	2.3	4.5	1.4	1.9	3.4	3.8	A-D;B-D
	S.D.	1.2	1.26	2.6	2.81	1.73	1.78	2.54	2.84	E-H;F-H
$A_{(H)mean}$	MEAN	0.3	0.4	0.6	0.6	0.15	0.2	0.4	0.5	
	S.D.	0.34	0.34	0.55	0.41	0.33	0.21	0.29	0.52	

Sagittal plane

Figures 21, 22 and 24 display the sagittal range of motion, sagittal angular extension velocity and acceleration for the lifts at two vertical lift levels (150mm and 500mm) and four angular displacements: symmetrical (0^0) and asymmetrical (30^0, 60^0 and 90^0).

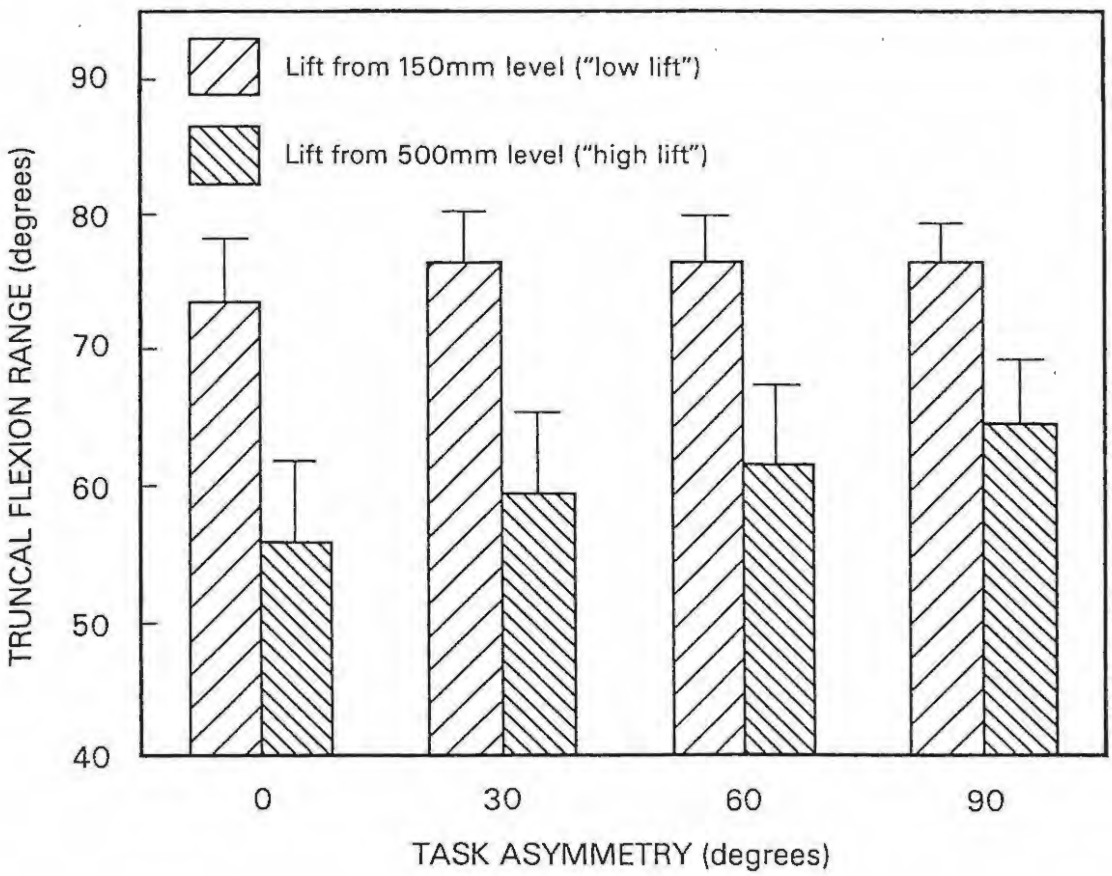

Figure 21 **Sagittal plane range of motion** (Variability indicators in this and subsequent bar graphs represent the standard deviation.)

Generally, as the lifting condition became lower and more asymmetric, the sagittal range of motion, peak velocity and peak acceleration increased. The range of motion, peak velocity and acceleration displayed more reactive response to asymmetry at the 500mm level than at the lower level (Figures 21, 23 and 25). The sagittal range of motion showed a statistically significant difference between the two levels.

The extension velocities for the lifts under four task asymmetry conditions followed similar patterns at the same vertical levels (Figure 22). The velocity reached a peak around 23% of total lifting time for all lifting tasks. This peak value increased with an increase in task asymmetry. Before and after the peak, velocities increased and decreased at almost the same rate at the same vertical level. At the 150mm level, the rate of decrease in velocity was arrested between 35% and 50% of the lifting cycle, (as shown in Figure 22 by two inflection points A and B). After point B the rate of decrease returned to its level before point A. The change in rate was more obvious for the two more symmetrical lifts (0^0 and 30^0) than for the two more asymmetrical lifts (60^0 and 90^0 deviations from the sagittal plane).

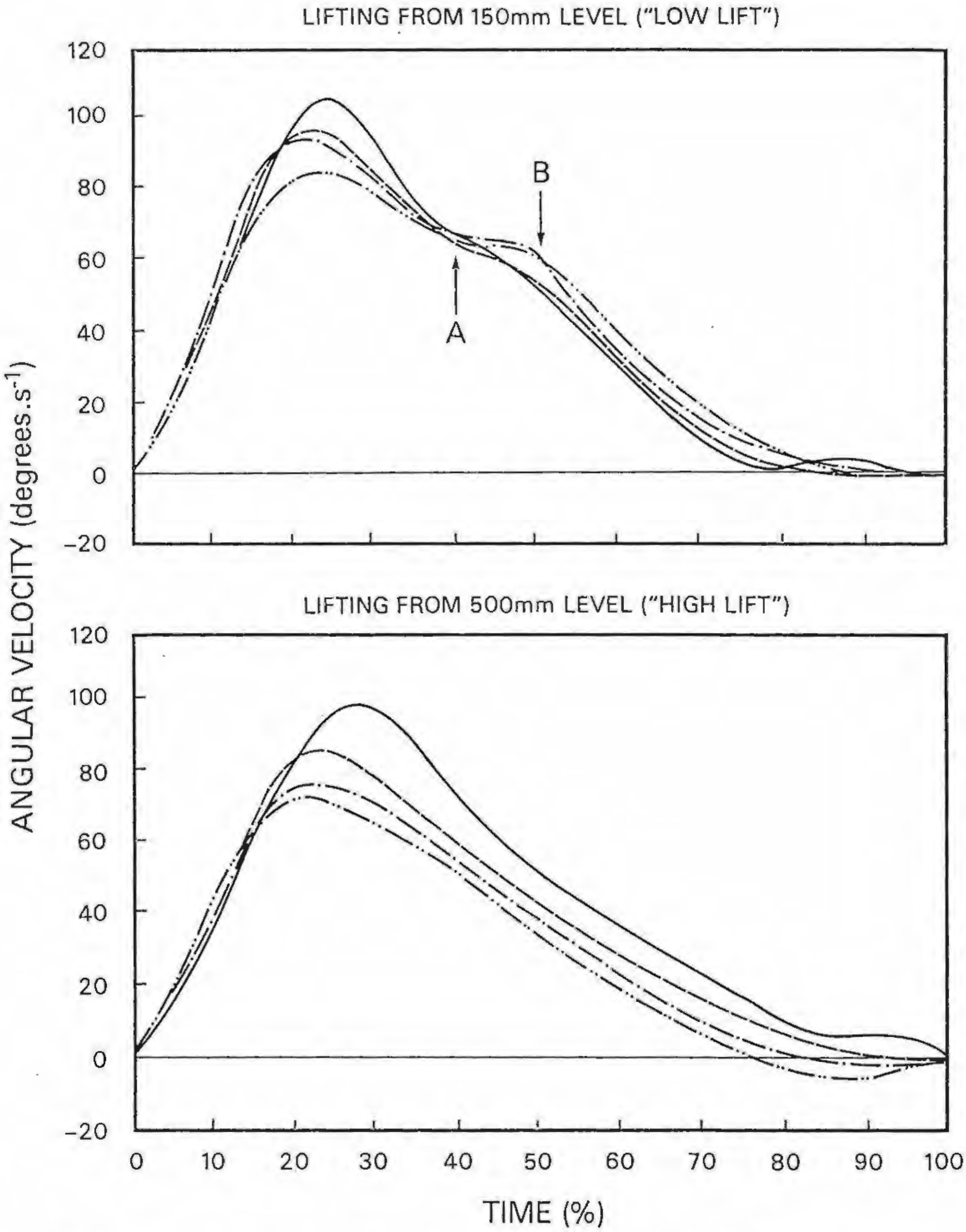

Figure 22 **Sagittal plane angular velocity traces** (Angular deviations from the sagittal plane indicated as follows: ——— (90°); — — (60°); —·— (30°); —··— (0°).)

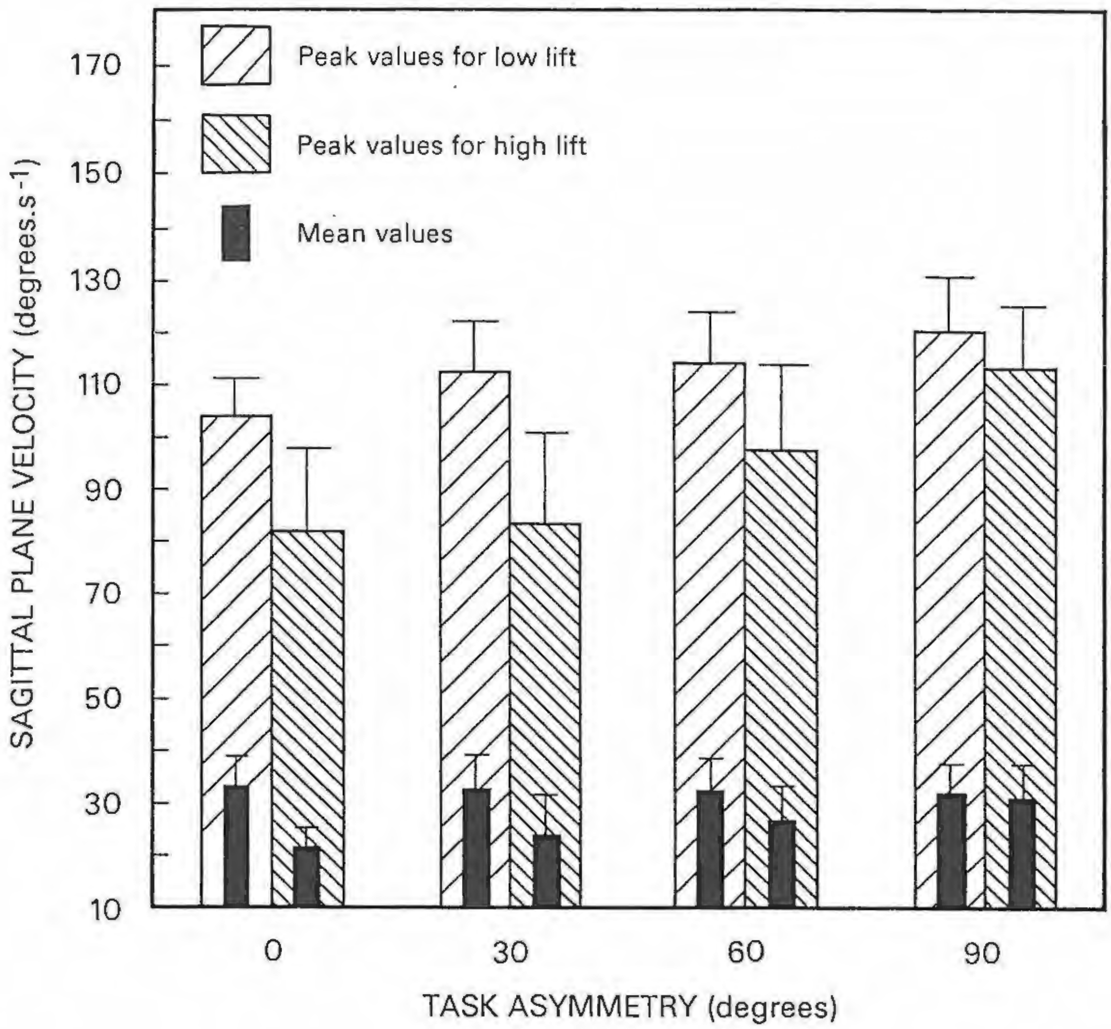

Figure 23 Sagittal plane peak and average spinal velocity

The average velocity increased with an increase in angular deviations of the load from the sagittal plane during "high" lifts (500mm) and slightly decreased during "low" lifts (150mm) (Figure 23). The average and peak velocities showed statistically significant differences between lifts as a function of initial lift height and a significant difference in the maximum velocity was found between the sagittal plane lifts and those involving the greatest asymmetry. The results of a two-way ANOVA indicated that the interaction of the vertical height and the task asymmetry significantly affected the sagittal average velocity. Therefore a one-way ANOVA was conducted separately on the sagittal average velocity at the two vertical levels. For low level lifts, the difference in average velocity between the sagittal lift and most asymmetrical lift was statistically significant.

The pattern of angular acceleration of the spine in the sagittal plane was more related to initial lift height than to degree of spinal twisting involved (Figure 24). Average accelerations at the 500mm level were higher than those at the 150mm level. It is evident from the graphic representation of the results in Figure 24 that peak accelerations were obtained around 10% of the lifting cycle for all lifting tasks. After 23% of the lifting cycle the trunk muscle started to decelerate the extension movement, producing an oscillating curve in the lifts initiated at the 150mm level. These waves were caused by changes in the decrease rate of the sagittal velocity at the 150mm vertical level (Figure 22).

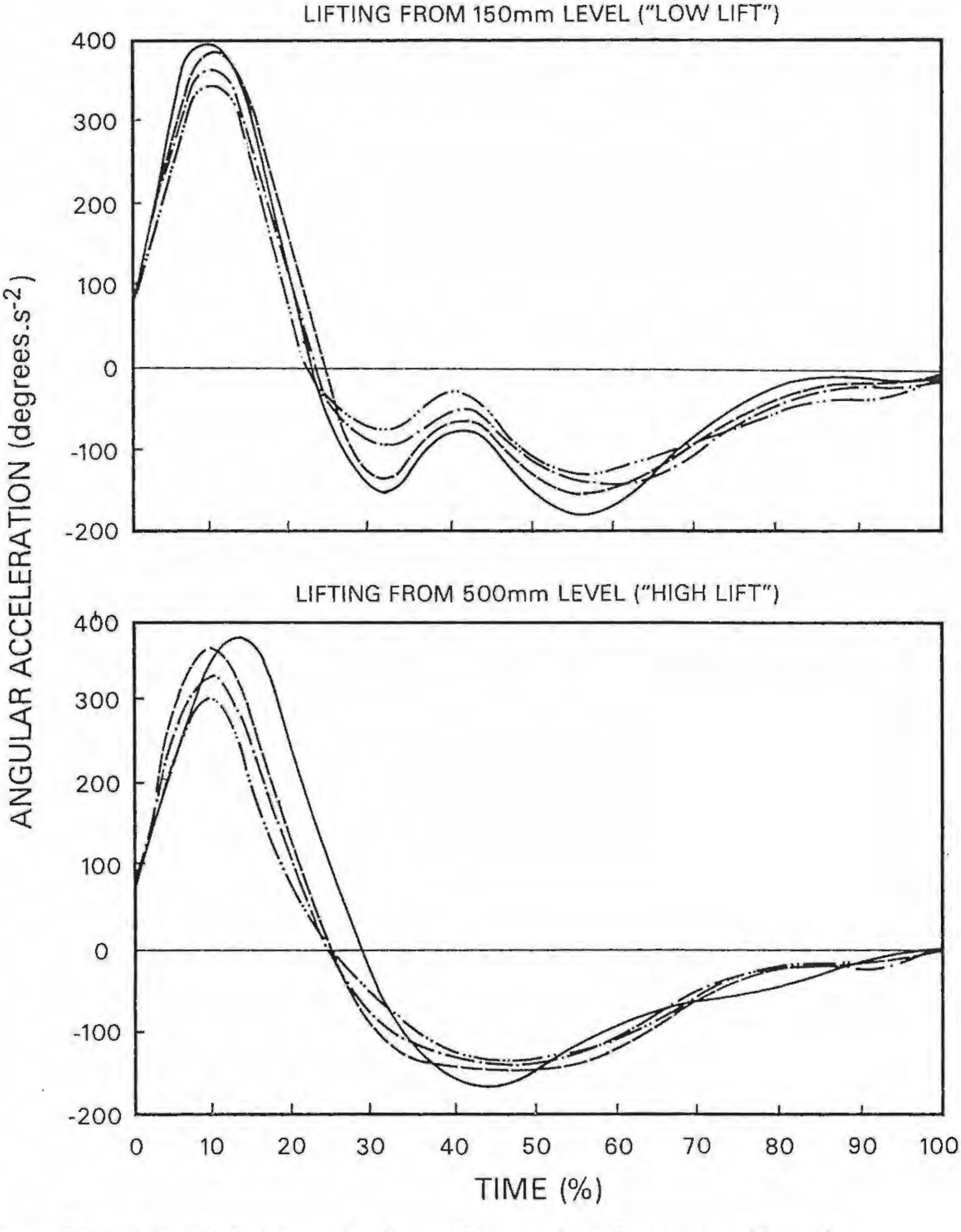

Figure 24 **Sagittal plane spinal angular acceleration traces** (Angular deviations from the sagittal plane indicated as follows:——— (90°); — — (60°); —·— (30°); —··— (0°).)

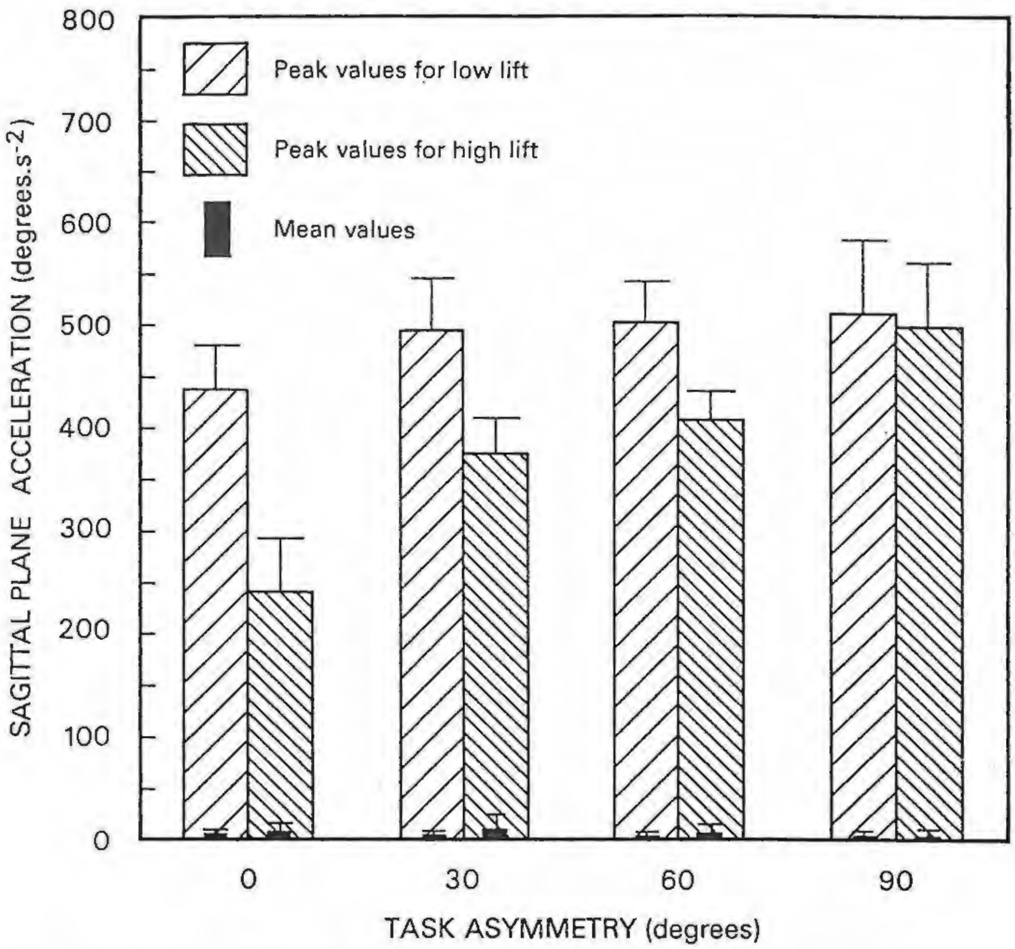

Figure 25 Sagittal plane peak and average spinal acceleration

It is clear that the lift undertaken at 90^0 to the sagittal plane provided the largest positive and negative accelerations, and the lift in the sagittal plane had the smallest peak positive and negative accelerations, regardless of initial lift height. The peak acceleration increased in tandem with task asymmetry and the average acceleration had a maximum value at 30^0 and lowest value at 90^0 (Figure 25). Both peak and average acceleration displayed a statistically significant difference between the lifts at the two vertical levels.

Frontal plane

The relationship between the frontal plane range of motion of the spine and the extent of task asymmetry is clearly linear (Figure 26). As expected the frontal plane range of motion increased as a function of increasing task asymmetry (ie. angular deviation of load from sagittal plane). The frontal plane range of motion for the sagittal lift was significantly different from the range of motion at 60^0 and 90^0 , regardless of initial height of object; and the lift at 90^0 was significantly different from those 30^0 and 60^0. There was no significant difference found in the frontal plane range of motion of the spine between two vertical levels.

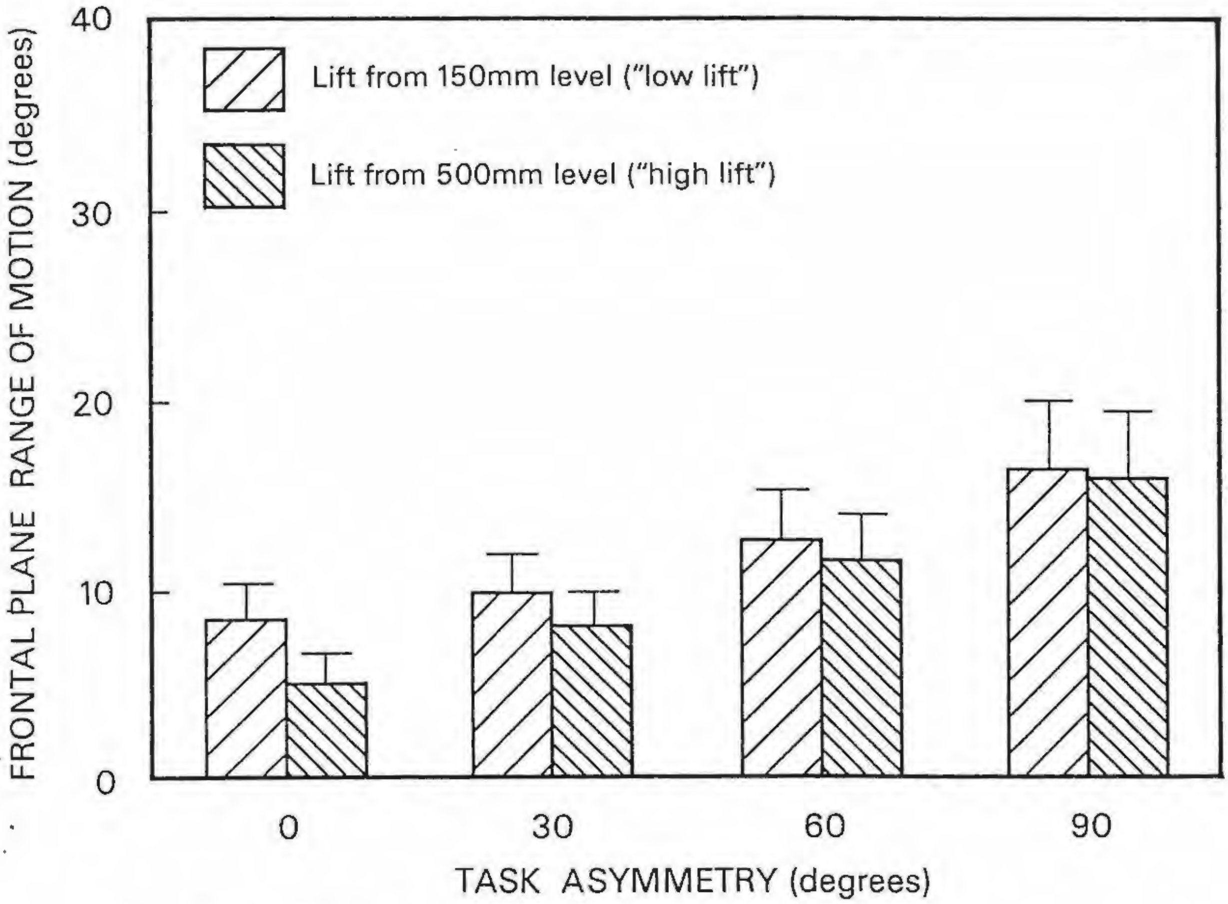

Figure 26 Frontal plane spinal range of motion

Figures 27 and 29 illustrate the frontal plane velocity and acceleration traces for the lifts at four task asymmetry conditions and two vertical lift heights. From Figure 27 and 29 it is clear that these velocities and accelerations fluctuated about the zero line during the range of lifting. However, the maximum peak velocity increased as a function of increasing task asymmetry (Figure 28). The results of a two-way ANOVA revealed that the maximum peak velocity was significantly different between the two vertical levels. The difference in the maximum peak velocity between lifts in the sagittal plane (0^0) and the lifts at 90^0 task asymmetry was statistically significant. At the 500mm level the maximum peak velocity increased at a much greater rate compared to the increase at the 150mm level. The difference between the maximum peak velocities at the two vertical levels decreased with an increase in task asymmetry. It is of interest to note that the maximum peak velocity for both levels achieved a similar value at 90^0 task asymmetry. The maximum peak velocity of the lift at 90^0 was significantly greater than that of the lift at 30^0 for both vertical lift heights.

Average velocity increased as a function of increasing task asymmetry for both vertical lift heights (Figure 28). At the 500mm level the average velocity exhibited dramatic increases in all deviations beyond 30^0 from the sagittal plane. A statistically significant difference in frontal plane average velocity was found between the lifts at 0^0 and at 90^0, and between the lifts at 30^0 and at 90^0.

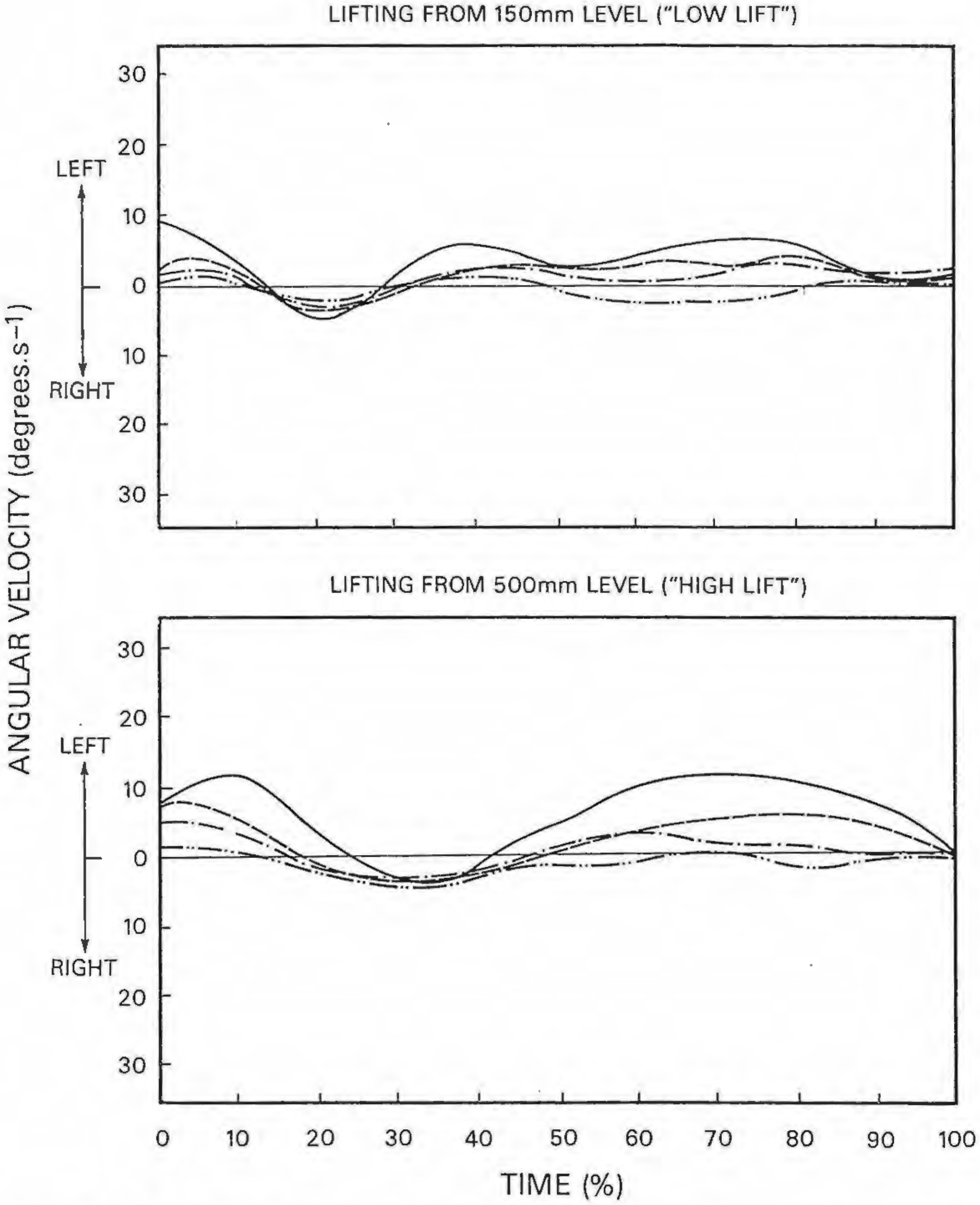

Figure 27 **Frontal plane angular velocity traces** (Angular deviations from the sagittal plane indicated as follows: ——— (90°); — — (60°); —·— (30°); —··— (0°).)

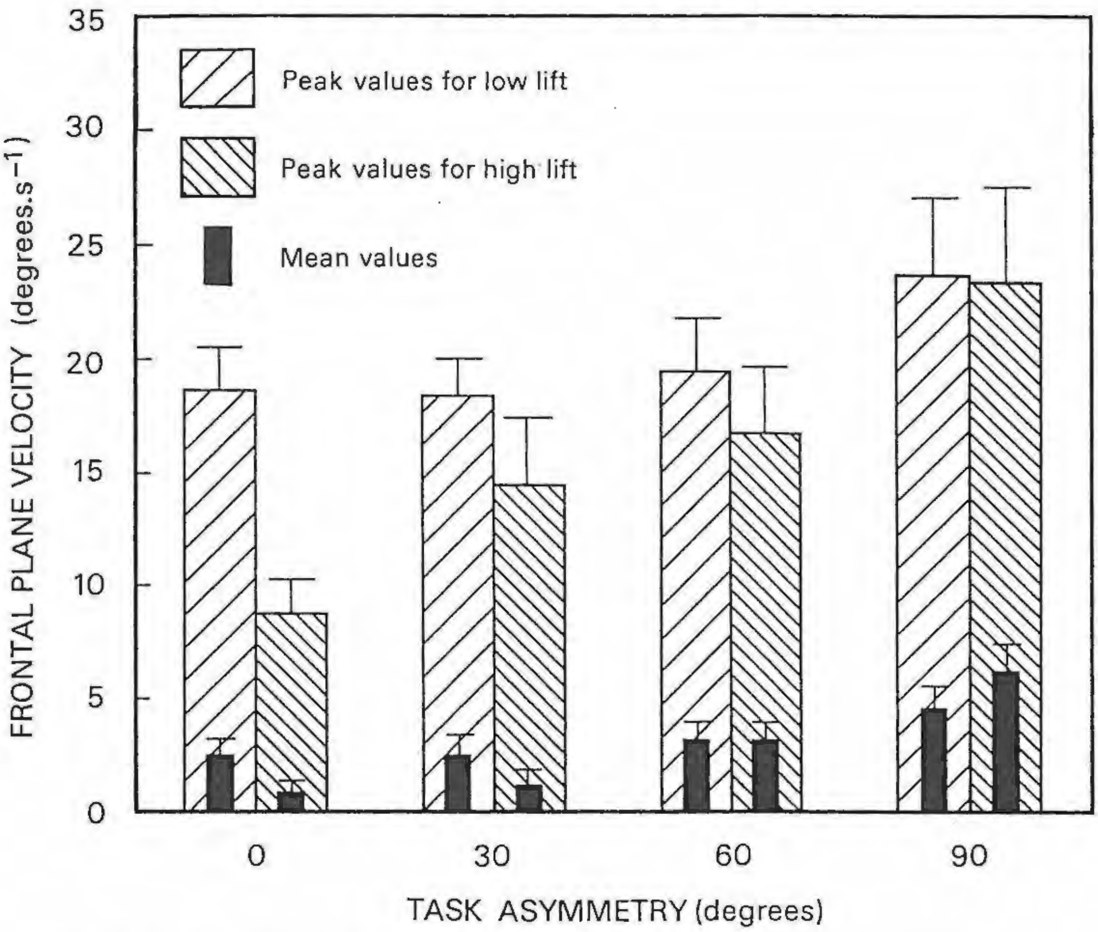

Figure 28 Frontal plane peak and average spinal velocity

The maximum peak acceleration remained almost constant between 0^0 to 60^0 during "low" level lifts and increased by 22% from 60^0 to 90^0 task asymmetry (Figure 30). In the "high" level lifts the maximum peak acceleration increased at a much reduced rate between 30^0 and 60^0 task asymmetry compared to the increases before 30^0 and after 60^0. The results of a two-way ANOVA indicated that the difference in the maximum peak acceleration was statistically significant between two of the vertical lift heights, and the maximum peak accelerations at 90^0 were significantly greater than those at 0^0 and 30^0 task asymmetry for both vertical levels.

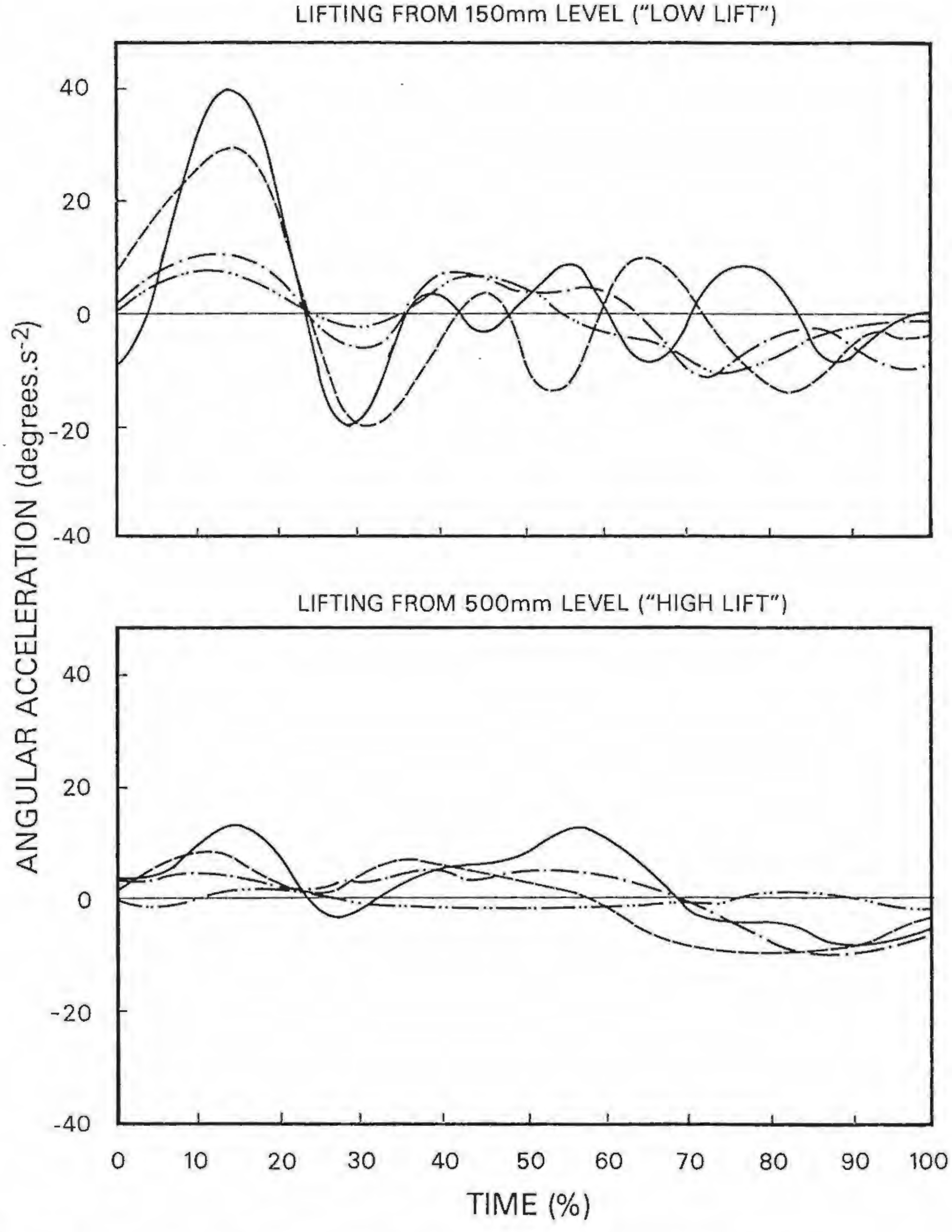

Figure 29 **Frontal plane spinal angular acceleration traces** (Angular deviations from the sagittal plane indicated as follows:——— (90^0); — — (60^0); —·— (30^0); —··— (0^0).)

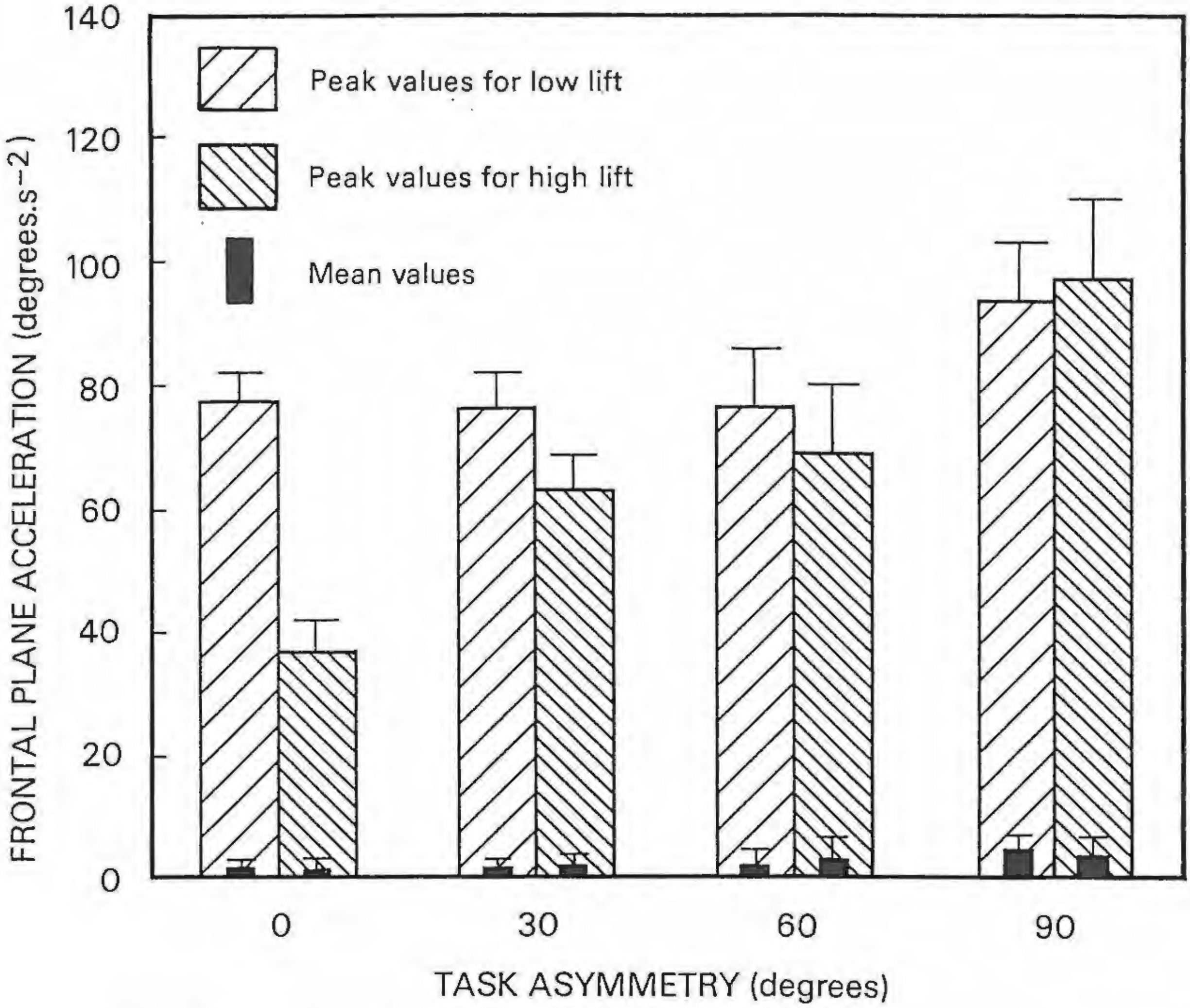

Figure 30 Frontal plane peak and average spinal acceleration

The average acceleration increased gradually with increased deviations from the sagittal plane (Figure 30). The increase was more dramatic beyond 30^0 asymmetry at the "low" level. The average accelerations for lifts at 30^0 and 60^0 deviations were lower in the case of 150mm lifts than 500mm lifts. Statistically significant differences in the average acceleration were found between sagittal plane lifts and those at 90^0 task asymmetry, and between 30^0 and 90^0 task asymmetry. There was no significant difference in the average acceleration between the two vertical levels.

Horizontal plane

Not surprisingly, the horizontal range of motion increased dramatically as a function of increasing task asymmetry (Figure 31). This increase occurred at a much greater rate at the higher lift level than at the lower level. The horizontal range of motion was significantly different between the two height levels.

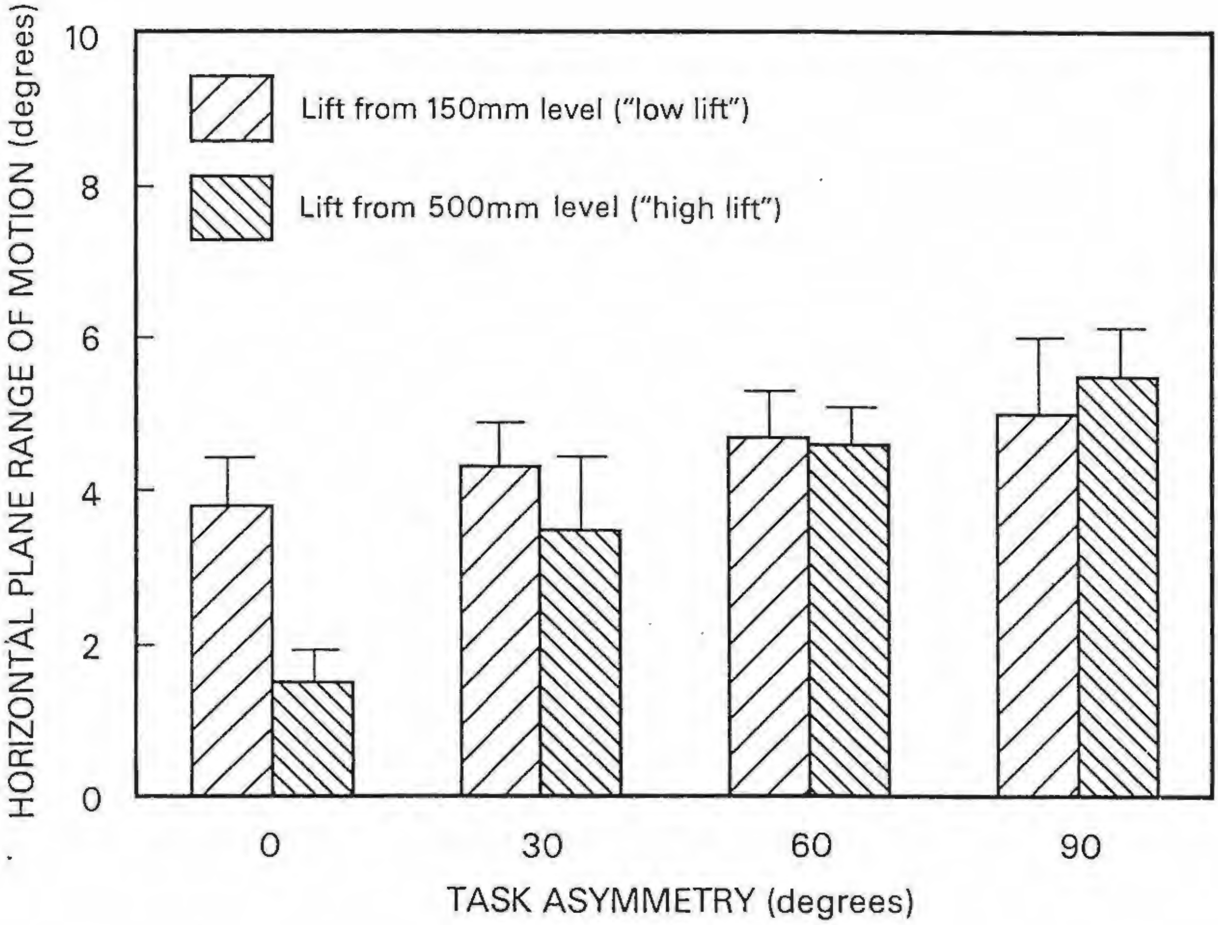

Figure 31 Horizontal plane spinal range of motion

A significant interaction was found between the vertical levels. The results of the one-way ANOVA showed that the range of motion in the sagittal plane at the 500mm level was significantly different from that at 30^0, 60^0 and 90^0 of task asymmetry. The difference in the range of motion between the lift at 30^0 and the lift at 90^0 was also significant.

Figure 32 demonstrates the horizontal plane angular velocity curves through the whole lifting motion for all eight lifting tasks. There was less movement in the horizontal plane when the lifts were performed closer to the sagittal plane. It is evident from Figure 32 that the velocities in the symmetrical lift(0^0) were closer to the zero line, particularly in the case of the lift at the 500mm level. Figure 32 also shows that the horizontal movement at the 150mm level was greater than that at the 500mm level. This may indicate that as lower lifts are performed, greater asymmetric movement is involved. The peak and average velocity increased while the task asymmetry increased from the sagittal plane 0^0 to 90^0 (Figure 33). At the 150mm level the average velocity increased at a constant rate and the peak velocity also increased at a constant rate after 30^0 task asymmetry. At the 500mm level the average and peak velocity generally increased but at a much reduced rate compared to the increase between 0^0 and 30^0 of task asymmetry. The results of the peak velocity analysis revealed statistically significant differences between the two vertical levels; however, there were no significant differences in average velocity between the two vertical levels. The results of the two-way ANOVA showed a significant interaction between the two vertical levels. Therefore, a one-way ANOVA was performed separately for the two vertical levels. The peak and average velocity during sagittal lifts at the 500mm level was significantly lower than peak and average velocities at 30°, 60° and 90° task asymmetry.

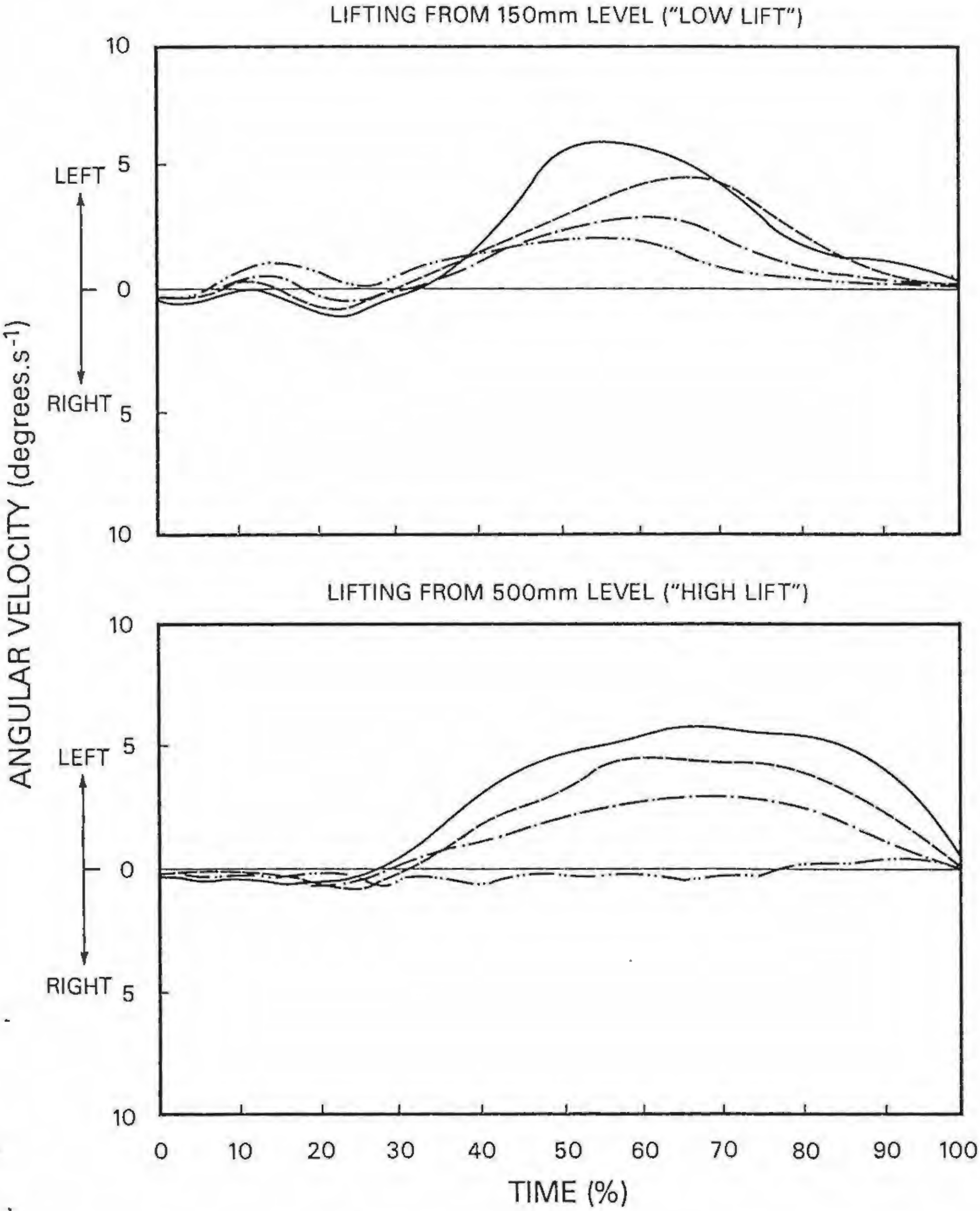

Figure 32 **Horizontal plane angular velocity traces** (Angular deviations from the sagittal plane indicated as follows: ——— (90°); — — (60°); —·— (30°); —··— (0°).)

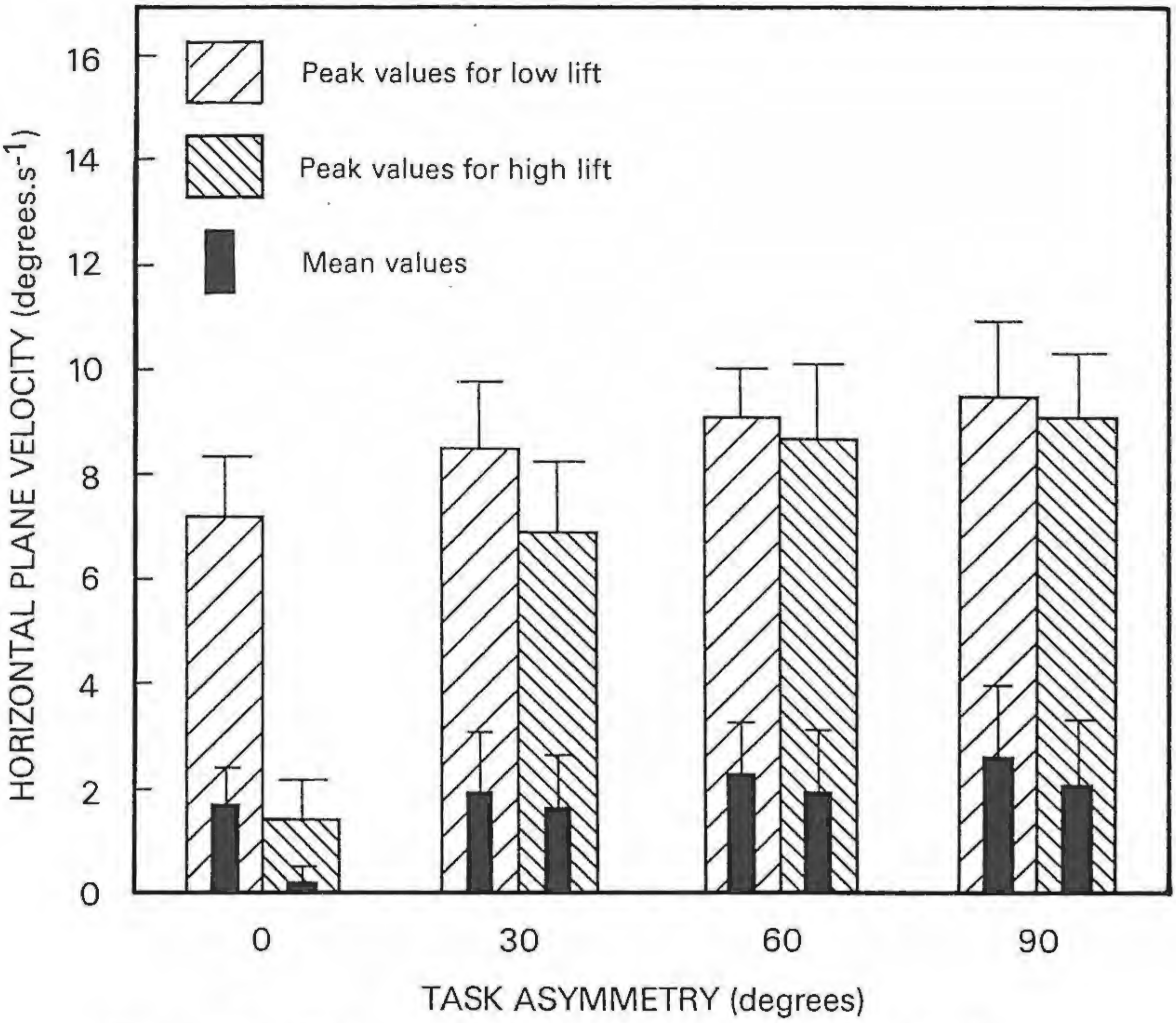

Figure 33 Horizontal plane peak and average spinal velocity

Figure 34 displays the horizontal acceleration curve patterns for the lifts at four task asymmetries and two vertical levels. Generally the horizontal acceleration showed an undulatory pattern. The amplitude and frequency of the undulation increased along with the increase of task asymmetry from 0^{0} to 90^{0} and decreased with an increase in vertical height from 150mm to 500mm. The undulatory patterns may indicate complex force applications in the horizontal plane during the manual lifting. The horizontal maximum and average acceleration increased as a function of increasing task asymmetry (Figure 35). At the level of 150mm the maximum acceleration increased at a slight greater rate from 0^0 to 30^0 than from 30^0 to 90^0. At the level of 500mm the maximum acceleration increased at a much higher rate from 0^0 to 60^0 task asymmetry than from 60^0 to 90^0 task asymmetry. The average acceleration increased in a very similar manner for the two vertical levels, and horizontal maximum acceleration showed a statistically significant difference between the levels. A two-way ANOVA showed significant interaction between levels, so a separate one-way ANOVA was carried out for each. The horizontal maximum acceleration at 0^0 at the level of 500mm was significantly different from those at 30^0, 60^0 and 90^0 deviations.

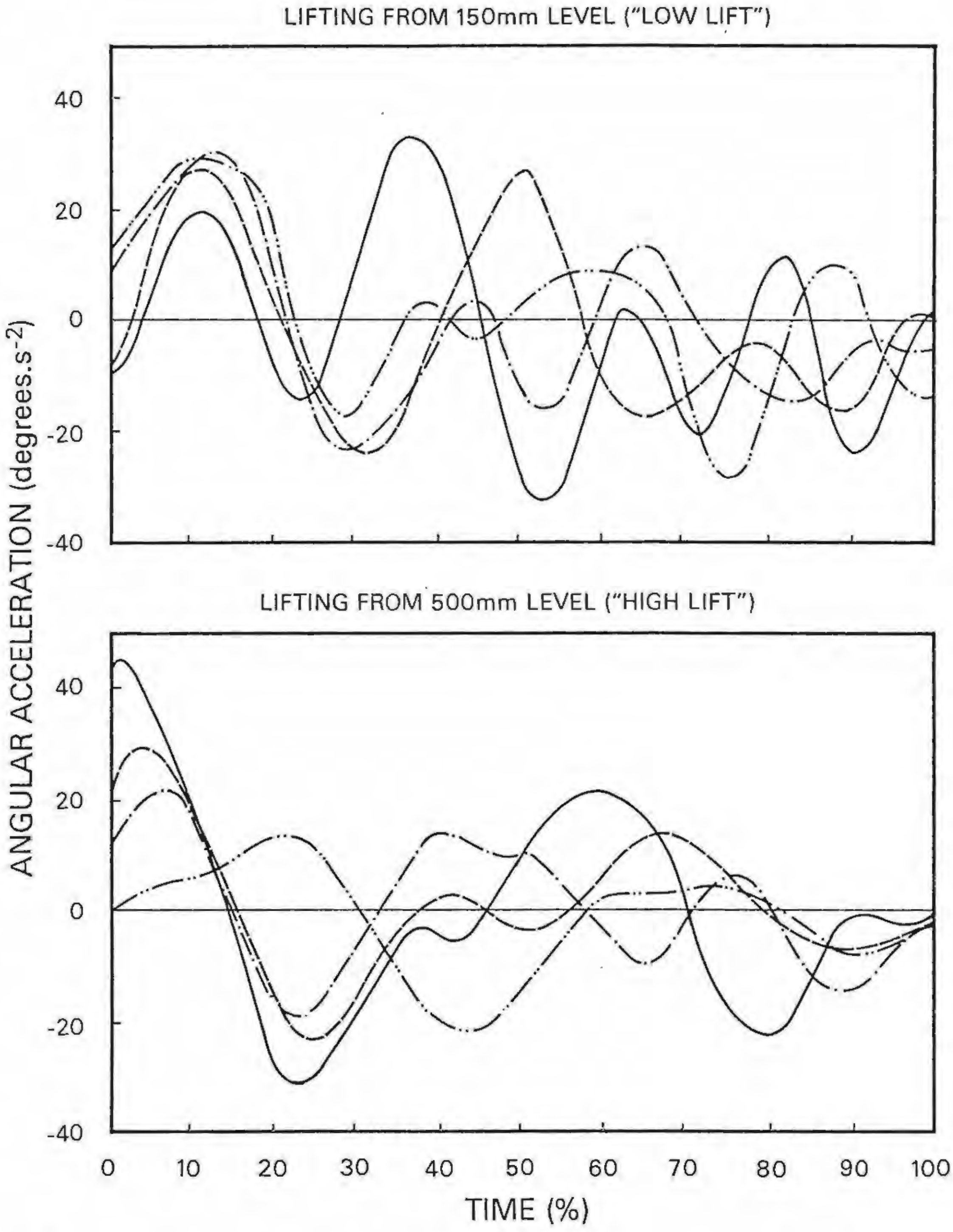

Figure 34 **Horizontal plane spinal angular acceleration traces** (Angular deviations from the sagittal plane indicated as follows:——— (90^0); — — (60^0); —·— (30^0); —··— (0^0).)

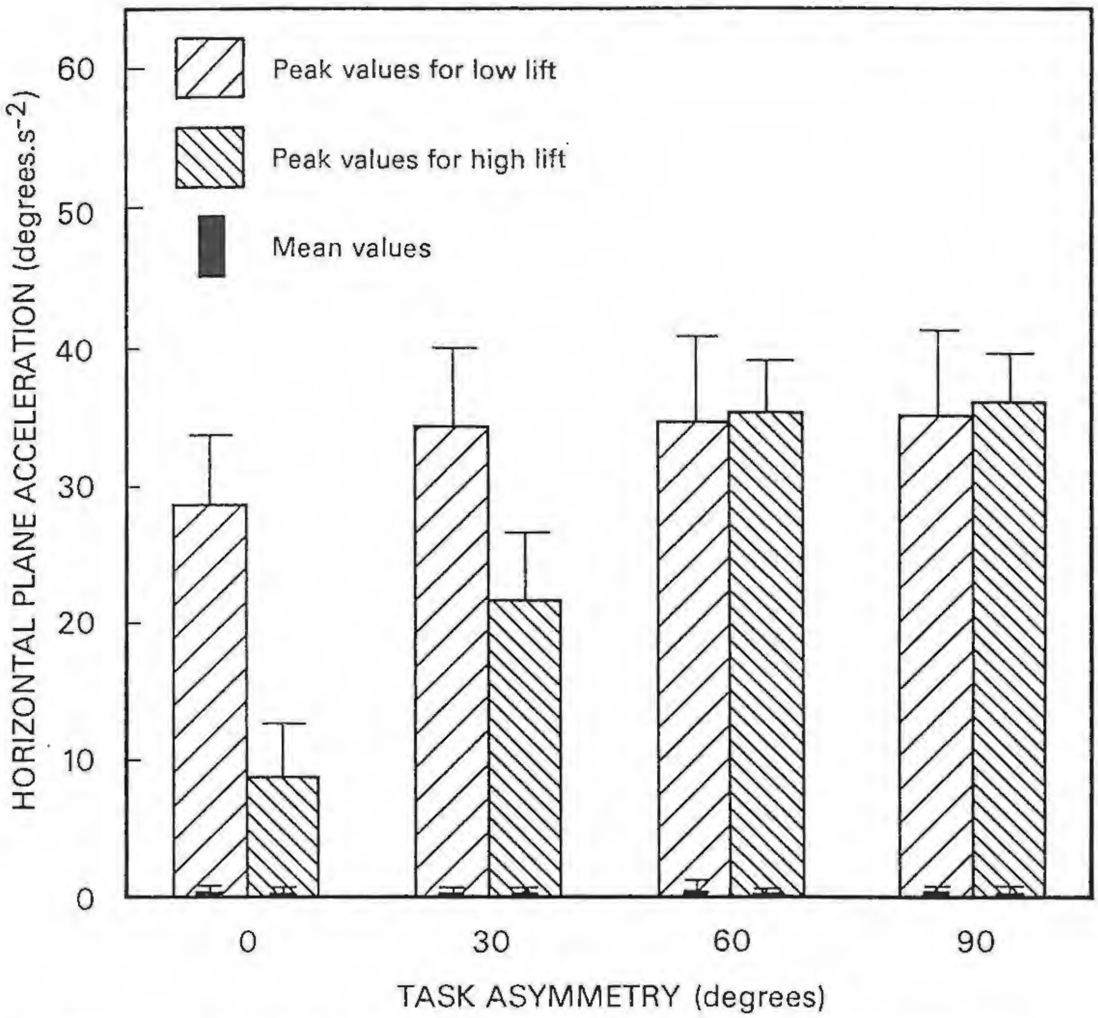

Figure 35 Horizontal plane maximum and average spinal acceleration

Discussion : kinematics of the human spine

Generally, the trunk motion factors increased as a function of increasing task asymmetry, except for the average acceleration in the sagittal plane. The rate of increase varied in the different planes and between the two vertical lift heights. In all asymmetrical conditions the motion factors showed a dramatic increase at the 500mm level compared to the increases at the 150mm level. The rate of increase in the horizontal and frontal planes was greater than that in the sagittal plane. Ferguson **et al.** (1992) reported the sagittal range of motion decreased, while horizontal range of motion increased as a function of task asymmetry. However, this trade-off in range of motion did not occur in the present study.

According to Ferguson et **al.**(1992), sagittal plane range of motion decreased as a function of increasing task asymmetry, but an increase was observed in the present investigation. The difference may be due to the fact that in the former study subjects performed the lifting tasks using different lifting techniques than those adopted by the subjects in the present study, as evidenced by the fact that the sagittal range of motion (ranging from 55.9^0 to 64.5^0) for four task asymmetries at 500mm height level in the present study was markedly greater than that (ranging from 42^0 to 47^0) observed by Ferguson and his colleagues for seven lifts from a 457mm vertical height level.

The finding that the sagittal peak velocity and peak acceleration increased as a function of increasing task asymmetry in the present study was similar to that reported by Ferguson et **al.**(1992). However, the peak velocities and peak accelerations in the present study were much greater than those in the above study. Three load levels: 6.4kg, 12.7kg and 19.1kg, were involved in Ferguson's study. Two of those load levels were heavier than the only load level(10kg) involved in the present study. It is arguable that the heavy load may have slowed down the lifting speed. Moreover student subjects were used in that study, whereas in the present study subjects were selected from a working population which was familiar with manual materials handling and therefore likely to be more efficient in manual lifting. Consequently a faster lifting style occurred in the present study. The different techniques used constitutes another possible reason for the differing lifting speeds.

While Ferguson **et al.** (1992) reported that frontal plane trunk motion parameters changed only slightly due to task asymmetry, a dramatic change was observed in the present study. Apparently, the subjects in the two studies used different lifting techniques in accordance with the different asymmetric situations.

According to Ferguson **et al.** (1992), a much greater horizontal range of motion (from 5^0 to 20^0) was observed for lifts at 0^0, 30^0, 60^0, 90^0, 120^0, 150^0 and 180^0 angles initiated from a 457mm vertical height. The horizontal plane range of motion ranged from 1.5^0 to 5.5^0 over the range of asymmetric angles analysed in the present study at 500mm height. The horizontal peak velocity was much smaller in the present study than that reported by Ferguson **et al.** (1992). Peak and average acceleration in the present study also showed a smaller values than those in Ferguson's study.

Marras and Mirka (1989) found that trunk strength decreased by 8.5% of maximum for every 15^0 of the trunk rotational position around L_5/S_1. This principle was used to predict the decrease in trunk strength under the eight different conditions of the present study and data are presented in Table IX.

In the present study trunk rotation angle was measured in the sagittal plane and under three task conditions of asymmetry, all at two vertical levels. Therefore, the percentage decrease in maximum trunk strength can be calculated at each task asymmetry and vertical level based on the horizontal range of motion and the claimed ratio of 8.5% decrease per 15^0 according to Marras and Mirka.

Table IX Measured trunk rotation in the transverse plane and predicted percentage decrease of maximum trunk strength based on the horizontal plane spinal range of motion (after Marras and Mirka (1989))

Task Asymmetry	150mm Lift Height		500mm Lift Height	
	Transversal Range of Motion	Decrease in Trunk Strength	Transversal Range of Motion	Decrease in Trunk Strength
0^0	3.8^0	2.2%	1.5^0	0.8%
30^0	4.3^0	2.4%	3.5^0	2.0%
60^0	4.7^0	2.7%	4.6^0	2.6%
90^0	5.0^0	2.8%	5.5^0	3.1%

The predicted percentage decrease in maximum trunk strength as a function of task asymmetry for each vertical level is presented in Table IX. This table indicates that the magnitude of force reduction caused by horizontal rotation is relatively small. Previously, Garg and Badger (1986) demonstrated that in three asymmetrical lifting postures the maximum acceptable weights reduced by 7—27% with respect to symmetric lifting, whereas the maximal voluntary isometric strength was reduced by 12—31%. Mital and Fard (1986) reported that the maximum acceptable weight decreased 8.5% at a 90^0 deviation from the sagittal plane.

However, Vink **et al.** (1992) found that the reduction in muscle force in asymmetric postures was dependent not only on horizontal rotation, but also on lateral and sagittal flexion. They proposed a decrease of 30% in trunk strength due to 30^0 lateral flexion. The results of the present study revealed that the subjects' trunk rotated in all three planes when performing the lifting tasks. Therefore, taking the effects of sagittal and frontal flexion into account, the total reduction in muscle force would probably become a significant quantity.

Marras and Mirka (1989) also reported that maximum trunk strength decreased approximately 0.33% for every degree-per-second increase in sagittal constant velocity. It could therefore be argued that the sagittal peak velocity may be used to determine the percentage of decrease in maximal trunk strength. Table X shows the predicted percentage decrease of maximal trunk strength under various task asymmetry conditions and vertical lift heights. This table indicates that dynamic exertion reduces trunk strength between 27% and 40% under the conditions imposed in the present study. This corresponds to the results of Vink **et al.** (1992) in that up to 40% reduction in muscle force was recorded when asymmetric postures were required to execute the lifting task.

Table X **Measured peak velocity in the sagittal plane and predicted percentage decrease of maximum trunk strength based on the sagittal plane peak spinal velocity** (after Marras and Mirka (1989))

Task Asymmetry	150mm Height		500mm Height	
	Peak Velocity(deg.s^{-1})	Decrease in Trunk Strength	Peak Velocity(deg.s^{-1})	Decrease in Trunk Strength
0^0	104.2	34.4%	82.1	27.1%
30^0	112.8	37.2%	83.5	27.6%
60^0	114.5	37.8%	97.7	32.2%
90^0	120.9	39.9%	113.7	37.5%

From Tables IX and X it is evident that predictions based on the spinal range of motion (ie. horizontal plane range of motion) underestimated the percentage decrease in maximum trunk strength caused by asymmetrical lifting; but spinal dynamic motion reduced back muscle force enormously. This may indicate that any fast and/or jerky lifting could bring about a tremendous hazard to the human back under both symmetric and asymmetric conditions. Therefore spinal dynamic variable should be recognized as a major risk factor in lifting analysis.

Based on the data of the present study, multiple regression models were developed for range of motion, peak and average velocity and acceleration in sagittal, frontal and horizontal planes (Table XI). The high r^2 values indicate that these parameter models should have good predictability.

Table XI Multiple regression models for trunk motion factors

Models	r^2
$ROM_{(S)} = 79.518571 - 0.043857 \cdot LEVEL + 0.061333 \cdot ASYM$	0.9684
$ROM_{(F)} = 7.900357 - 0.004786 \cdot LEVEL + 0.104833 \cdot ASYM$	0.9477
$ROM_{(H)} = 0.003942 \cdot LEVEL + 0.050627 \cdot ASYM$	0.8585
$V_{(S)} = 109.118571 - 0.053857 \cdot LEVEL + 0.269 \cdot ASYM$	0.8614
$V_{(S)mean} = 0.037788 \cdot LEVEL + 0.263125 \cdot ASYM$	0.7573
$V_{(F)} = 0.017178 \cdot LEVEL + 0.051428 \cdot ASYM$	0.8260
$V_{(F)mean} = 0.00159 \cdot LEVEL + 0.051428 \cdot ASYM$	0.8998
$V_{(H)} = 0.006071 \cdot LEVEL + 0.098958 \cdot ASYM$	0.8206
$V_{(H)mean} = 0.001063 \cdot LEVEL + 0.027444 \cdot ASYM$	0.8282
$A_{(S)} = 0.469799 \cdot LEVEL + 4.651526 \cdot ASYM$	0.7950
$A_{(S)mean} = 0.01459 \cdot LEVEL - 0.000238 \cdot ASYM$	0.8632
$A_{(F)} = 0.07673 \cdot LEVEL + 0.841494 \cdot ASYM$	0.8271
$A_{(F)mean} = 0.002147 \cdot LEVEL + 0.037224 \cdot ASYM$	0.9517
$A_{(H)} = 0.025329 \cdot LEVEL + 0.371805 \cdot ASYM$	0.8146
$A_{(H)mean} = 0.000161 \cdot LEVEL + 0.006275 \cdot ASYM$	0.8503

where:

LEVEL = initial vertical lift height(mm);

ASYM = asymmetric lift angle from sagittal plane(degree).

3.2.2 Kinetic Characteristics of Asymmetrical Lifting

Tables XII, XIII and XIV summarize the kinematic and kinetic measures of asymmetrical lifting in the present study. The variable descriptions are presented in Tables IV. The three-dimensional co-ordinate system established in the study is shown in Figure 17.

The duration of lifting (T_{max}) varied with subject, task asymmetry and vertical lift height. The lifting cycle increased in duration slightly with an increase in task asymmetry (Table XII), and was significantly different at the two vertical heights. The instantaneous measures of time on the Z-axis ($T_{V(z)}$, $T_{A(z)}$ and $T_{F(z)}$) showed no significant difference across the task asymmetries for both vertical lift heights. The time taken to attain peak velocity on the X-axis ($T_{V(x)}$) was significantly shorter at 90^0 task asymmetry than at 0^0 and 60^0 for both vertical lift levels. The time taken to achieve maximum acceleration on the X-axis ($T_{A(x)}$) and on the Y-axis ($T_{A(y)}$) decreased as a function of increasing task asymmetry at both lift levels. The values of $T_{A(x)}$ at the two vertical levels were significantly different. $T_{A(x)}$ at 90^0 asymmetry was significantly smaller than that at 0^0 and 30^0. $T_{A(y)}$ in sagittal symmetry was significantly greater than $T_{A(y)}$ in the three asymmetrical conditions at both vertical lift levels. Similarly, the time taken to attain peak force on the X-axis ($T_{F(x)}$) and on the Y-axis ($T_{F(y)}$) decreased with an increase in task asymmetry.

$T_{F(x)}$ was significantly different between the two vertical heights. $T_{F(x)}$ at 90^0 task asymmetry was significantly different from $T_{F(x)}$ at 0^0 and 30^0 . The time taken to achieve maximum force on the Y-axis at 90^0 was significantly different from that at 0^0, 30^0 and 60^0.

Table XII Temporal factors of asymmetrical lifting

Variables		(1) 150mm Vertical Height				(2) 500mm Vertical Height				Significance ($p<0.05$)
		A 0°	B 30°	C 60°	D 90°	E 0°	F 30°	G 60°	H 90°	
T_{max}	MEAN	1.33	1.34	1.35	1.39	1.2	1.21	1.25	1.26	1-2
	S.D.	0.15	0.17	0.15	0.22	0.24	0.16	0.15	0.31	
T_{Vmax}	MEAN	0.46	0.4	0.41	0.32	0.35	0.36	0.37	0.29	
	S.D.	0.07	0.12	0.15	0.13	0.08	0.06	0.05	0.36	
$T_{V(X)}$	MEAN	0.9	0.73	0.79	0.5	0.73	0.68	0.7	0.26	A-D;C-D
	S.D.	0.12	0.26	0.13	0.34	0.17	0.11	0.16	0.27	E-H;G-H
$T_{V(Y)}$	MEAN	0.88	0.63	0.64	0.52	0.44	0.53	0.51	0.26	1-2
	S.D.	0.25	0.24	0.36	0.38	0.25	0.14	0.16	0.27	
$T_{V(Z)}$	MEAN	0.45	0.44	0.4	0.29	0.35	0.34	0.33	0.34	
	S.D.	0.07	0.2	0.15	0.12	0.09	0.05	0.04	0.43	
T_{Amax}	MEAN	0.46	0.49	0.38	0.19	0.41	0.43	0.23	0.29	A-D;B-D
	S.D.	0.2	0.22	0.23	0.16	0.15	0.23	0.13	0.36	E-H;F-H
$T_{A(X)}$	MEAN	0.64	0.55	0.49	0.26	0.39	0.41	0.36	0.3	1-2;A-D;8-D
	S.D.	0.2	0.14	0.18	0.27	0.15	0.09	0.13	0.31	E-H;F-H
$T_{A(Y)}$	MEAN	0.65	0.38	0.32	0.24	0.44	0.29	0.2	0.33	A-B;A-C;A-D
	S.D.	0.2	0.23	0.23	0.18	0.24	0.17	0.14	0.29	E-F;E-G;E-H
$T_{A(Z)}$	MEAN	0.27	0.25	0.24	0.17	0.2	0.18	0.18	0.3	
	S.D.	0.1	0.11	0.14	0.09	0.07	0.03	0.03	0.27	
T_{Fmax}	MEAN	0.27	0.26	0.24	0.16	0.2	0.19	0.18	0.26	
	S.D.	0.1	0.11	0.14	0.09	0.07	0.04	0.03	0.32	
$T_{F(X)}$	MEAN	0.64	0.55	0.49	0.26	0.43	0.41	0.37	0.34	1-2;A-D;8-D
	S.D.	0.2	0.14	0.18	0.27	0.13	0.09	0.13	0.27	E-H;F-H
$T_{F(Y)}$	MEAN	0.65	0.38	0.32	0.24	0.45	0.29	0.2	0.28	A-B;A-C;A-D
	S.D.	0.21	0.23	0.23	0.18	0.25	0.17	0.14	0.36	E-F;E-G;E-H
$T_{F(Z)}$	MEAN	0.27	0.26	0.24	0.17	0.2	0.18	0.18	0.26	
	S.D.	0.1	0.11	0.14	0.09	0.07	0.03	0.03	0.27	
T_{Pmax}	MEAN	0.35	0.33	0.31	0.22	0.29	0.28	0.28	0.33	
	S.D.	0.07	0.09	0.14	0.12	0.08	0.05	0.04	0.28	

Table XIII Kinematic factors of asymmetrical lifting

Variables		(1) 150mm Vertical Height				(2) 500mm Vertical Height				Significance ($p<0.05$)
		A 0°	B 30°	C 60°	D 90°	E 0°	F 30°	G 60°	H 90°	
V_{max}	MEAN	1.98	2.23	3.06	3.13	1.68	1.69	1.75	2.11	1-2
	S.D.	0.32	0.38	0.32	0.43	0.25	0.3	0.32	0.44	
$V_{(X)}$	MEAN	1.0	1.03	1.1	1.23	0.99	1.06	1.13	1.29	A-C;A-D;B-D;C-D
	S.D.	0.12	0.1	0.16	0.2	0.11	0.15	0.1	0.16	E-G;E-H;F-H;G-H
$V_{(Y)}$	MEAN	0.18	0.49	0.65	0.84	0.13	0.38	0.67	0.8	A-B;A-C;A-D;B-C;B-D
	S.D.	0.08	0.15	0.26	0.44	0.09	0.11	0.13	0.14	E-F;E-G;E-H;F-G;F-H
$V_{(Z)}$	MEAN	1.96	2.2	2.48	2.56	1.65	1.64	1.6	1.54	1-2
	S.D.	0.33	0.36	0.8	0.75	0.25	0.31	0.32	0.36	
V_{mean}	MEAN	0.8	0.93	0.98	1.09	0.8	0.83	0.88	0.97	
	S.D.	0.27	0.11	0.24	0.17	0.1	0.12	0.1	0.18	
$V_{(X)mean}$	MEAN	0.3	0.33	0.44	0.61	0.3	0.38	0.47	0.65	A-C;A-D;B-C;B-D;C-D
	S.D.	0.04	0.05	0.07	0.13	0.08	0.08	0.07	0.14	E-G;E-H;F-G;F-H;G-H
$V_{(Y)mean}$	MEAN	0.02	0.14	0.28	0.31	0.03	0.16	0.29	0.32	A-B;A-C;A-D;B-C;B-D
	S.D.	0.02	0.03	0.06	0.07	0.05	0.03	0.04	0.06	E-F;E-G;E-H;F-G;F-H
$V_{(Z)mean}$	MEAN	0.52	0.54	0.56	0.57	0.45	0.4	0.39	0.38	1-2
	S.D.	0.06	0.07	0.15	0.09	0.11	0.05	0.04	0.07	
A_{max}	MEAN	8.38	9.97	11.07	12.93	7.76	7.6	8.43	9.48	1-2;A-D
	S.D.	2.77	2.47	4.0	5.65	2.0	1.91	1.59	3.14	E-H
$A_{(X)}$	MEAN	3.68	3.41	3.26	5.2	4.06	4.21	3.51	4.83	A-D;B-D;C-D
	S.D.	1.6	0.65	0.72	1.86	0.97	0.92	0.89	1.33	E-H;F-H;G-H
$A_{(Y)}$	MEAN	1.25	2.12	2.8	3.92	0.8	1.53	2.47	3.2	1-2;A-C;A-D;B-D;C-D
	S.D.	0.54	0.88	1.28	2.13	0.47	0.62	0.84	1.33	E-G;E-H;F-H;G-H
$A_{(Z)}$	MEAN	7.03	9.09	10.28	11.24	7.07	6.95	6.86	7.56	1-2
	S.D.	2.03	2.83	4.07	5.26	1.54	1.71	1.49	2.9	
A_{mean}	MEAN	4.05	4.31	4.61	4.64	4.18	4.23	4.08	4.11	
	S.D.	1.11	1.0	0.93	1.72	1.0	1.04	0.91	1.1	
$A_{(X)mean}$	MEAN	0.04	0.04	0.03	0.15	0.05	0.03	0.02	0.07	A-D;B-D;C-D
	S.D.	0.02	0.04	0.02	0.15	0.03	0.02	0.02	0.07	E-H;F-H;G-H
$A_{(Y)mean}$	MEAN	0.01	0.03	0.05	0.12	0.01	0.01	0.02	0.05	1-2;A-D;B-D;C-D
	S.D.	0.01	0.04	0.06	0.14	0.01	0.06	0.02	0.07	E-H;F-H;G-H
$A_{(Z)mean}$	MEAN	0.04	0.11	0.17	0.39	0.06	0.01	0.06	0.18	1-2;A-D;B-D;C-D
	S.D.	0.02	0.14	0.25	0.43	0.03	0.03	0.01	0.2	E-H;F-H;G-H

Table XIV Kinetic factors of asymmetrical lifting

Variables		(1) 150mm Vertical Height				(1) 500mm Vertical Height				Significance ($p < 0.05$)
		A 0°	B 30°	C 60°	D 90°	E 0°	F 30°	G 60°	H 90°	
F_{max}	MEAN	170.3	182.5	202.7	209.1	169.4	168.2	173.2	182.5	1-2
	S.D.	12.7	16.3	17.41	17.18	11.6	11.11	15.7	16.2	
$F_{(X)}$	MEAN	36.82	35.0	32.56	52.01	40.6	42.1	35.14	48.64	A-D;B-D;C-D
	S.D.	16.04	5.67	7.21	18.64	9.73	9.24	8.92	13.14	E-H;F-H;G-H
$F_{(Y)}$	MEAN	12.48	21.18	28.0	39.78	8.91	15.26	24.73	32.03	1-2;A-C;A-D;B-D;C-D
	S.D.	2.13	2.53	3.62	6.2	1.67	2.41	2.42	4.21	E-G;E-H;F-H;G-H
$F_{(Z)}$	MEAN	168.5	173.4	200.9	210.5	168.8	166.7	166.7	193.7	1-2
	S.D.	20.29	41.44	40.73	52.64	15.44	17.7	14.92	28.98	
F_{mean}	MEAN	102.4	103.0	103.2	103.2	101.9	102.1	101.5	101.3	
	S.D.	4.82	2.85	3.76	6.23	2.35	2.69	1.41	2.4	
$F_{(X)mean}$	MEAN	0.36	0.42	0.32	1.45	0.49	0.3	0.2	0.7	A-D;8-D;C-D
	S.D.	0.24	0.43	0.24	1.52	0.27	0.24	0.17	0.72	E-H;F-D;G-H
$F_{(Y)mean}$	MEAN	0.08	0.33	0.44	1.23	0.11	0.09	0.25	0.56	1-2;A-D;B-D;C-D
	S.D.	0.08	0.34	0.61	1.64	0.12	0.08	0.17	0.67	E-H;F-H;G-H
$F_{(Z)mean}$	MEAN	97.66	98.13	96.42	95.29	97.44	97.09	97.54	97.03	
	S.D.	0.2	2.94	2.54	5.22	0.27	1.62	0.15	2.44	
P_{max}	MEAN	255.3	342.6	475.2	516.0	221.2	219.3	221.4	292.6	1-2
	S.D.	61.93	205.5	448.9	342.6	45.5	57.34	58.39	97.59	
P_{mean}	MEAN	86.44	91.72	127.7	114.2	76.26	77.81	82.19	91.33	1-2
	S.D.	14.34	10.63	109.0	35.31	9.96	12.56	8.47	15.43	
W_k	MEAN	119.1	123.5	163.5	172.5	91.79	93.58	102.2	112.1	1-2
	S.D.	8.67	17.4	13.12	22.83	18.26	5.33	5.4	6.37	

Velocity

The velocity of the lift constitutes an important and complicated component of any lifting task, as the timing of movement significantly affects acceleration and therefore force. In the present study considerable variation in velocity was observed throughout the lift cycle.

Figure 36 shows the velocity traces on X-, Y- and Z-axes separately. The velocities on the X- and Z-axes at the two vertical levels followed very similar patterns. The velocities on the X-axis attained a peak around 60% of lifting cycle in the case of both lower and higher lifts, while the velocity on the Z-axis achieved a peak about 30% of lifting cycle.

Velocity on the Y-axis indicated a different pattern for the lifts at different levels of task asymmetry and vertical height. The velocity on the Y-axis fluctuated around the zero line for the sagittal lift at both vertical heights. The range of variation was greater at the lower level (150mm) than at the higher level (500mm). This supports the observation that, as lower lifts are performed greater asymmetry of movement is involved, as stated previously in this chapter. Figure 36 indicates that less movement on the Y-axis occurred when the lift was performed closer to the sagittal plane. The velocity trace deviated from the zero line as task asymmetry increased from 0^0 to 90^0. This corresponds to the earlier discussion of human trunk movement in the horizontal plane. It further corroborates the observation that the movement on the Y-axis is basically activated by shoulder swinging and trunk twisting (Li, 1993). According to the results of this study, movements in the vertical direction (Z-axis) dominated the whole range of lifting, irrespective of task asymmetry and vertical lift height.

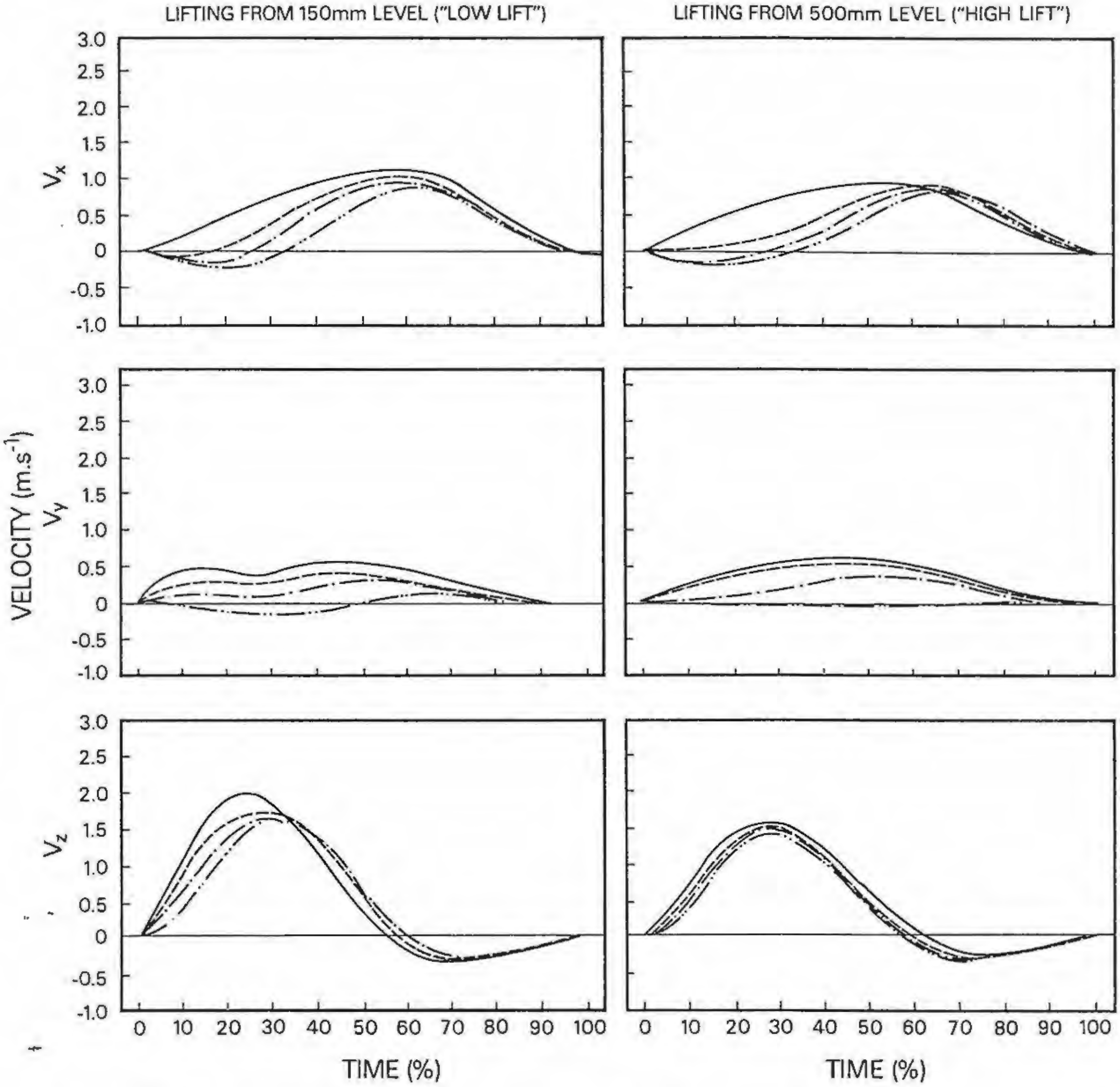

Figure 36 **Velocity traces on x-, y- and z-axes** (Angular deviations from the sagittal plane indicated as follows: ——— (90^0); — — (60^0); —·— (30^0); —··— (0^0).)

Figure 37 depicts the maximum and average velocities on X-, Y- and Z-axes, against task asymmetry. The maximum and average velocities on the X-axis increased as a function of increasing task asymmetry for lifts at both vertical heights. The maximum velocities on the X-axis at 90^0 task asymmetry were significantly greater than those at 0^0, 30^0 and 60^0 task asymmetries for the lifts at both vertical heights. The difference in maximum velocity on the X-axis between sagittal plane lifting and 60^0 twist lifting was statistically significant. The average velocities on the X-axis in sagittal symmetry and in 30^0 task asymmetry were significantly smaller than those at 60^0 and 90^0 task asymmetries. Similarly, the maximum and average velocities on the Y-axis increased as a function of increasing task asymmetry for the lifts at both vertical levels. The maximum and average velocities on the Y-axis in the sagittal lift and at 30^0 of twist were significantly smaller than those at 60^0 and 90^0 for both lift heights. The difference in maximum average velocity on the Y-axis between 0^0 and 30^0 was also statistically significant, but the difference in maximum and average velocity on X- and Y-axes between the two vertical heights was not. However, the maximum and average velocities on the Z-axis increased at the lower level and slightly decreased at the higher level while task asymmetry increased from 0^0 to 90^0. This may indicate that a threshold of change of velocity on the Z-axis is reached between 150mm and 500mm vertical lift heights with increasing task asymmetry. The velocity on the Z-axis was significantly different between the two vertical heights. No significant difference in velocity on the Z-axis was found between the lifts at various levels of task asymmetry.

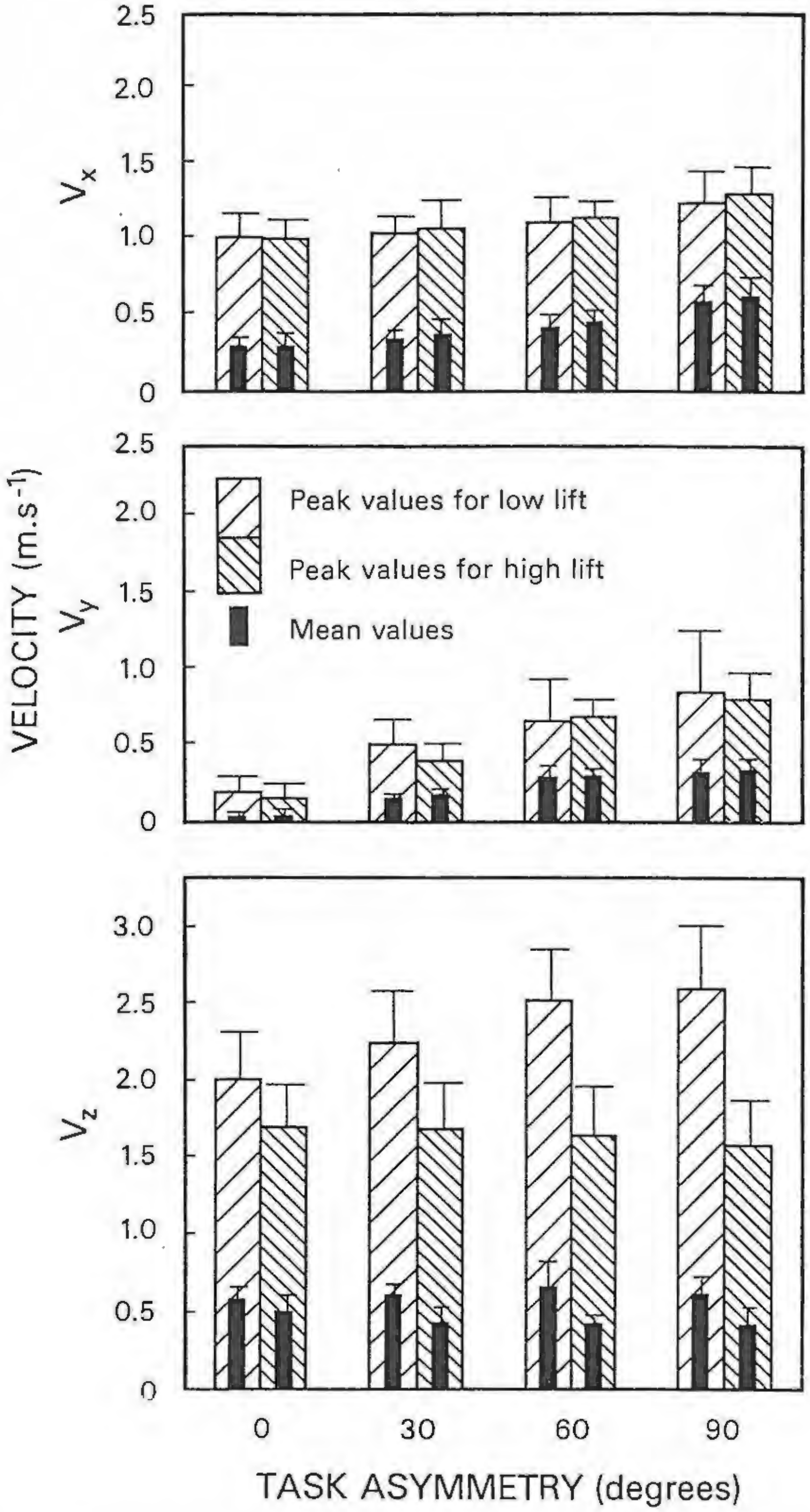

Figure 37 **Maximum and average velocity on x-, y- and z-axes**

Maximum and average velocities on X- and Y-axes increased at almost the same rate for the lifts at both vertical heights. However, the velocity on the Y-axis showed a more reactive response to asymmetry. Due to the very low original values, dramatic increase rates of 441% and 1212% were observed for the maximum and the average velocity respectively on the Y-axis when the task asymmetry increased from 0^0 to 90^0. Velocity on the X-axis increased by 27% and 110% for maximum and average respectively. At the 150mm vertical height (lower level), the maximum and average velocities on the Z-axis increased by 31% and 10%, respectively. The maximum and average velocities on the Z-axis decreased by 7% and 16% respectively at the 500mm vertical height (higher level).

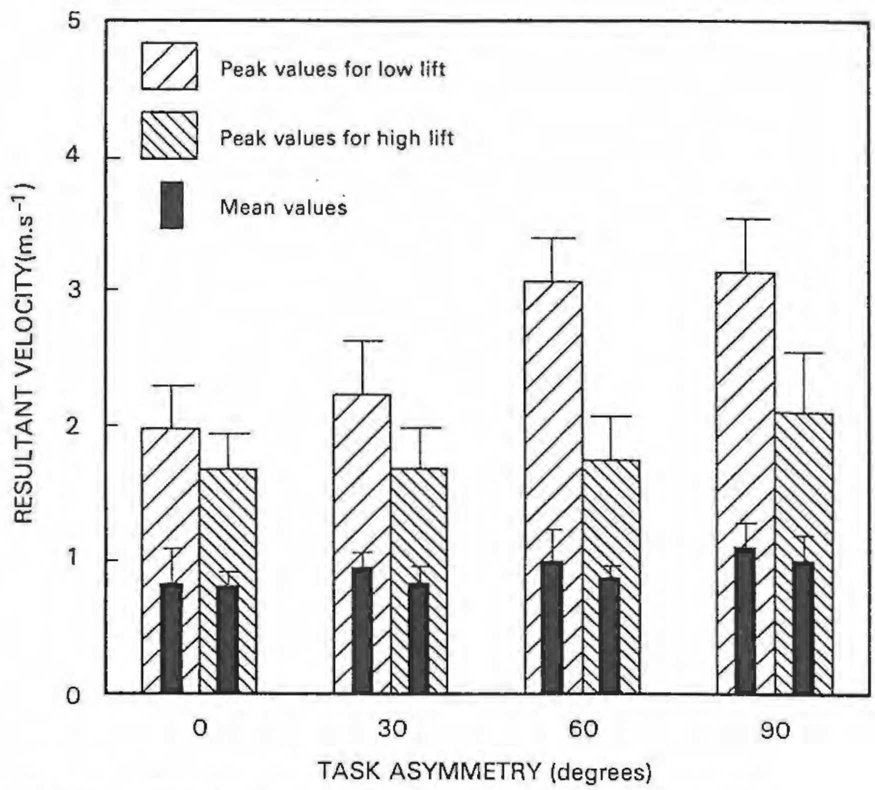

Figure 38 Maximum and average resultant velocity

Generally, resultant velocity increased along with an increase in task asymmetry (Figure 38), the rate of increase being greater at the lower level than at the higher level for both peak and average velocities. The maximum and average resultant velocities increased by 58% and 36% at the lower level, and 26% and 21% at the higher level.

Force/acceleration

Force/acceleration on X- and Z-axes exhibited similar sinus-curve patterns, while force/acceleration traces on Y-axis deviated to a marked degree for different lifts at various angle of task asymmetry (Figure 39).

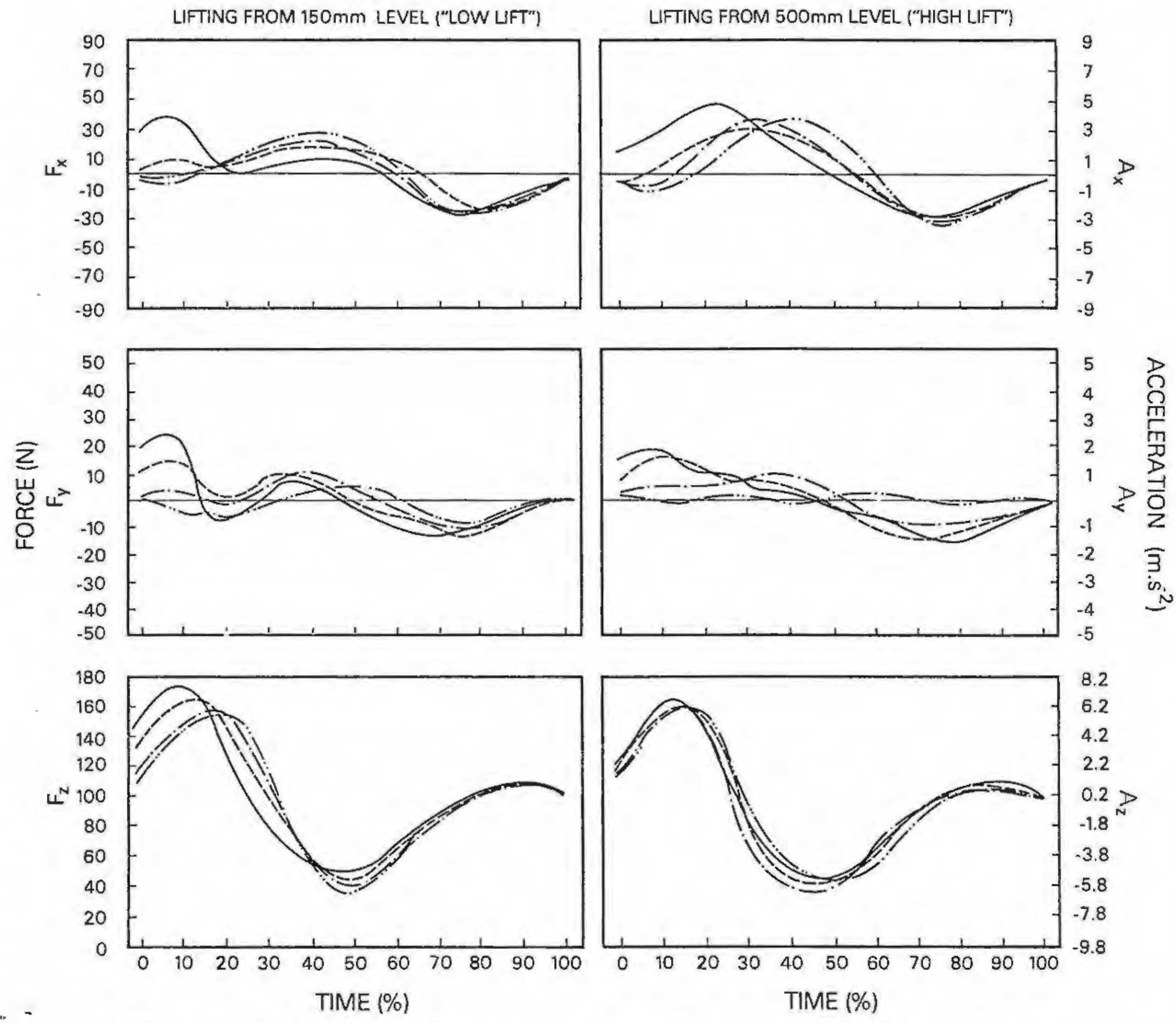

Figure 39 **Force/acceleration trace on x-, y- and z-axes** (Angular deviations from the sagittal plane indicated as follows: ——— (90°); — — (60°); —·— (30°); —··— (0°).)

At 0° the force/acceleration on the Y-axis showed a weak pulsation. This pulsation increased with an increase in task asymmetry at both vertical lift heights.

Based on the observation of this investigation, most force was applied in the vertical direction (Z-axis) for the accomplishment of lifting action through the full lifting cycle. The force applied on the Y-axis was the least. The force/acceleration on the X-axis achieved peak around 45% of lift cycle for the lifts at 0^0 and 30^0 at the lower level. Two peaks occurred for the lifts at 60^0 and 90^0, the first occurring shortly after lift-off(about 5% of lift cycle) and the second around 40% of lift cycle. At the higher level, the times taken to attain the peak on the X-axis were 20%, 30%, 35% and 40% of lift cycle for the lifts at 0^0, 30^0, 60^0 and 90^0 respectively. On the Y-axis, the force/acceleration smoothly oscillated around zero for the lifts at 0^0 and 30^0. However, a peak occurred shortly after lift-off (about 10% of lift cycle) for the lifts at 60^0 and 90^0. The time taken to achieve the peak on the Z-axis at the lower level (150mm) varied from 10 to 20% of lift cycle. The further away from the sagittal plane in which the lift was performed, the shorter was the delay to peak after lift-off. Force/acceleration attained a trough on the Z-axis around 50% of lift cycle for the lifts at the lower level. At the higher level (500mm), the peak on the Z-axis occurred around 15% of lift cycle and a trough was evident around 45% of lift cycle.

Maximum and average force/acceleration showed different tendencies on the X-, Y- and Z-axes respectively (Figure 40). The maximum and average force/ acceleration on the X-axis displayed a slight decrease from 0^0 to 60^0, and a sudden increase after 60^0. Both maximal and average force/acceleration on the X- and Y-axes at 90^0 task asymmetry were significantly greater than those involved at 0^0, 30^0 and 60^0 of twist; and the difference between 0^0 and 60^0 on the Y-axis was also significant.

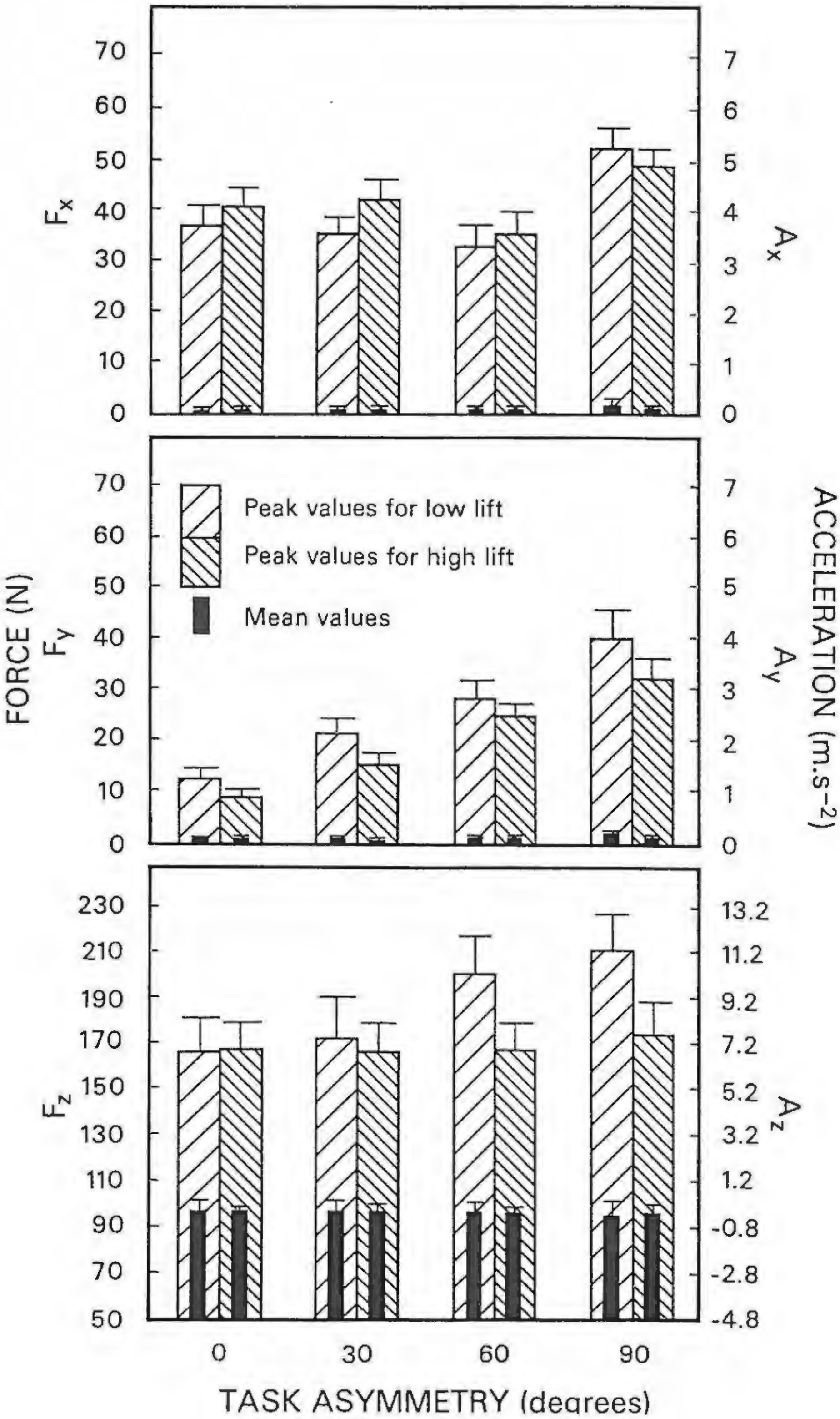

Figure 40 **Maximum and average force/acceleration**

Basically, maximum and average force/acceleration on the Y-axis increased as a function of increasing task asymmetry. The maximal force/acceleration on the Y-axis increased by 239% on average of the two vertical levels. A significant difference in force/acceleration was observed between the lower and higher lift levels. The maximum force/acceleration on the Z-axis increased at a much greater rate at the lower level than at the higher level. However, the average force/acceleration decreased with an increase in task asymmetry. The difference in maximum force/ acceleration on the Z-axis was significant between the two lift levels.

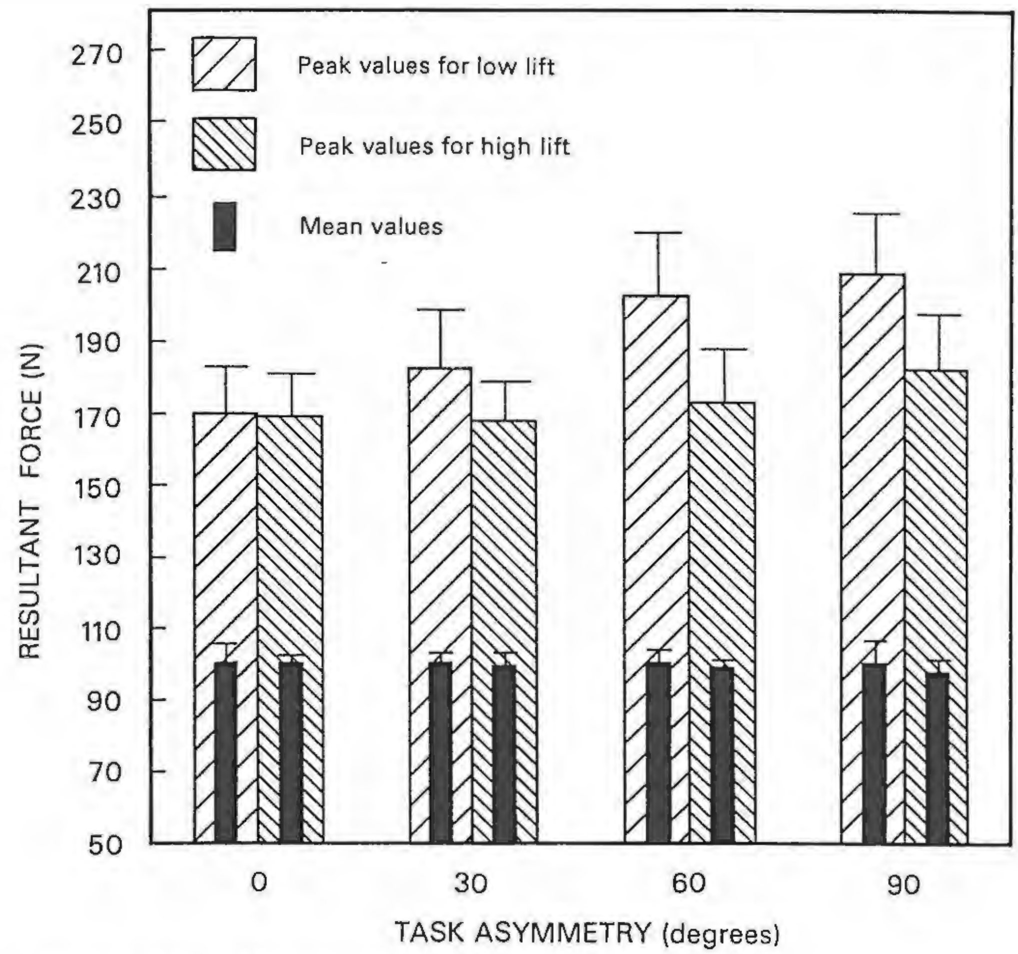

Figure 41 Maximum and average resultant force

The maximum resultant force increased with an increase in task asymmetry (Figure 41). However, the average resultant force increased at the lower level and decreased at the higher level. A significant difference was found between the two lift levels for the maximum resultant force.

Work and power

External work done by the subjects to accomplish the lifting task under the various conditions is displayed in Figure 42.

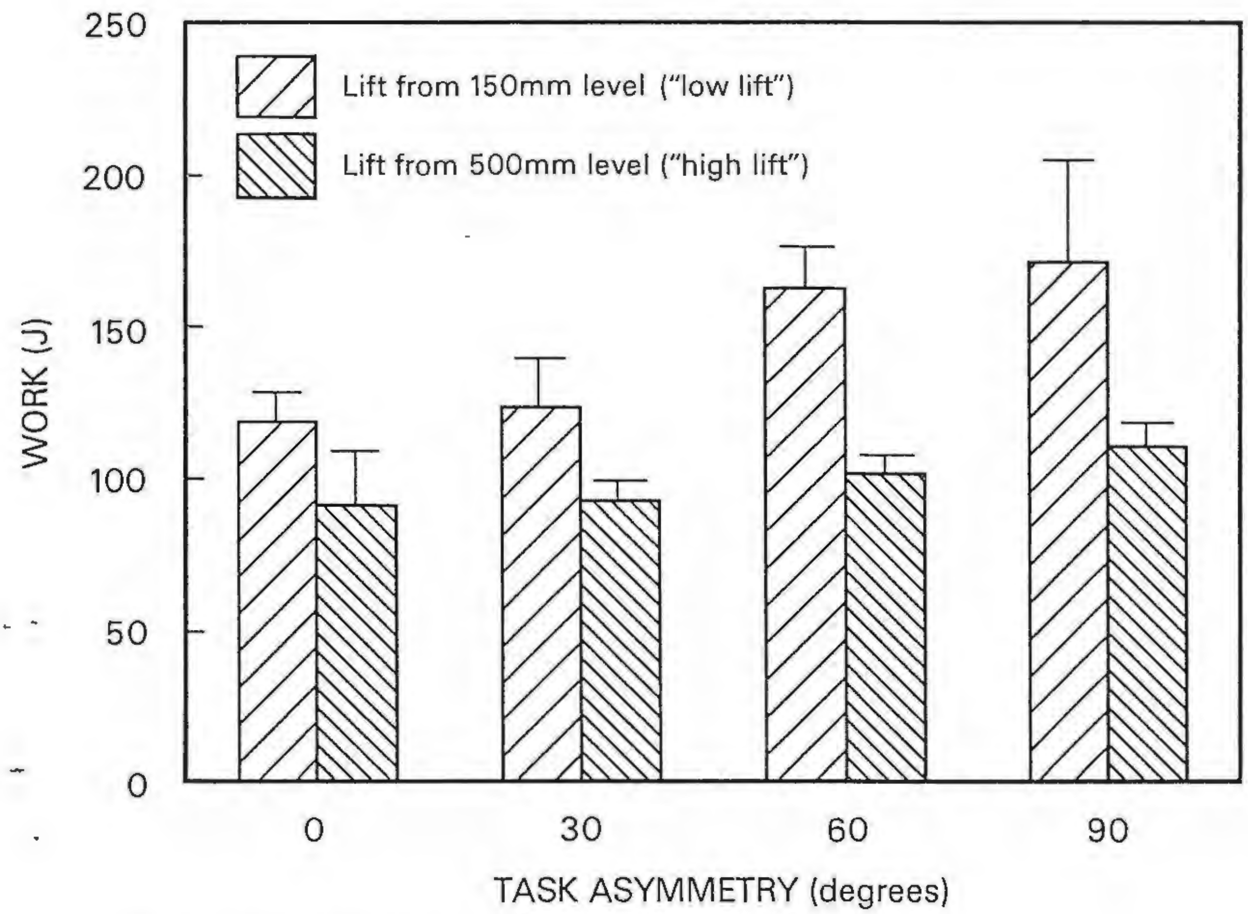

Figure 42 Maximum work

As expected the work increased with an increase in task asymmetry; and the increase was more dramatic at the lower level (150mm) than at the higher level (500mm). The difference between the two lift levels was statistically significant.

The instantaneous power output of the subjects doing the external work is shown in Figure 43. The power traces resembled each other in a curve pattern, irrespective of task asymmetry and vertical lift height. However, the maximum and average power increased along with an increase in task asymmetry and was significantly different between the two vertical lift levels (Table XIV). The power traces have two peaks with a single trough between. At the lower level, the time taken to attain the first peak varied from 10% to 30% of lift cycle for the lifts at various asymmetric angles. The further away from the sagittal plane in which the lift was performed, the closer was the peak to the point of lift-off. At the higher level, the power achieved in the first peak at almost the same time (around 25% of lift cycle) for the lifts with various degrees of twist. The trough and the second peak occurred at around 50% and 70% of the lift cycle respectively, regardless of task asymmetry and lift height.

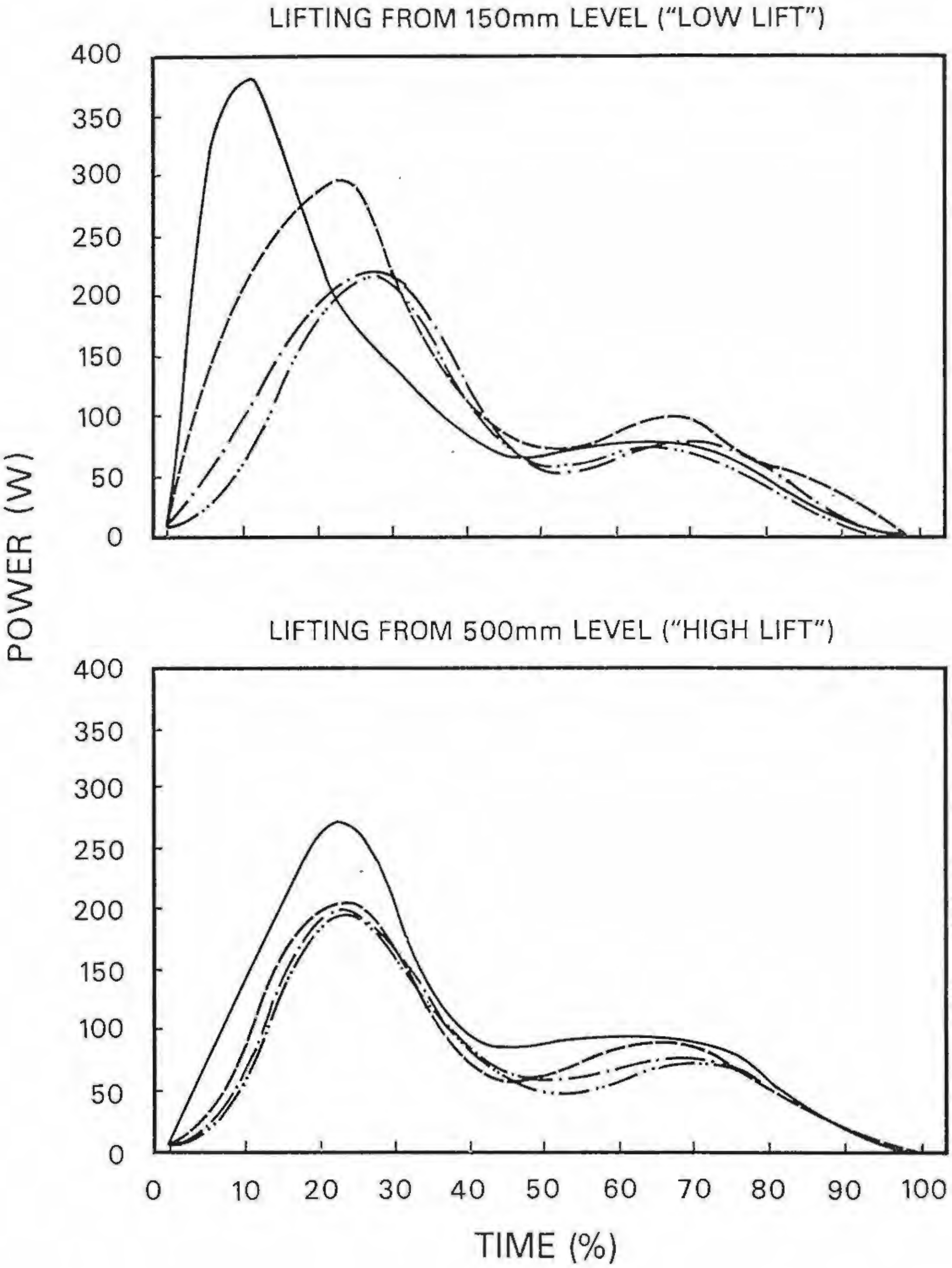

Figure 43 **Instantaneous power on x-, y- and z-axes** (Angular deviations from the sagittal plane indicated as follows: ——— (90^0); — — (60^0); —·— (30^0); —··— (0^0).)

Discussion: kinetics of asymmetrical lifting

The results of the present study revealed that the kinematic and kinetic measures of asymmetrical lifting increased generally with an increase in task asymmetry. The increase was more considerable on the Y-axis. However, motion factors of the object on the Z-axis (vertical direction) dominated the full range of the lift, irrespective of task asymmetry and lift height. This indicated that the lifting action was still the dominant component under the asymmetric conditions in the present study. It is clearly shown that measures in the vertical direction were significantly different between the lower (150mm) and the higher (500mm) levels, but that lift height had no significant effect on the variables on the X-axis, regardless of task asymmetry. However, the task asymmetry had a significant effect on the variables on the X-axis, but none on the variables on the Z-axis. Interestingly, both task asymmetry and lift height had a significant effect on the measures on the Y-axis, except in the case of velocity.

According to the results here presented, raising lift height may not only bring about a smaller trunk flexion angle (Li, 1992), but should also significantly reduce the magnitude of the dynamic factors of the object and thus lower the inertial effects on the lumbar spine moment in lifting. Li(1993) noted that the motion factors on the X- and Y-axes are the ones most likely related to shear forces on the lower back. Therefore the shear forces may possibly be cut down by increasing lift height and decreasing task asymmetry.

The lifting velocity recorded in the present study tended to be high, compared to results of some previous studies. The maximum peak velocity in the vertical direction (Z-axis) at sagittal symmetry was 1.41—2.39m.s^{-1} and 1.40—1.97m.s^{-1} for the lifts at the lower and the higher levels, respectively. Bush-Joseph **et al.**(1988) studied lifts of a 150N box from floor to a 1m shelf. The lift was restricted to the sagittal plane. These authors observed that the vertical peak lifting speeds were 0.8—0.9m.s^{-1}, 1.1—1.2m.s^{-1} and 1.7m.s^{-1} for slow, normal and fast lifts respectively. According to Leskinen **et al.**(1983), the vertical peak velocity of 1.4—1.6m.s^{-1} occurred when subjects lifted a 147N box from 100mm to knuckle height. However, Lindbeck and Arborelius (1991) reported a vertical lifting speed of 1.4—2.3m.s^{-1}, which is similar to the velocity observed in the present study. In their investigation, subjects were required to lift a box of 126N from floor to a height corresponding to 61% of the lifter's stature. In Gagnon and Smyth's study the lifts were executed from a height of 150mm to a height of 1800mm with two different loads (62.8N and 112.8N) in the sagittal plane, and the vertical peak lifting velocity ranged from 1.1m.s^{-1} to 2.2m.s^{-1} (Gagnon and Smyth, 1992).

The difference in the lifting velocity may be due to the fact that different load weights were lifted in the various studies and varying techniques were employed by the subjects to perform the lifting task. Furthermore, the results of the present study have indicated that the peak velocity increased with an increase in task asymmetry. Even in the sagittal plane, lifting action may be executed asymmetrically, as already discussed in this study. Consequently, a higher lifting speed is involved in asymmetrical lifting. However, a fast lifting speed may result in a high moment and force at the L_5/S_1 level (Bush-Joseph **et al.**, 1988; Jager and Luttmann, 1989 and Tsuang **et al.**, 1992).

It could be argued that three-dimensional movement is involved in the vast majority of manual lifting unless the lift is performed on a specially designed machine, such as Kroemer's incremental lifting machine (ILM) (Kroemer, 1983). The present study has demonstrated that movements in the horizontal plane (X- and Y-axes) are a part of lifting action, even in the case of lifts ostensibly in the sagittal plane. Table XV shows component velocities (V_X, V_Y and V_Z) as a proportion of the resultant velocity. The vertical proportion of the movement (V_Z) decreased as task asymmetry increased from 0^0 to 90^0, and the horizontal proportion increased on the Y-axis. This indicates that the movements are re-orientated in three-dimensional space in accordance with the change of asymmetric situation.

Table XV Peak component velocities expressed as a proportion of the peak resultant velocity

TASK	(1) 150mm Lift Height			(2) 500mm Lift Height		
ASYMMETRY	V_X	V_Y	V_Z	V_X	V_Y	V_Z
0^0	0.51	0.09	0.99	0.59	0.08	0.98
30^0	0.46	0.22	0.99	0.63	0.22	0.97
60^0	0.36	0.21	0.81	0.65	0.38	0.91
90^0	0.39	0.27	0.82	0.61	0.38	0.73

Grieve (1975) reported that resultant hand force for fast lifting was an average of 3.83 times the load for a 38N load, 2.58 times the load for a 137N load and 1.75 times the load for a 285N load. Danz and Ayoub (1992) directly measured the hand forces in the vertical and horizontal directions (2-D) during a manual lifting, using a strain gauge apparatus. However, 3-D force application is certainly involved in lifting action in the real world. In this study, the force applied on the object by the subject was resolved into three components and calculated using Newton's Laws of Motion. The quantitative information of 3-D hand force will facilitate a more realistic simulation of lifting.

Table XVI Peak forces expressed as a proportion of the load

TASK	(1) 150mm Lift Height				(2) 500mm Lift Height			
ASYMMETRY	F_X	F_Y	F_Z	F	F_X	F_Y	F_Z	F
0^0	0.37	0.13	1.72	1.74	0.41	0.09	1.72	1.73
30^0	0.36	0.22	1.77	1.86	0.43	0.16	1.70	1.71
60^0	0.33	0.29	2.05	2.07	0.36	0.25	1.70	1.77
90^0	0.53	0.41	2.14	2.13	0.50	0.33	1.97	1.86

As shown in Table XVI, the magnitude of the vertical peak force (F_Z) and the peak resultant force (F) recorded in this study always exceeded the weight of the load, regardless of task asymmetry and lift height. This is not difficult to understand because lifting the load would be impossible if the peak hand force merely reached the weight of the load. The peak resultant force in the present study showed a much smaller proportion than that reported by Grieve (1975) and Danz and Ayoub (1992). Referring to the previous discussions, subjects in this study used a fast lifting style, but this did not cause greater force application. This may demonstrate that the subject executed the lifting task smoothly and efficiently, and consequently less force was involved. In addition, the time taken to attain the peak in vertical and horizontal force was much longer than that reported previously (Grieve, 1975; Danz and Ayoub, 1992). This reaffirms that the subjects did lift smoothly in the present study. The total proportion of horizontal forces (F_X plus F_Y) was 0.5 times the load under the sagittal condition, and increased at almost the same rate from 0^0 to 90^0 task asymmetry for both lift heights. Danz and Ayoub

(1992) observed that horizontal hand force was equal to the magnitude of the load at fast lifting speed and half the magnitude of the load at normal speed. Corresponding to the present study, the total proportion of horizontal forces was approximately equal to the magnitude of the load at 90^0 task asymmetry and half the magnitude of the load at 0^0 task asymmetry.

Multiple regression analyses were conducted based on the results of the current study. The prediction equations of dynamic factors derived with the independent variables of lift height (mm) and task asymmetry (degree) are listed in Tables XVII and XVIII.

Table XVII   Multiple regression models for kinematic factors

Model	r^2
T_M = 1.37425-0.00035·LEVEL+0.000683·ASYM	0.9809
V_{max} = 2.517393-0.002264·LEVEL+0.009383·ASYM	0.7976
$V_{(X)}$ = 0.948464+0.000079·LEVEL+0.002883·ASYM	0.8972
$V_{(Y)}$ = 0.226286-0.000129·LEVEL+0.0074·ASYM	0.9684
$V_{(Z)}$ = 2.468536-0.001979·LEVEL+0.00285·ASYM	0.8121
V_{mean} = 0.873286-0.000229·LEVEL+0.002467·ASYM	0.8992
$V_{(X)mean}$ = 0.243643+0.000086·LEVEL+0.003633·ASYM	0.9130
$V_{(Y)mean}$ = 0.031393+0.000036·LEVEL+0.00335·ASYM	0.9233
$V_{(Z)mean}$ = 0.612321-0.000407·LEVEL-0.000083·ASYM	0.8658
A_{max} = 0.004857-0.006486·LEVEL+0.034567·ASYM	0.8562
$A_{(X)}$ = 0.006399·LEVEL-0.0313·ASYM	0.8550
$A_{(Y)}$ = 1.484179-0.001493·LEVEL+0.02805·ASYM	0.9905
$A_{(Z)}$ = 0.009144·LEVEL+0.084592·ASYM	0.7653
A_{mean} = 0.006721·LEVEL+0.030902·ASYM	0.8030
$A_{(X)mean}$ = 0.000015·LEVEL+0.000917·ASYM	0.6507
$A_{(Y)mean}$ = -0.000036·LEVEL+0.000988·ASYM	0.8170
$A_{(Z)mean}$ = -0.000105·LEVEL+0.003214·ASYM	0.7956

where:

LEVEL = initial vertical lift height(mm);

ASYM = asymmetric lift angle from sagittal plane(degree).

Table XVIII Multiple regression models for kinetic factors

Models	r^2
$F_{max} = 0.255423 \cdot LEVEL + 1.476868 \cdot ASYM$	0.7922
$F_{(X)} = 0.064089 \cdot LEVEL + 0.314884 \cdot ASYM$	0.8543
$F_{(Y)} = 14.99125 - 0.01465 \cdot LEVEL + 0.27925 \cdot ASYM$	0.9839
$F_{(Z)} = 0.257204 \cdot LEVEL + 1.458161 \cdot ASYM$	0.8009
$F_{mean} = 0.172514 \cdot LEVEL + 0.656862 \cdot ASYM$	0.8099
$F_{(X)mean} = 0.000154 \cdot LEVEL + 0.009061 \cdot ASYM$	0.6709
$F_{(Y)mean} = -0.000332 \cdot LEVEL + 0.010078 \cdot ASYM$	0.8308
$F_{(Z)mean} = 0.166591 \cdot LEVEL + 0.607613 \cdot ASYM$	0.8119
$P_{max} = 380.45425 - 0.45335 \cdot LEVEL + 1.885183 \cdot ASYM$	0.7816
$P_{mean} = 0.10761 \cdot LEVEL + 0.936002 \cdot ASYM$	0.7748
$W_k = 143.615357 - 0.127886 \cdot LEVEL + 0.4495 \cdot ASYM$	0.8551

where :
LEVEL = initial vertical lift height(mm);

ASYM = asymmetric lift angle from sagittal plane(degree).

CHAPTER 4: PHASE II:THE CONCEPTUAL STUDY

A THREE-DIMENSIONAL FORCE MODEL OF THE LOWER BACK

4.1 INTRODUCTION

Biomechanical modelling has been widely used for evaluating the forces and moments acting on the human body during manual materials handling. Basically these models assume that the human body is a simple mechanical structure amenable to the laws of mechanics in respect of analysis of forces and moments. A general review of biomechanical modelling in manual lifting was presented in Chapter 2.

In the view of Kromodihardjo and Mital (1986), the purpose of developing a biomechanical model for analysing manual lifting is to quantify the stresses acting upon the L_5/S_1 disc of the spinal column. These authors summarized two distinct approaches used for estimating spinal stress: static and dynamic analysis. The major difference between these approaches is that in dynamic analysis the effects of motion are taken into consideration while in static analysis they are ignored. In fact human body motion in dynamic analyses is interpolated from a series of static positions.

The modelling approach considers the human body as a system of links and connecting joints. The majority of the biomechanical models developed in the past have been restricted to the sagittal plane. These uni-planar models may assume the human body to be comprised of as few as four rigid segments linked by three hinge joints (Leskinen **et al.**, 1983; Bejjani **et al.**, 1984), or as many as eight segments with seven joints (Freivalds **et al.**, 1984). The trunk is often regarded as a simple one-link or two-link system. In Aspden's model the human spine was assumed to function as an arch rather than a cantilever (Aspden, 1988). In reality the spine consists of five lumbar and twelve thoracic vertebrae in the trunk region. These vertebrae rotate in diverse ways and to different degrees during manual lifting activities, resulting in each vertebra being stressed differently. Acknowledging the complexity of the human spine, Frigo (1990) considered it a four-link system in his three-dimensional model. Jager and Luttmann (1989), using a complex 19-segment model, investigated the different force values at each of the five intervertebral discs of the lumbar spine.

That lifting actions involve three-dimensional motion has been clearly demonstrated in the previous chapter of the present study. In order to accomplish lifting and to maintain balance under asymmetrical conditions, multiple muscle force applications are needed. Schultz and Andersson (1981) developed a three-dimensional lower back model which includes ten muscle forces. However, calculation of these muscle forces constitutes a statistically indeterminate problem as the number of unknown muscle forces is more than the number of equations of equilibrium. More recently Tracy (1990) reduced the 10-muscle model to a 6-muscle model, without loss of predictability.

Earlier models focused mainly on resolving compressive and shear forces at the L_5/S_1 junction. Two shear forces (lateral and anterior) were introduced in Schultz and Andersson's lower back model. In fact the human trunk always rotates a certain extent to right or left in the transverse plane during asymmetrical lifting. The relative rotational movement between upper trunk and pelvis causes a torsional moment in the lumbar spine which may, according to Farfan **et al.** (1970), be the main cause of wear-related disc degeneration, and in turn one of the main sources of lower back pain (Rowe, 1969; 1971).

Consequently, the main focus of the model proposed in the present study was to ascertain the three-dimensional activity within the lower back when performing manual lifting, and is offered in an attempt to show how torsional moments in the lumbar spine during asymmetrical lifting may be calculated.

4.2 BIOMECHANICAL MODEL OF THE LOWER BACK

The empirical study (Chapter 3) clearly demonstrates that three-dimensional movements of the human lumbar spine always accompany manual lifting; and that these spinal movements are always exacerbated under asymmetrical conditions. Moreover, dynamic motions of the object occurring outside the human body in three-dimensional space increased as a function of task asymmetry. In the empirical phase of this project, analysis of the spinal kinematics identified a dramatic increase in horizontal and frontal planes; the kinematics and kinetics of the object, however, fluctuated between vertical and horizontal while performing an asymmetrical lifting task (Tables XV and XVI). These changes indicated that the force application, both inside and outside the human body, is altered under asymmetrical lifting conditions. It may be further deduced that the musculoskeletal system is stressed differently due to this alteration.

Apparently, uni-planar models are not capable of examining external and internal force alteration under asymmetrical lifting conditions. Furthermore, although the human body is constructed in three dimensions and is largely bi-lateral, the left side of the body is not identical to the right side. Particularly the musculoskeletal system tends to asymmetry of geometric configuration and strength due to the fact that one side of the body is dominant over the other. This results in an asymmetrical force application by involved muscles during manual lifting, and further causes three-dimensional movements of the body segments. As reported in the empirical study, asymmetrical movements of the lumbar spine and of the object were observed, even when subjects were performing a symmetrical lifting task. It could thus be argued that anatomical asymmetry is a causative factor in the asymmetrical motions.

Uni-planar models consider only the erector spinae to be engaged in lifting actions; clearly, there are other muscles in the lumbar region which contribute to lifting as well. Kumar (1980) reported that the electromyographic activity of the external oblique muscles was consistently lower in sagittal lifts and higher in lateral and oblique lifts. The external obliques, according to Gallagher **et al.** (1994), were 31.5% more active in the asymmetrical conditions than when lifting symmetrically. The internal and external obliques have a moment arm about the spine that would make them well placed for development of axial rotation under asymmetrical lifting conditions. During trunk extension, according to Pope **et al.** (1986), a considerable amount of the muscle contraction is used to control the posture. The inter-abdominal pressure is another factor in the development of support for the spine (Troup, 1979). However, Schultz **et al.** (1982) found inter-abdominal pressure to be low and to have little effect on trunk mechanics.

Any simple body movement needs an integrated effort of various muscles. The muscles have to co-contract concentrically and eccentrically in combination, and to maintain whole-body balance during a dynamic movement. Under the dynamic equilibrium, forces generated by various muscles are not symmetrically distributed all the time. At one moment the forces on one side might become greater or lesser than on the other side and the equilibrium is temporarily upset. In order to restore the state of equilibrium, the muscles on the opposite side have accordingly to produce greater or lesser forces. The process of maintaining a dynamic equilibrium occurs almost instantaneously. The alteration of force distribution affects the motions of body segments. If the forces on one side of the movement are greater than on the other , the movement will be pulled away from its neutral (symmetrical) path. In the present study it was found that these deviations of movement occur in a see-saw fashion: e.g. when subjects were performing a symmetrical lifting task, the motions of the lumbar spine fluctuated to right and left in both frontal and horizontal planes (see Figures 27 and 32); and the movement of the object deviated to right and left in the horizontal plane as well (see Figure 36). The results of the empirical study also demonstrate that these deviations in movements of the lumbar spine and of the object were exacerbated under asymmetrical lifting conditions. Generally, asymmetrical movement is a feature of manual lifting and cannot be neglected. From the point of view of this rationale, analysis of asymmetrical movements and prediction of the stresses imposed by them requires the development of three-dimensional biomechanical models.

In the empirical study, the lumbar range of motion varied from 1.5^0 to 5.5^0 in the horizontal plane and from 5.1^0 to 16.5^0 in the frontal plane over the range of asymmetrical angles analyzed. Although some lifting tasks were performed within the sagittal plane, tangible motion of the lumbar spine was observed in both horizontal and frontal planes. Similarly, though the object was predominantly moved within the sagittal plane, movements were also recorded outside the sagittal plane for ostensibly uni-planar lifting. Based on the findings revealed in the empirical study, questions were raised against the validity of uni-planar lifting models developed in the past. These uni-planar models only consider the human spine and the load being lifted as moving in the sagittal plane, and therefore 2-D force analyses have been applied. However, cross-planar lifting involves three-dimensional movements and consequently requires 3-D force application modelling. Several such models have been proposed. Three-dimensional force distribution was included in the biomechanical models developed by Schultz and Andersson (1981) and Tracy (1990), but these regarded the lower back as a rigid body. In other words, the lower back was assumed not to rotate around its longitudinal axis in those models. Obviously, this assumption is inconsistent with the facts, certainly as revealed in the present study.

Hutson (1993) noted that the human spine is a system of articulated segments superimposed upon one another. The basic movements of the spine are flexion/extension in the sagittal plane, lateral bending in the coronal plane and rotation in the transverse plane. The lumbar spine has its greatest excursions in flexion and extension, with a limited range in rotation because of restriction of facet planes (White and Panjabi, 1978). Radin **et al.** (1979), however, found that side bending and axial rotation are always associated with any spinal movements. This phenomenon associated with spinal kinematics was also observed in the present empirical study. As reported previously, the lumbar motions in three reference planes are coupled with each other under various lifting conditions. Spinal rotation causes a torsional moment on the intervertebral discs. In general the disc serves to absorb loads and distribute the forces applied to the spine. Disc degeneration has been reported as the prime cause of the lower back pain (Rowe, 1969; Jayson, 1981; NIOSH, 1981; Hutson, 1993). According to Farfan et al. (1970), a small amount of twist will produce a relatively large loss of disc volume. Thus it may be easier to reduce disc volume by torsion than by axial compression. In addition, rotation creates both tension and shear in the annulus fibrosus of the intervertebral disc (White and Panjabi, 1978; Soderberg, 1986). Shear takes place in the horizontal plane around which the rotation is occurring. Tension develops in the fibres oriented in the direction of the rotation. White and Panjabi (1978) pointed out that the greatest stresses are expected in the peripheral segments of the annular fibres because motion is greatest in the fibres that are farthest from the centre of rotation. Although the spinal compression force has been widely recognized as a critical value in lifting task assessment, loading of the spine in a real lifting action is a combination of compression, shear and torsion. Nordin and Frankel (1989) found that clinical examination of fracture patterns indicates that most fractures are produced by a combination of several loading modes. Therefore, combined loading of the spine should be included in biomechanical analyses of manual lifting.

The musculoskeletal system of the human body is one of the most complicated structures. The force distributions among muscles inside the body are partially based on the geometrical configuration of these structures. The present study demonstrated that working postures are clearly changed under asymmetrical lifting conditions. This results in a transformation in spinal ranges of motion in all three planes. When the upper body twists to the right or left to perform an asymmetrical lifting task, muscles in the lumbar region are caused to re-orient their geometric configuration. Muscles such as erector spinae and rectus abdominis have to change their direction of action from vertical to oblique. However, some oblique muscles have to be disengaged in order to complete an asymmetrical lifting task; e.g. the internal oblique abdominis on the left side of the body and external oblique abdominis on the right side are expected to be inactive while performing a lateral lift from right to left and the left internal and right external oblique abdominis should not be involved for a lateral lifting from left to right. Rotational movement of the lower back in the transverse plane may cause the muscles to change direction, resulting in re-orientation of forces in the lumbar region.

The anatomical mechanism discussed above can be modelled as shown in Figure 44. There is a force **F** applying to the top edge of a cylinder at point **P**. The direction of the force is vertically downward and the line of application is the link between **P** and **Q**. Assume that an external force is applied to rotate the top (anticlockwise to an angle of θ) relative to the bottom of the cylinder (Figure 44b). The force (**F**) still acts at point **P** with the line of application connecting **P** and **Q**. The direction of the force (**F**) however is changed to run obliquely downwards at an angle of δ to the vertical. The angle δ may be called a "strain angle of twist".

Under this condition, the force (**F**) may be resolved into vertical and horizontal components. The magnitude of the vertical component force (**F_1**) and horizontal component force (**F_2**) can be calculated by the following equations:

$$F_1 = F \cdot Cos\delta$$

$$F_2 = F \cdot Sin\delta$$

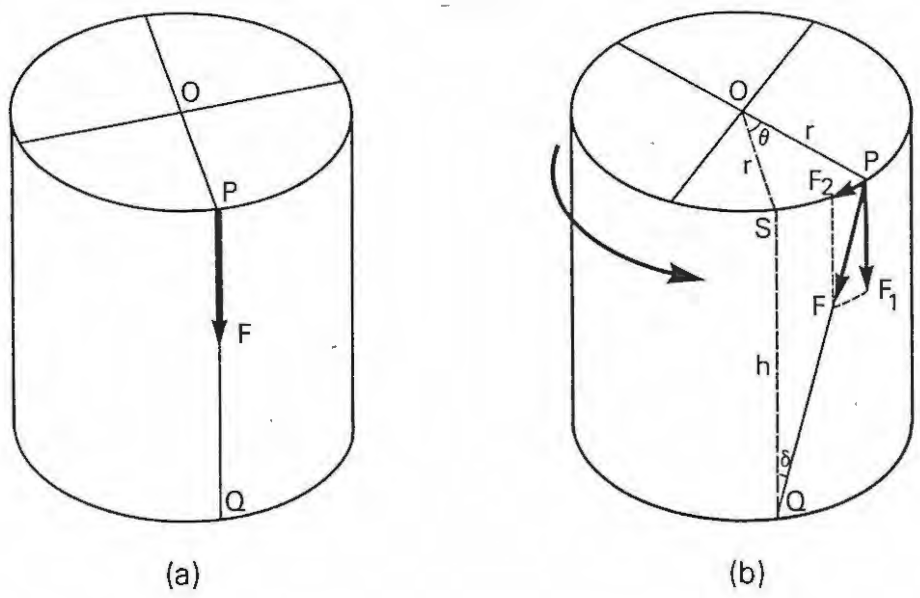

Figure 44 A twisting cylinder with force acting on it

Obviously, the vertical component force decreases and the horizontal component force increases as the strain angle of twisting increases. Suppose that **r** and **h** represent the radius and height of the cylinder respectively. Considering the right triangle PQS in Figure 44b, the following equation for computing strain angle $\boldsymbol{\delta}$ is derived:

$$\delta = \arctan\left[\frac{PS}{SQ}\right] \tag{1}$$

In the triangle POS (Figure 44b), the link PS can be obtained as follows:

$$PS = r\cdot\sqrt{2\cdot(1 - Cos\theta)} \tag{2}$$

Link QS is the height of the cylinder,

$$QS = h \tag{3}$$

Equations (1), (2) and (3) provide the following:

$$\delta = \arctan\left[\frac{r\cdot\sqrt{2\cdot(1 - Cos\theta)}}{h}\right] \tag{4}$$

In the case of the human trunk, erector spinae muscle force (**E**) and rectus abdominis muscle force (**R**) will act at angles to the vertical when the trunk rotates to either right or left side; consequently horizontal and vertical components may be obtained. Therefore, the erector spinae and rectus abdominis muscle forces have vertical components to be involved in trunk extension and flexion, and horizontal components to be involved in trunk rotation under asymmetrical conditions. The oblique abdominis muscle forces change their directions accordingly. In other words, trunk muscle forces are re-orientated in the lumbar region as the human trunk rotates to the right or left of the sagittal plane. These muscle contractions, along with intra-abdominal pressure, compression, shear and torsional moments in the spine, make up the net reaction forces and moments at the lumbo-sacral level.

From insights gained from an extensive literature review, and from the results of the three-dimensional empirical study here reported, the author has developed his own three-dimensional biomechanical model of the lower back (Figure 45). The internal forces and moments considered in this model are: erector spinae muscle force (**E**) assumed to act at $(-x_e,0)$; rectus abdominis force (**R**) at $(x_a,0)$; external oblique abdominis forces on the right ($\mathbf{X}_r$) at $(0,y_o)$, and the left side ($\mathbf{X}_l$) at $(0,-y_o)$; internal oblique abdominis forces on the right ($\mathbf{I}_r$) at $(0,y_o)$, and the left side ($\mathbf{I}_l$) at $(0,-y_o)$; compression ($\mathbf{F}_c$); anterior and lateral shear forces ($\mathbf{S}_x$ and $\mathbf{S}_y$); abdominal pressure force (**P**) at $(x_p,0)$ and torsional moment (**M**) at the L_5/S_1 interspace.

Axes (X,Y,Z) are centred on the spine; X- and Y-axes are in the transverse plane with X directed anteriorly and Y to the left of the body; Z-axis directed upwards. The resultant reaction forces at the level of the section are $\mathbf{F}_X$, $\mathbf{F}_Y$ and $\mathbf{F}_Z$. The resultant reaction moments at the level of the section are $\mathbf{M}_X$, $\mathbf{M}_Y$ and $\mathbf{M}_Z$.

The following assumptions are made with regard to the direction and point of application of the muscle force when the body is in the upright position:

1) external oblique muscle forces run downward-forward at 45^0 to the vertical;
2) internal oblique muscle forces run downward-backward at 45^0 to the vertical;
3) oblique muscle forces are modelled to act in the sagittal plane with their points of application in the same coronal

plane as the spine. They are thus assumed not to be involved in flexion/extension moments;

4) erector spinae and rectus abdominis muscle forces act vertically downwards in the sagittal plane;

5) intra-abdominal pressure force runs upwards vertically in the sagittal plane.

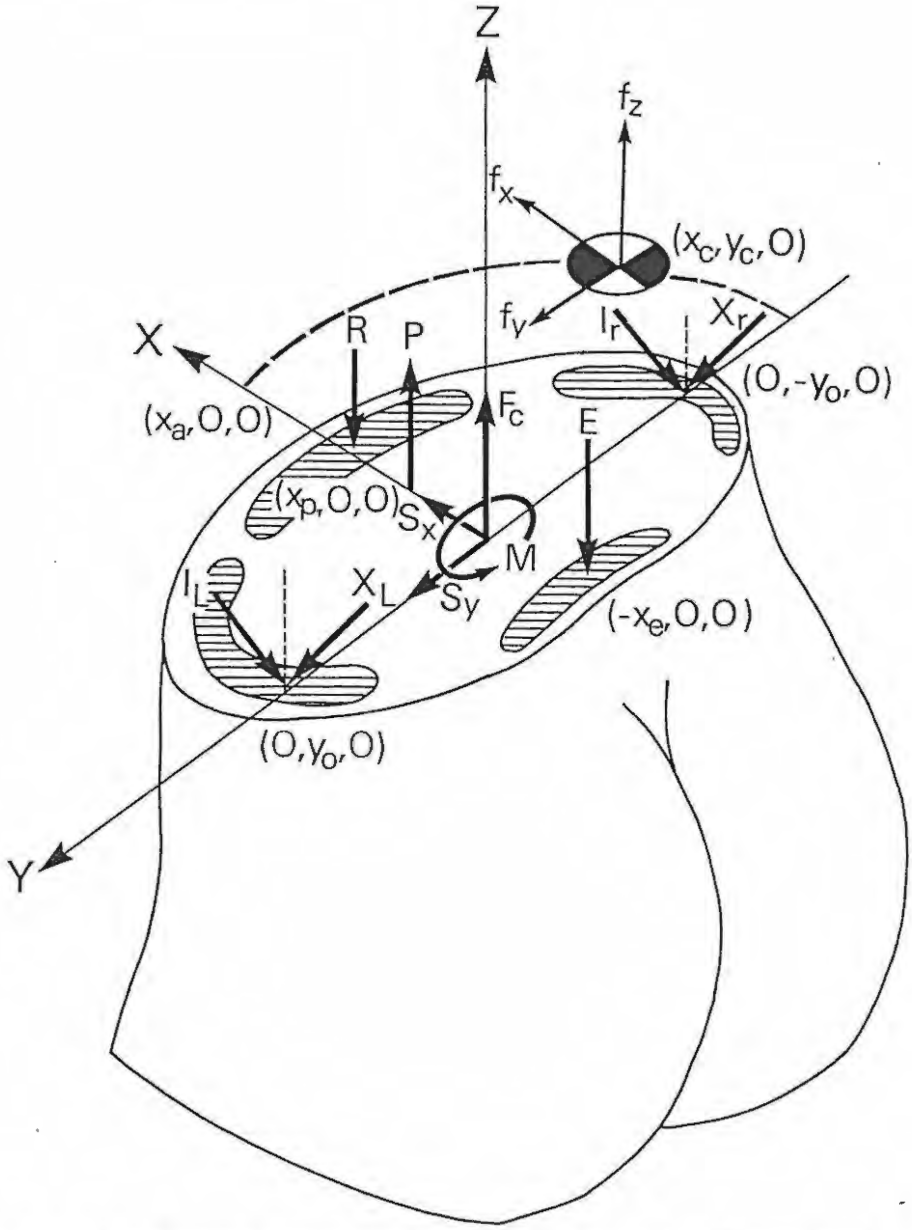

Figure 45 The three-dimensional force model of the lower back

When the human trunk rotates to right or left, the strain angle of twist is different for the various muscles in the lumbar region due to their different rotational radii around the L_5/S_1 disc. Assume that $\boldsymbol{\alpha}$, $\boldsymbol{\beta}$ and $\boldsymbol{\gamma}$ represent the strain angles for erector spinae, oblique abdominis and rectus abdominis, respectively and that anti-clockwise rotation is defined as positive. The equations of equilibrium are as follows:

$$F_X = S_X + X_r \cdot Sin(45° - \beta) - I_r \cdot Sin(45° + \beta) + X_l \cdot Sin(45° + \beta) - I_l \cdot Sin(45° - \beta) \tag{5}$$

$$F_Y = S_Y + E \cdot Sin\alpha - R \cdot Sin\gamma \tag{6}$$

$$F_Z = P + F_C - E \cdot Cos\alpha - R \cdot Cos\beta - X_r \cdot Cos(45° - \beta) - I_r \cdot Cos(45° + \beta) - X_l \cdot Cos(45° + \beta) - I_l \cdot Cos(45° - \beta) \tag{7}$$

$$M_X = I_r \cdot Cos(45° + \beta) \cdot y_O + X_r \cdot Cos(45° - \beta) \cdot y_O - I_l \cdot Cos(45° - \beta) \cdot y_O - X_l \cdot Cos(45° + \beta) \cdot y_O \tag{8}$$

$$M_Y = R \cdot Cos\gamma \cdot x_a - P \cdot x_p - E \cdot Cos\alpha \cdot x_e \tag{9}$$

$$M_Z = M - R \cdot Sin\gamma \cdot x_a - E \cdot Sin\alpha \cdot x_e + X_r \cdot Sin(45° - \beta) \cdot y_O - I_r \cdot Sin(45° + \beta) \cdot y_O - X_l \cdot Sin(45° + \beta) \cdot y_O + I_l \cdot Sin(45° - \beta) \cdot y_O \tag{10}$$

Suppose that a 10kg load(98.1N) is placed at $(x_c, y_c, 0)$ and deviating 60^0 to the right of the sagittal plane. The hand forces ($\mathbf{f_x}$, $\mathbf{f_y}$ and $\mathbf{f_z}$) are recorded at the initiation of the lift. The following basic rules, proposed by Tracy (1990), are adopted here to determine muscle contraction forces.

The rectus abdominis (**R**) should stay inactive if the trunk is providing an extension moment. On the other hand, if the trunk is to provide a flexion moment the erector spinae (**E**) should be inactive. In a situation in which the trunk has to provide mostly lateral flexion to the left, $\mathbf{I_l}$ and $\mathbf{X_l}$ should exert the most force, and $\mathbf{I_r}$ and $\mathbf{X_r}$ should be inactive. On the contrary, $\mathbf{I_l}$ and $\mathbf{X_l}$ should be inactive if $\mathbf{I_r}$ and $\mathbf{X_r}$ are providing a lateral flexion moment to the right. An anti-clockwise twisting moment could be achieved by a combination of $\mathbf{I_l}$ and $\mathbf{X_r}$ while $\mathbf{I_r}$ and $\mathbf{X_l}$ stay inactive. A clockwise twisting moment should be provided by $\mathbf{I_r}$ and $\mathbf{X_l}$ when $\mathbf{I_l}$ and $\mathbf{X_r}$ are inactive.

Reasonable assumptions in this particular lifting task are:

$R=0$; $P=0$; $I_r=0$; $X_l=0$.

Based on Equations (5) to (10), we have the following:

$$F_X = S_X + X_r \cdot Sin(45° - \beta) - I_l \cdot Sin(45° - \beta) \tag{11}$$

$$F_Y = S_Y + E \cdot Sin\alpha \tag{12}$$

$$F_Z = F_C - E \cdot Cos\alpha - X_r \cdot Cos(45° - \beta) - I_l \cdot Cos(45° - \beta) \tag{13}$$

$$M_X = X_r \cdot Cos(45° - \beta) \cdot y_O - I_l \cdot Cos(45° - \beta) \cdot y_O \tag{14}$$

$$M_Y = - E \cdot Cos\alpha \cdot x_e \tag{15}$$

$$M_Z = M - E \cdot Sin\alpha \cdot x_e + X_r \cdot Sin(45° - \beta) \cdot y_O + I_l \cdot Sin(45° - \beta) \cdot y_O \tag{16}$$

The results of the empirical analysis revealed that task asymmetry causes the human trunk to rotate in the transverse plane, and that associated differences in vertical lift height also affect trunk rotation. Transverse rotations of the lumbar spine are presented as a function of vertical lift height and task asymmetry in Table XI. This multiple regression equation has the following form:

$$ROM_{(H)} = 0.003942 \cdot LEVEL + 0.050627 \cdot ASYM \tag{17}$$

where: $ROM_{(H)}$ = transverse rotation (degrees);

LEVEL = vertical lift height (mm);

ASYM = task asymmetry (degrees).

In order to lift the required load, the upper body needs to rotate to the right of the sagittal plane. To illustrate, assume the load is placed 900mm above the floor. The trunk rotation angle of 7^0 is obtained from Equation (17). As discussed earlier, the strain angle of twisting is different for different muscles in the lumbar region. The following trunk cross-sectional geometrical data are obtained from the studies of Kumar (1988) and Tracy **et al.** (1989).

x_a = 80mm; (moment arm for rectus abdominis about L_5/S_1)

x_e = 50mm; (moment arm for erector spinae about L_5/S_1)

x_p = 48mm; (moment arm for abdominal pressure about L_5/S_1)

y_o = 122mm. (moment arm for oblique abdominis about L_5/S_1)

Suppose the link representing the $T_{12}/L_1;L_5/S_1$ interspace has a length of 184mm (Chaffin and Andersson, 1984). The strain angles are calculated as follows using Equation (4):

(1) erector spinae muscle group

$\alpha = 2^0$

(2) oblique abdomini muscle

$\beta = 5^0$

(3) rectus abdomini muscle

$\gamma = 3^0$

Based on body-segment weight data and assuming that the upper body weight (above L_5/S_1) is 53.6% of total body weight (Chaffin and Andersson, 1991) the net reaction forces and moments due to external load and upper body weight can be computed as follows (Figure 45):

$F_x = -f_x;$

$F_y = -f_y;$

$F_z = -f_z;$

$M_x = f_z \cdot y_c + W_b \cdot y_b;$

$M_y = f_z \cdot x_c + W_b \cdot x_b;$

$M_z = -(f_x \cdot y_c + f_y \cdot x_c).$

where: $f_{X,Y,Z}$ = hand forces in three dimensions;

W_b = upper body weight;

(x_b, y_b) = upper body weight acting point on the cross-section;

(x_c, y_c) = hand force acting point on the cross-section.

If the numerical values are assumed as:

$f_x = 10N,$

$f_y = 20N,$

$f_z = 120N, \quad x_c = 200mm, \quad y_c = 350mm;$

$W_b = 400N, \quad x_b = 60mm, \quad y_b = 20mm.$

The net reaction forces and moments are:

$F_x = -10N;$

$F_y = -20N;$

$F_z = -520N;$

$M_x = 50Nm;$

$M_y = 48Nm;$

$M_z = -7.5Nm.$

When the net reaction components and all of the given data are entered into Equations (11) to (16), we have the following:

$$S_x + X_r \cdot Sin50^0 - I_l \cdot Sin50^0 - 10 = 0 \quad (18)$$

$$S_y - E \cdot Sin2^0 - 20 = 0 \quad (19)$$

$$F_c - E \cdot Cos2^0 - I_l \cdot Cos50^0 - X_r \cdot Cos50^0 - 520 = 0 \quad (20)$$

$$X_r \cdot Cos50^0 \cdot 0.122 - I_l \cdot Cos50^0 \cdot 0.122 + 50 = 0 \quad (21)$$

$$48 - E \cdot Cos2^0 \cdot 0.05 = 0 \quad (22)$$

$$M + E \cdot Sin2^0 \cdot 0.05 + X_r \cdot Sin50^0 \cdot 0.122 + I_l \cdot Sin50^0 \cdot 0.122 - 7.5 = 0 \quad (23)$$

From Equations (18), (19), (21) and (22), we find:

$E = 961N$

$S_x = 498N$

$S_y = 54N$

There are still four unknown variables. In order to solve the problem additional assumptions must be made or optimization techniques must be employed.

Linear programming is proposed as the simplest procedure of mathematical optimization. Bean and Chaffin (1988) demonstrated the use of linear programming to calculate muscle contraction forces in biomechanical models. Kromodihardjo and Mital (1986) used linear programming to assess spinal forces. The choice of the "objective function" is the crucial step for optimization procedures. In the present example, the author chose to minimize compression force subject to the constraint that muscle forces not be negative. Equation (20) provides as follows:

$$F_c = 520 + E \cdot Cos2^0 + (I_l + X_r) \cdot Cos50^0$$
$$= 1480 + (I_l + X_r) \cdot Cos50^0$$

From Equation (21), we have: $I_l = X_r + 638$. Then,

$$MINIMIZE \quad F_C = 1890 + 2 \cdot X_r \cdot Cos50^o$$

When $X_r = 0$, $\mathbf{F_c}$ attains the minimum value. Therefore, the internal forces and moment are found to be:

$F_c = 1890N$

$S_x = 498N$ $\qquad$ $S_y = 54N$

$M = -54Nm$

$E = 961N$

$X_r = 0$ $\qquad$ $X_l = 0$

$I_r = 0$ $\qquad$ $I_l = 638N$

The torsional moment on the L_5/S_1 is 54Nm (clockwise). The compression force of 1890N predicted in this example is far below the safe limit of 3400N recommended by NIOSH'81 and NIOSH'91 (NIOSH, 1981; Waters **et al.**, 1993). However, the predicted torsional moment of 54Nm is about 61% of the torque at failure of the intervertebral joint tested by Farfan **et al.** (1970). The author has taken measurements on eight adult human lumbar spines. The average diameter of the fourth and fifth lumbar vertebra was found to be 48.5mm with a coefficient of variation only 0.95%. Dividing the torsional moment by its force arm, the torsional force acting on the peripheral segment of the intervertebral disc can be estimated. A large torsional force of 2250N was obtained for this example.

4.3 DISCUSSION

Tables XIX to XXIII show comparisons between the predictions obtained by the present model and those of Schultz and Andersson (1981). The task, subject stature (1830mm) and body mass (75.6kg) were chosen to conform to the data reported by Schultz **et al.** (1982).

There are only minor differences in predictions by the present model, compared to those of Schultz and Andersson, when tasks are performed symmetrically (Tables XIX, XX and XXI). In force calculation via the present model, however, it should be noted that the upper body superior to the third lumbar vertebra, the head and both arms, were considered as one unit.

Table XIX Symmetrical: standing relaxed

Forces(N) & Moment(Nm)	Present Model		Schultz and Andersson's Model	
L_3-Compression	469		470	
L_3-Anterior Shear	0		0	
L_3-Lateral Shear	0		0	
L_3-Torsional Moment	0		—	
Muscle Forces(N):	Left	Right	Left	Right
Erector spinae	60	60	60	60
External oblique	0	0	0	0
Internal oblique	0	0	0	0

Table XX Symmetrical: arms abducted holding 2kg load in each hand

Forces(N) & Moment(Nm)	Present Model		Schultz and Andersson's Model	
L_3-Compression	522		500	
L_3-Anterior Shear	0		0	
L_3-Lateral Shear	0		0	
L_3-Torsional Moment	0		—	
Muscle Forces(N):	Left	Right	Left	Right
Erector spinae	66	66	50	50
External oblique	0	0	0	0
Internal oblique	0	0	0	0

Table XXI Symmetrical: resisting a 44.3Nm flexion moment

Forces(N) & Moment(Nm)	Present Model		Schultz and Andersson's Model	
L_3-Compression	1355		1390	
L_3-Anterior Shear	147		150	
L_3-Lateral Shear	0		0	
L_3-Torsional Moment	0		—	
Muscle Forces(N):	Left	Right	Left	Right
Erector spinae	503	503	510	510
External oblique	0	0	0	0
Internal oblique	0	0	0	0

For a standing position, holding a 2kg (19.2N) load in each hand (Table XX), the erector spinae muscle force predicted by the present model is 32% higher than that estimated with Schultz and Andersson's model, due to the fact that the latter model allows latissimus dorsi to share the required effort with the erector spinae. They do this efficiently because they act over a large lever arm. Spinal compression force via the present model is 4.4% higher than that of Schultz and Andersson.

For tasks involving a twisting action (Tables XXII and XXIII), present estimates of external oblique muscle force and shear forces are much higher, while erector spinae muscle force is lower, than that predicted by Schultz and Andersson's model. The present model considered only left-side external oblique muscles to be involved in twists to the right. Compared to the symmetrical tasks (Tables XIX and XX), twisting decreases erector spinae muscle forces and increases oblique muscle force in the present model. In other words, twisting brings in a new distribution of muscle forces in the lumbar region. The decrease in compression caused by twisting is very small, but a dramatic increase in anterior shear force occurs; and the proposed model brings a torsional moment into effect.

Table XXII Asymmetrical: standing with 45^0 twist to the right at shoulder level*

Forces(N) & Moment(Nm)	Present Model		Schultz and Andersson's Model	
L_3-Compression	469		480	
L_3-Anterior Shear	−28		0	
L_3-Lateral Shear	3		0	
L_3-Torsional Moment	3		—	
Muscle Forces(N):	Left	Right	Left	Right
Erector spinae	42	42	50	50
External oblique	45	0	10	10
Internal oblique	0	0	0	0

* assume this case is similar to that of the 90^0 task asymmetry condition in the empirical study

Table XXIII Asymmetrical: arms lateral holding 2kg load in each hand with 45^0 twist to the right at shoulder level*

Forces(N) & Moment(Nm)	Present Model		Schultz and Andersson's Model	
L_3-Compression	521		680	
L_3-Anterior Shear	−32		0	
L_3-Lateral Shear	4		0	
L_3-Torsional Moment	4		—	
Muscle Forces(N):	Left	Right	Left	Right
Erector spinae	47	47	100	150
External oblique	50	0	0	10
Internal oblique	0	0	0	10

* assume this case is similar to that of the 90^0 task asymmetry condition in the empirical study

In Schultz and Andersson's model the influence of twist on muscular tension, compression and shear forces is not consistent: twisting dramatically increases erector spinae muscle forces when abducted arms hold 2kg bilaterally. Compression, however, remains the same and erector spinae muscle forces decrease when standing relaxed with 45^0 twist; i.e., twist shows no effect on shear forces.

By 1991 the original NIOSH'81 lifting equation had been extended to apply to a larger percentage of lifting tasks (Waters **et al.**, 1993). An asymmetric multiplier was included in the revised 1991 lifting equation, the effect of which was that a 30% reduction in the weight limit was recommended for lifts involving an asymmetrical twist of 90^0. This asymmetric multiplier (AM) was determined as follows:

$$\mathbf{AM = [1 - (0.0032 \bullet A)]}$$

where: A = the angle between the sagittal plane and the plane of asymmetry.

Tables XIX to XXIII generally show that compression and shear forces and torsional moments predicted by the present model are greater in the asymmetrical condition than that in the symmetrical condition. Obviously, these predicted forces and moments will increase as a function of increased asymmetry. Table XXIV displays the forces and moment predicted via the present model, along with the weight limits recommended by the NIOSH'91 lifting equation under various lifting conditions.

Table XXIV Force and moment predictions by the Li model, along with the load limit recommended by NIOSH'91

Asymmetry	NIOSH'91	Forces and Moment Predicted via the Present Model (10kg load)						
	Load Limit(kg)	F_C (N)	S_X (N)	S_Y (N)	M (Nm)	E (N)	I_l (N)	X_r (N)
0^0	23.0	1092.4	0	0	0	618.3	0	0
30^0	20.8	1137.4	126.2	11.5	16.0	549.4	170.1	0
60^0	18.6	1030.4	230.4	9.9	28.6	356.1	305.3	0
90^0	16.4	802.6	283.7	9.8	35.1	90.9	370.4	0

The average stature (1572mm) and body mass (71.5kg) of the subjects of the present empirical study were used in the force computations via the present model. The subject is assumed to be lifting a 98.1N load placed first in the sagittal plane, and then successively at 30^0, 60^0 and 90^0 to the right of the sagittal plane. The vertical and horizontal locations were maintained at 740mm and 250mm, respectively, under the various asymmetrical conditions. Ideal conditions are assumed for the vertical, horizontal, lift distance, frequency and coupling multipliers when the recommended weight limit is calculated using the 1991 lifting equation as the asymmetrical angle increases from 0^0 to 90^0.

Generally the weight limits recommended by NIOSH'91 decrease with an increase in asymmetrical angle, while the forces predicted via the Li model show a different trend. The compression (F_C) increases slightly when asymmetrical angle increases from 0^0 to 30^0, and decreases after 30^0 asymmetry. The erector spinae muscle force decreases with increasing asymmetrical angle. This is due to the fact that when the asymmetrical angle increases from 0^0 to 90^0, the effective force arm of the load to the lateral axis decreases. This results in a decrease of flexion moment caused by the external load. Consequently, the erector spinae produce lesser force to counteract the forward bending torque at more asymmetric positions. However, the effective force arm of the load to the posterior-anterior axis increases while increasing asymmetrical angle. This causes a greater side bending moment about the posterior-anterior axis. Therefore, the oblique muscles on the opposite side have to produce a greater force to counterbalance the side bending moment. It is evident that the internal oblique muscle force on the left side (I_l) increases dramatically when asymmetrical angle increases from 0^0 to 90^0. As a result anterior shear force and torsional moment increase. The anterior shear forces (S_X), internal oblique muscle force on the left (I_l) and torsional moment (M) have similar increase rates. These are greatest from 0^0 to 30^0 and least from 60^0 to 90^0. This indicates that the increases in these forces and in the moment are not well-proportioned over 90^0 asymmetry. This is congruent with the empirical findings of the present study, in which the lumbar motion factors in the horizontal plane, especially at the higher level, increased at a much reduced rate from 60^0 to 90^0 compared to the increase before 30^0. With less discrimination the weight limit recommended by NIOSH'91 shows a rate of decrease that is evenly distributed over the 90^0 asymmetry; approximately 10.7% per every 30^0 deviation from sagittal plane.

The model constructed in the present study was developed primarily on the basis of the 10-muscle model of Schultz and Andersson (1981), but it incorporates the simplifications proposed by Tracy (1990), and the muscle forces considered are as those in Tracy's model. What is innovatively proposed in the present model, however, is that the lower back is modelled to be able to rotate around its longitudinal axis, and it embraces the re-orientation of muscle forces that must occur as a result of trunk rotation during asymmetrical lifting. In addition, a combined loading of the spine including compression, shear forces and torsion, is incorporated into the new model. From the examples included it is evident that the model here developed is highly congruent with Schultz and Andersson's (1990) model under symmetrical lifting conditions. However, during asymmetrical lifting, forces generated by the present model differ from those estimated by Schultz and Andersson under similar circumstances, because of the force re-orientation in the lumbar region caused by incorporation of a "strain angle of twist". The torsional moment at the L_5/S_1 disc is also taken into account in the present model under asymmetrical conditions.

During manual lifting, three types of stress vectors are transmitted through the human spine: compression, shear and torsional forces. Although twisting, besides work intensity, has been identified as a common cause of lower back pain, only compressive force is traditionally selected as the critical vector in risk assessment (NIOSH, 1981; Waters **et al.**, 1993). A safe limit of 3.4kN compression on the L_5/S_1 is recommended by NIOSH'81. However, asymmetrical lifting studies (e.g. Mital and Kromodihardjo, 1986; Gallagher **et al.**, 1994) have found that lumbar compression is lower when a lifting task is asymmetric; and the cost of this decrease in compression is that shear forces acting on the lumbar are increased. According to Snook (1978), asymmetrical lifting is associated with a large proportion of lower back pain cases. Based on this evidence, Gallagher and associates speculated that the traditional biomechanical parameter used for ergonomic design of lifting tasks (compression) may not relate well to the higher incidence of lower back pain associated with asymmetrical motion.

The present study has clearly demonstrated that three-dimensional asymmetrical movement is a feature of manual lifting. The task asymmetry has a significant effect on the kinematics of the lumbar spine in both horizontal and frontal planes. Similarly, the kinetics of asymmetrical lifting in the horizontal plane, particularly in a lateral direction, increase dramatically with an increase in task asymmetry. It was also found that the kinetics are transferred from vertical to horizontal under asymmetrical lifting conditions. These kinematic and kinetic characteristics in the horizontal plane are most likely to connect with shear and torsional stresses acting on the lumbar spine. Therefore, it could be deduced that asymmetrical lifting exacerbates shear and torsional stresses on the spine.

Twisting while lifting is commonly recognized as more hazardous to the musculoskeletal system than symmetrical lifting because of the combined effects of flexion, lateral flexion and accompanying axial rotation of the spine. Farfan **et al.** (1970) conducted a cadaveral study and tested the torque strength of the human lumbar spine. The average torque for the intact whole intervertebral joint was found to be 88.14Nm. Jager and Luttmann (1992) found a large variability in the torsional strength recorded in various investigations. This may be due to declines in lumbar strength with age, bone mineral content, and degenerative changes (Hansson and Roos, 1981). Unfortunately, no study has been undertaken to explore the direct relationship between lifting-related lower back pain and predicted torsional stress on the L_5/S_1 disc. Possible reasons for this are: technological difficulties in capturing 3-D asymmetrical motion; complication of 3-D force analysis; and lack of a suitable biomechanical model to estimate torsional forces acting on the intervertebral joints of the human spine. The Li model proposed in the present study is capable of estimating the combined stresses imposed on the human spine: compression, shear forces and torsion. This model may enable us to explore the role of torsional force, in association with compression and shear forces, in production of lower back pain.

The choice of any model depends on the level of detail required by its estimates. In manual materials handling research there has been much emphasis on calculating sagittal plane spinal compression forces. This may be because "NIOSH 1981" had set limits for these forces in symmetrical lifting. However, discrepancies have been revealed between observed and estimated risks during asymmetrical lifting when only compression force has been considered in risk assessment (Mital and Kromodihardjo, 1986; Gallagher **et al.**, 1994). The model here proposed incorporates a torsional moment at L_5/S_1 which surely should be considered as an additional risk factor in asymmetrical lifting analyses.

CHAPTER 5: SUMMARY, CONCLUSIONS AND RECOMMENDATIONS

5.1 INTRODUCTION

Most manual lifting involves three-dimensional dynamic motion of the human spine and of the object lifted. A number of investigators have evaluated dynamic effects on the loading of the human spine. One consistent finding has been that dynamic motion greatly increases the biomechanical stresses on the spine. It is generally accepted that the combined effects of flexion and axial rotation of the spine caused by asymmetrical lifting exacerbate the stresses on the musculoskeletal system. In the present study, the kinematics of the human spine and the kinetics of lifting were investigated during asymmetrical handling tasks.

Considering the torsional moment on the spine and the mechanism of muscle force re-orientation in the lower back caused by trunk rotation, a three-dimensional force model of the lower back was developed as part of this project, which was carried out in two phases: an empirical phase, during which kinematic and kinetic data were collected under symmetrical and asymmetrical lifting conditions, and a conceptual phase in which a 3-D biomechanical model was constructed to account for the observed kinematics of asymmetrical lifting. The following research hypotheses were tested in the empirical study ($p<0.05$ level of significance):

1) Spinal kinematic measures during manual lifting are not affected by differences in task asymmetry or initial lift height.

2) The kinetic characteristics of manual lifting are not affected by differences in task asymmetry or initial lift height.

5.2 SUMMARY OF PROCEDURES IN THE EMPIRICAL STUDY

In the present investigation, asymmetrical lifting was studied in the sagittal plane (0^0), and 30^0, 60^0 and 90^0 lateral planes to the right. A box was placed perpendicular to one of those planes. The 10kg load lifted in the study was determined by applying the NIOSH'81 guideline for manual lifting. Subjects were instructed to keep their feet in the sagittal plane and required to lift the box using both hands from two initial vertical grip heights of 150mm and 500mm to an 800mm high bench in the sagittal plane. Each subject lifted the box using whatever working posture was found to be most comfortable. The initial horizontal distance between the mid-point of the link between the ankles and the centre of the box was controlled at 350mm. A total of eight lifting tasks (2 vertical heights x 4 task asymmetries) was randomly assigned to each of the subjects in the study.

A V-scope ultrasonic motion analyzer was used for measuring the movements of the box in three-dimensional space. The V-scope system consists of a microcomputer, three towers and a set of buttons. The system measures the time of flight of the ultrasonic signal between button and towers. The position of the buttons in three-dimensional space is computed by multiplying the speed of sound in air by the time delay. The velocity and acceleration are obtained from a time-function. In the present study, three V-scope towers were installed above the working area and a V-scope button was attached to the centre of the box being lifted. The sampling time was set at 10ms. A personal computer was used for further data process and display.

During the manual lifting activities, the kinematics of the lumbar spine of each subject were recorded by a Lumbar Motion Monitor (LMM). The lumbar motion monitor is essentially an exoskeleton of the spine that has been instrumented with sensors. The output of these sensors is transmitted through an umbilical cable to a laptop computer, where instantaneous position, velocity and acceleration of the lumbar spine are calculated. The data collection rate of the LMM is 60Hz.

Eleven South African Black male manual workers, ranging in age between 27 and 56 years volunteered to participate in this study. Subjects were informed as to the purpose of the experiments and signed an informed consent. Thereafter, the subject's anthropometric parameters and isometric strength were measured. Each subject performed the eight lifting tasks to become familiar with the experimental procedures, laboratory setting and equipment.

During the data collection, the lumbar motion monitor was mounted on the subject's back. The kinematic data reflecting lumbar spinal motion and that of the box being lifted were simultaneously recorded by the lumbar motion monitor and the V-scope system, while the subject was engaged in the lifting tasks. The data were processed using the analytical protocol developed for the study.

Results were analysed using one- and two-way analyses of variance for assessing task asymmetry and lift grip-height effects on kinematics of the human spine and kinetics of the load. Multiple regression was applied to evaluate the relationship between independent and dependent variables.

5.3 SUMMARY OF RESULTS AND CONCLUSIONS: EMPIRICAL STUDY

5.3.1 Kinematics of the Human Spine

The range of motion (ROM) of the human spine in three reference planes increased as a function of increasing task asymmetry. ROM in the sagittal and horizontal planes at the lower level (150mm) was greater ($p<0.05$) than that at the higher level (500mm). There was no significant difference in the frontal plane ROM between the two vertical lift heights. ROM differed ($p<0.05$) at various task asymmetries in the frontal plane at both vertical levels and only at the higher level in the horizontal plane, while showing no difference under various task asymmetries in the sagittal plane.

The angular peak and average velocity increased in the three reference planes when task asymmetry increased from 0^0 to 90^0. The peak velocity showed a more reactive response to task asymmetry at the higher level (500mm). There was a significant difference in peak velocity between the two vertical lift heights in the three reference planes and in average velocity in the sagittal plane. Task asymmetry displayed a significant effect on the peak and average velocity at both vertical levels in the sagittal and frontal planes, but only at the higher level in the horizontal plane.

The peak and average angular acceleration in the three reference planes increased with an increase in task asymmetry, except for the case of average acceleration in the sagittal plane. The increase was more dramatic at the higher lift level (500mm). A significant difference in the peak acceleration was found between the two vertical levels in the three reference planes. Task asymmetry had a significant effect on peak and average acceleration in the frontal plane, but only on peak acceleration at the higher level in the horizontal plane. The average acceleration in the sagittal plane differed ($p<0.05$) between the two vertical levels.

5.3.2 Kinetics of Asymmetrical Lifting

The duration of asymmetrical lifting increased nominally with an increase in task asymmetry and differed significantly between the two vertical heights. The lift cycle times were 1.33—1.39s for the 150mm lift height and 1.20—1.26s for the 500mm lift height.

The maximum and average velocities increased as a function of increasing task asymmetry. The peak resultant velocities ranged from $1.98 m.s^{-1}$ to $3.13 m.s^{-1}$ and $1.68 m.s^{-1}$ to $2.11 m.s^{-1}$ for the lower (150mm) and higher (500mm) lift levels, respectively. The average resultant velocities were $0.80-1.09 m.s^{-1}$ for the lower level and $0.80-0.97 m.s^{-1}$ for the higher level. While the velocity in the vertical direction at the lower level was greater than that at the higher level, there was no difference in the velocity in the horizontal plane between the two levels. Task asymmetry showed significant effects on the velocities in the horizontal plane.

The resultant peak accelerations were 8.38—12.93m.s^{-2} and 7.76—9.48m.s^{-2} for the lower and higher levels, respectively. The average accelerations were 4.05—4.64m.s^{-2} for the lower level and 4.08—4.23m.s^{-2} for the higher level. The acceleration in the vertical direction differed markedly between the two vertical levels. Task asymmetry displayed significant effects on accelerations only in the horizontal plane.

The peak resultant forces ranged from 170.3N to 209.1N for the lower lift level (150mm) and 169.4N to 182.5N for the higher lift level (500mm). The average resultant forces were 102.4—103.2N and 101.3—102.1N for the lower and higher levels, respectively. The vertical peak force was significantly different between the two vertical levels. Peak and average forces in the horizontal plane differed ($p<0.05$) at various task asymmetries.

Both maximum and mean power was significantly different between the two vertical levels. Maximum power outputs were 255.3—516.0W and 221.2—292.6W for the lower (150mm) and higher (500mm) levels, respectively. Average power was 86.4—114.2W for the lower level and 76.3—91.3W for the higher level. Work differed significantly between the two vertical levels and ranged from 119.1J to 172.5J at the lower level and 91.8J to 112.1J at the higher level, but was not different between various task asymmetries.

5.3.3 Conclusions

The kinematic characteristics of the human lumbar spine as studied in this project generally increased as a function of increasing task asymmetry, the sole exception being average acceleration in the sagittal plane. The rate of this increase in the horizontal and frontal planes was greater than that in the sagittal plane. The increase showed a more reactive response to task asymmetry at the 500mm lift height than at the 150mm lift height. However, sagittal plane measures dominated through the full range of motion.

The kinetic characteristics of manual lifting in general increased with an increase in task asymmetry. This increase was more dramatic in the lateral direction in the horizontal plane. Kinetic measures in the vertical direction dominated through the full range of the lifting.

In light of the results of the empirical study, the following conclusions can be drawn:

Hypothesis 1:

Between Un-planar and Cross-planar Lifts	
Significant differences were found in:	**No differences were found in:**
(1) Sagittal plane: Peak velocity; Average velocity at the high level.	(1) Sagittal plane: Range of motion; Average velocity at the low level; Peak and average accelerations.
(2) Frontal plane: All measured spinal kinematic parameters.	(2) Frontal plane: ———
(3) Horizontal plane: All measured spinal kinematic parameters at the high level, except for average acceleration.	(3) Horizontal plane: All measured spinal kinematic parameters at the low level; Average acceleration at the high level.
Decision: H_0 Rejected	Decision: H_0 Retained

Between Low and High Lift Height Levels	
Significant differences were found in:	**No differences were found in:**
(1) Sagittal plane: All measured spinal kinematic parameters.	(1) Sagittal plane: ———
(2) Frontal plane: Peak velocity and acceleration.	(2) Frontal plane: Range of motion; Average velocity and acceleration.
(3) Horizontal plane: Range of motion; Peak velocity and acceleration.	(3) Horizontal plane: Average velocity and acceleration.
Decision: H_0 Rejected	Decision: H_0 Retained

Thus, in the sagittal plane the hypothesis 1 was rejected in respect of lift height and retained in respect of task asymmetry. In the frontal plane, the hypothesis was rejected in respect of task asymmetry. With regard to the horizontal kinematics, the hypothesis was rejected between the uni-planar and cross-planar lifts only at the high level. The lift height had a significant effect on the peak velocity and acceleration in both frontal and horizontal planes, and range of motion in the horizontal plane.

Hypothesis 2:

Between Un-planar and Cross-planar Lifts	
Significant differences were found in:	**No differences were found in:**
(1) Vertical direction: Average acceleration.	(1) Vertical direction: All measured kinetic characteristics, except for average acceleration.
(2) Anterior-posterior direction: All measured kinetic characteristics.	(2) Anterior-posterior direction: ————
(3) Lateral direction: All measured kinetic characteristics.	(3) Lateral direction: ————
(4) Trajectory: Peak resultant acceleration.	(4) Trajectory: All measured kinetic characteristics, except for peak resultant acceleration.
Decision: H_0 Rejected	Decision: H_0 Retained

Between Low and High Lift Height Levels	
Significant differences were found in:	**No differences were found in:**
(1) Vertical direction: All measured kinetic characteristics, except for average force.	(1) Vertical direction: Average force.
(2) Anterior-posterior direction: ————	(2) Anterior-posterior direction: All measured kinetic characteristics.
(3) Lateral direction: Peak and average accelerations; Peak and average forces.	(3) Lateral direction: Peak and average velocities.
(4) Trajectory: Peak resultant velocity; Peak resultant acceleration; Peak resultant force; Peak and average powers; Work.	(4) Trajectory: Average resultant velocity; Average resultant acceleration; Average resultant force.
Decision: H_0 Rejected	Decision: H_0 Retained

Thus, the hypothesis 2 was rejected in respect of lift height except for average force, and retained in respect of task asymmetry in the vertical direction except for average acceleration. In the anterior-posterior direction, hypothesis was rejected in respect of task asymmetry and retained for lift height. In the lateral direction the hypothesis was rejected in respect of task asymmetry and lift height with the exception of velocity. With regard to the trajectorial measures, the hypothesis was retained in respect of task asymmetry, except for the peak resultant acceleration and rejected in respect of lift height with regard to the trajectorial peak values and work.

5.4 THREE-DIMENSIONAL BIOMECHANICAL FORCE MODEL OF THE LOWER BACK

The human spine is not a single rod but rather a series of small cylinders with flexible intervertebral discs interposed. Any relative movement between these vertebrae brings stress to bear on the associated discs. Compression and shear forces are commonly recognised risk factors in occupational lower back disorders. However, torsional force caused by spinal rotation in the transverse plane appears not to have been taken into account in previous biomechanical models. Yet, according to Farfan **et al.** (1970), it is easier to deform the disc by torsion than by axial compression, in which case torsional moments may cause disc annular protrusion, which is one of the main sources of lower back pain.

Almost all lifting tasks involve three dimensional motions of the human spine. The rotations of the spine in the transverse plane change the geometric configuration of muscles in the lumbar region and consequently result in a re-orientation of muscle forces during lifting. This force re-orientation exacerbates the stresses on the disc in the transverse plane, specifically shear force and torsional moment.

The model proposed in the present study accounts not only for compression and shear forces on the disc, but the torsional moment as well. The effect of spinal rotation on muscle force re-orientation is also taken into account in the model. The present model has similar predictability to that of Schultz and Andersson (1981), when the human trunk exerts only a flexion-extension moment in the sagittal plane without involving spinal rotation. Under the condition of asymmetrical loading, however, stresses on the human spine predicted by the present model differ from the estimates in Schultz and Andersson's model. This difference is most dramatic in the transverse plane. The advantage of the present model is that the lower back is considered as a flexible joint with three-dimensional force distribution; it is capable of quantifying torsional stress in the human spine and is offered as a further improvement over previous models for use with tasks requiring twisting movements of the trunk.

Although "NIOSH'91" has acknowledged torsional force as one of the stress vectors on the spine, compressive force was still chosen as the biomechanical criterion for the revised lifting equation. The present model has demonstrated that the torsional moment and shear forces on the spine increase dramatically with an increase in asymmetric angle, while the increase in compression is inconsiderable. NIOSH'91 incorporates a 30% decrease in the weight limit for 90^0 asymmetrical lifting, and this 30% reduction is evenly distributed over 90^0. However, the model developed in the present study indicates that increases in the torsional moment and shear forces are not equally divided within 90^0 asymmetry.

5.4.1 Utility of the Model

The three-dimensional low back force model developed in the present study includes consideration of the lower back being able to rotate around its longitudinal axis. The conceptual background of the model was to incorporate a torsional moment on the human spine and the mechanism of muscle force re-orientation in the lumbar region during asymmetrical lifting, and to permit the evaluation of compression, shear force and torsional stress on the human spine.

Nominal differences were observed between the predictions by the present model and Schultz and Andersson's model under symmetric conditions. However, the further away from the sagittal plane in which the lift was performed, the greater were these difference.

The model proposed by the present study predicts dramatic increases in shear forces, oblique muscle forces and torsional moment under asymmetrical lifting conditions. The increase rates in these forces and moment were not linearly related over task asymmetric angle.

5.5 RECOMMENDATIONS

The following recommendations for further study merit consideration:

1) The kinetics of manual lifting should be studied in three dimensions: two-dimensional analysis tends to obfuscate, rather than add clarity to our understanding of this complex area of study.

2) The kinematics of the lumbar spine should be investigated at diverse vertical levels in order to further clarify the effects of lift height on spinal dynamic motions under asymmetrical lifting conditions.

3) The relationship between lumbar spinal rotation and angle of task asymmetry at any vertical lift height should be further evaluated.

4) The mechanism of muscle force re-orientation in the lumbar region during asymmetrical lifting should be further investigated and justified.

5) Epidemiological studies should be undertaken to explore the specific relationship between back pain and spinal torsion and shear forces.

6) The future project designed to test the 3-D model developed in the present study in conjunction with EMG analysis, should provide a fruitful new avenue for research in asymmetrical lifting.

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APPENDIX 1

SUBJECT CONSENT FORM

RHODES UNIVERSITY

DEPARTMENT OF HUMAN MOVEMENT STUDIES

SUBJECT CONSENT FORM

TITLE OF RESEARCH : "**THREE DIMENSIONAL KINETIC ANALYSIS OF ASYMMETRICAL LIFTING**"

PROCEDURES, RISKS AND BENEFITS

Only two sessions (each 45 minutes) are required for each selected subject to participate in the research. In the first session the body mass, stature, arm strength, back strength and leg strength will be measured. This will be followed by a habituation of the experimental technology and equipment,and manual lifting activities of 10 minutes practice. In the second session subjects are asked to lift a box of 10kg in the sagittal plane (0^0), and in 30^0, 60^0 and 90^0 lateral planes to the right, from two initial vertical heights of 150mm and 500mm to a height of 800mm. These eight lifting tasks will be randomly assigned to the subject. Each task will last 5 seconds with more than five minutes rest between lifts. The kinematics of the box and the lumbar spine of the subject will be measured using a V-scope Ultrasonic Motion Monitor and a Lumbar Motion Monitor in three dimensional space.

There are no additional risks that may be encountered during those data collection sessions. The task will be performed infrequently, less than once per five minutes. According to "Work Practice Guide for Manual Lifts" (Technical Report, National Institute of Occupational Safety and Health, U.S.A., Department of Health and Human Service, Cincinnati, Ohio, 1981), the 10kg weight of load to be lifted in the experiment is below the Action Limit (AL) which represents the lifting capabilities of 99% of the general male population.

The benefits to be obtained include personal information on your arm strength, back strength and leg strength. Furthermore, you will be providing a valuable service to the advancement of our knowledge in this area of human performance.

I, ______________________ have been fully informed of the nature of the research entitled **THREE DIMENSIONAL KINETIC ANALYSIS OF ASYMMETRICAL LIFTING** and do hereby give my consent to act as a subject in the above-named research.

I am fully aware of the procedures involved as the potential risks and benefits attendant to my participating as explained to me verbally and in writing. In agreeing to participate in this research, I waive any legal recourse against the researchers or Rhodes University, from any and all claims resulting from personal injuries sustained while being tested. This waiver shall be binding upon my heirs and personal representatives. I realize that it is necessary for me to promptly report to the researcher any signs or symptoms of discomfort indicating any abnormality or distress.

I am aware that I may withdraw my consent and withdraw from participation in the research at any time. I am aware that my anonymity will be protected at all times, and agree that the information collected may be used and published for statistical or scientific purposes.

I have read the foregoing and I understand it. Any questions which may have occurred to me have been answered to my satisfaction.

PERSON INVOLVED	PRINT NAME	SIGNATURE	DATE
Subject (or legal representative)			
Person Administering Informed Consent			
Witness			
Project Supervisor			

APPENDIX 2

KINEMATIC CHARACTERISTICS OF THE LUMBAR SPINE

0^0-150mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.6667	0.45	0.35	0.4333	0.2833	0.2333	0.2667	0.3167	0.4833	0.6333	0.2333	0.39	0.15
2	Tv(f)	1.0667	1.1	1.3833	1.1667	1.4	1.4	1.15	0.6167	1.933	1.4166	1.0833	1.24	0.16
3	Tv(h)	0.6667	0.2666	1.2833	1.0833	0.9	0.9333	1.2333	0.5834	1.0833	0.9666	0.8	0.89	0.15
4	Ta(s)	0.1334	0.3166	0.2166	0.1333	0.15	0.1167	0.15	0.15	0.35	0.1166	0.1167	0.89	0.30
5	Ta(f)	0.2167	0.3833	0.9166	0.1667	1.3	0.8333	1.1	0.85	1.25	0.5	0.2167	0.18	0.08
6	Ta(h)	0.45	0.9166	0.9166	0.6167	0.8833	0.75	1.15	0.5	0.9833	0.9	0.6833	0.8	0.10
7	ROM(s)	63	80	83	68	86	68	72	57	78	66	87	73.5	9.96
8	ROM(f)	11	6	6	10	5	13	6	6	4	15	12	8.5	3.75
9	ROM(h)	5	3	2	5	3	4	2	6	4	4	4	3.8	1.25
10	V(s)	96	124	114	105	149	97	100	57	110	81	113	104.2	23.47
11	V(f)	23	18	14	15	18	31	17	19	10	19	22	18.7	5.44
12	V(h)	10	9	6	8	4	11	5	7	6	5	8	7.2	2.23
13	V(s)mean	36.1	40.6	31.1	30.8	41.6	31.2	26.9	29.3	32.8	25.8	40.9	33.4	5.62
14	V(f)mean	4.4	2.8	2.0	1.7	1.9	4.9	0.1	3.9	1.3	1.2	3.7	2.5	1.52
15	V(h)mean	2.7	1.8	1.0	2.3	0.5	2.2	0.6	2.6	1.7	1.3	1.8	1.7	0.76
16	A(s)	417	478	486	308	671	476	487	235	520	197	554	439	141.01
17	A(f)	106	86	73	68	72	75	64	83	51	76	100	77.6	15.67
18	A(h)	45	30	22	17	20	43	27	30	28	15	40	28.8	10.25
19	A(s)mean	5.2	0.7	2.7	5.1	6.8	5.4	0.6	2.1	8.9	3.9	3.6	4.1	2.54
20	A(f)mean	1.1	0.2	0.2	2.4	4.1	1.6	1.9	0.4	0.5	0.7	2.1	1.4	1.2
21	A(h)mean	1.0	0.7	0.3	0.5	0.4	1.0	0.1	1.0	0.6	0.3	0.1	0.3	0.34

30^0-150mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.3334	0.7166	0.4167	0.2834	0.2333	0.2334	0.2833	0.6333	0.4166	0.2833	0.2333	0.37	0.17
2	Tv(f)	0.0167	1.0	0.2833	1.3167	0.4833	1.4834	1.0833	1.4	0.0116	1.45	1.1167	0.88	0.26
3	Tv(h)	0.9834	0.3666	1.4833	0.9834	1.1	0.8	1.1	0.6166	1.4166	0.3	0.7167	0.9	0.19
4	Ta(s)	0.2	0.35	0.2833	0.15	0.1166	0.1167	0.15	0.1833	0.2833	0.15	0.1167	0.19	0.08
5	Ta(f)	0.25	0.4	0.1167	1.1167	0.3833	0.2167	0.1833	0.1	0.4166	1.3666	1.0333	0.42	0.41
6	Ta(h)	0.8667	0.75	1.0333	0.6834	0.15	0.65	1	0.4	0.9166	0.25	0.5667	0.66	0.15
7	ROM(s)	69	75	82	83	83	69	71	64	86	75	83	76.4	7.42
8	ROM(f)	8	8	11	8	14	11	8	11	4	13	13	9.9	2.98
9	ROM(h)	5	3	3	6	3	4	4	6	5	4	4	4.3	1.1
10	V(s)	102	108	118	123	165	105	109	91	115	96	109	112.8	19.6
11	V(f)	17	21	23	16	22	14	25	19	8	15	22	18.4	4.9
12	V(h)	8	7	6	12	7	8	11	12	8	6	9	8.5	2.2
13	V(s)mean	31.8	38.5	30.7	38.2	43.3	22.3	23.2	30.2	36.8	27.6	39.1	32.8	6.7
14	V(f)mean	3.3	3.6	4.1	0.8	7.0	1.4	0.3	4.8	0.4	1.6	1.3	2.6	2.13
15	V(h)mean	2.4	1.2	0.8	3.6	0.1	1.7	1.5	4.3	3.0	1.9	2.0	2.0	1.23
16	A(s)	407	381	466	561	821	519	521	282	513	447	531	495.4	135.19
17	A(f)	40	75	97	70	108	74	113	82	31	57	91	76.2	26.11
18	A(h)	30	34	26	45	32	27	53	38	27	31	37	34.5	8.34
19	A(s)mean	2.4	0.2	0.5	5.8	7.5	3.6	1.0	6.4	16.2	2.9	3.0	4.5	4.56
20	A(f)mean	1.8	0.7	0.0	1.5	1.6	0.1	0.8	0.3	0.6	4.0	3.0	1.5	1.26
21	A(h)mean	0.03	0.3	0	0.9	0.3	0.7	0.1	0.9	0.6	0.1	0.4	0.4	0.34

60^0-150mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.3333	0.45	0.45	0.2333	0.3833	0.2666	0.25	0.6667	0.2333	0.35	0.2333	0.35	0.14
2	Tv(f)	1.25	0.8834	0.3667	1.2	0.8666	0.0833	1.0167	1.2333	0.1333	0.0333	0.95	0.73	0.24
3	Tv(h)	0.8667	0.35	1.3	0.8	0.85	0.35	1.0167	0.7667	1.3333	1.1167	0.55	0.85	0.17
4	Ta(s)	0.2	0.3	0.3	0.1166	0.25	0.1333	0.1333	0.1167	0.1	0.2167	0.1166	0.18	0.08
5	Ta(f)	0.2	0.4	0.2	0.1833	0.1666	0.3333	0.1833	0.7	0.0667	1.25	0.8666	0.41	0.37
6	Ta(h)	0.65	0.8667	0.9333	0.6333	0.2166	0.2833	0.95	0.6833	1.25	1.0333	0.4	0.72	0.16
7	ROM(s)	67	77	84	80	81	68	69	70	85	77	82	76.4	6.73
8	ROM(f)	9	7	21	17	21	15	9	11	6	16	8	12.7	5.5
9	ROM(h)	6	3	4	4	5	5	4	7	6	4	4	4.7	1.19
10	V(s)	106	124	111	115	165	103	108	88	114	106	120	114.5	19.25
11	V(f)	13	16	26	18	32	17	27	24	5	19	18	19.5	7.42
12	V(h)	9	7	12	10	11	5	10	13	10	8	5	9.1	2.63
13	V(s)mean	39.5	32.4	36.8	41.3	32.9	26.0	32.2	24.9	28.1	29.2	39.5	33.0	5.67
14	V(f)mean	2.6	0	10.6	2.6	11.0	0.8	1.2	3.4	1.2	0.8	2.5	3.3	3.83
15	V(h)mean	3.5	0.9	1.1	2.7	0.6	1.7	2.0	3.3	2.3	1.5	1.2	2.4	0.97
16	A(s)	469.	439	375	548	763	500	527	341	496	497	595	504.5	112.52
17	A(f)	59	72	112	72	97	71	106	104	31	39	82	76.8	26.81
18	A(h)	37	25	37	46	56	23	41	31	25	40	23	34.9	10.65
19	A(s)mean	0.6	0.4	0.4	6.3	0.6	2.6	1.5	5.6	17.6	1.4	3.5	3.7	5.06
20	A(f)mean	0.05	0.6	0.1	4.9	0.2	3.3	0.7	1.0	1.4	7.2	6.0	2.3	2.6
21	A(h)mean	0.9	0.2	0.1	0.6	0.3	0.3	0.5	0.3	0.1	1.6	1.6	0.6	0.55

90^0-150mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.4	0.55	0.45	0.2833	0.4	0.35	0.2667	0.3167	0.4	0.25	0.5333	0.38	0.1
2	Tv(f)	1.0833	1.2167	0.7	1.1	0.85	0	1.1334	1.1667	0	0.0666	0.6	0.72	0.24
3	Tv(h)	0.8	0.9334	1.4833	0.95	1.0667	0.85	1.1167	0.8667	1.15	1.05	1.2333	1.05	0.1
4	Ta(s)	0.25	0.4	0.3	0.15	0.2833	0.2	0.1334	0.15	0.2667	0.1333	0.0833	0.21	0.1
5	Ta(f)	0.3833	0.7667	1.3	0.9333	0.6833	1.4167	0.15	0.2	0.5667	0.7833	0.55	0.7	0.4
6	Ta(h)	0.6833	1.25	0.8333	0.6	0.25	0.7333	1.0167	0.15	0.7333	0.95	0.4666	0.7	0.16
7	ROM(s)	66	80	83	80	85	72	70	75	74	78	78	76.5	5.7
8	ROM(f)	8	15	29	15	29	22	14	9	12	15	14	16.5	7.15
9	ROM(h)	6	2	5	4	5	5	4	9	7	5	3	5	1.90
10	V(s)	109	134	135	126	154	100	91	100	120	111	150	120.9	20.85
11	V(f)	15	17	28	19	54	20	26	22	14	22	25	23.8	10.95
12	V(h)	13	6	10	10	10	7	11	15	8	7	8	9.5	2.73
13	V(s)mean	32.5	28.5	36.4	41.9	38.0	29.6	21.4	30.7	26.0	30.7	26.1	32.3	5.91
14	V(f)mean	0.9	3.4	12.4	3	14.0	4.8	0.5	3.2	4.1	0.1	3.7	4.6	4.55
15	V(h)mean	3.1	0.8	0.6	3.1	0	1.3	1.7	4.6	2.6	1.5	0.3	2.7	1.42
16	A(s)	452	480	524	577	690	385	434	427	546	544	589	513.5	88.43
17	A(f)	63	74	101	80	182	81	133	104	50	53	113	94.0	38.88
18	A(h)	55	30	37	42	50	33	55	33	19	37	41	35.2	10.98
19	A(s)mean	1.3	1.5	0.4	4.7	0.6	1.7	1.0	3.4	0.8	2.3	6.8	2.2	1.99
20	A(f)mean	3.0	1.4	3.8	1.0	0.8	8.3	3.7	6.0	6.2	7.4	8.0	4.5	2.81
21	A(h)mean	0.2	0	0.4	0.6	0.7	1.0	0.5	0.3	0.4	1.0	1.4	0.6	0.41

0^0-500mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.2333	0.2667	0.45	0.2334	0.1667	0.4666	0.2834	0.3667	0.2167	0.2834	0.35	0.30	0.10
2	Tv(f)	0.8166	0.3667	0.1833	0	0.0333	0.95	0.0334	0.4	0.0667	1.4334	0.8	0.46	0.23
3	Tv(h)	0.3166	0.65	1.1333	0.3	0.5667	0	0.2167	1.65	0.9833	1.3667	1.05	0.75	0.27
4	Ta(s)	0.1166	0.1333	0.2833	0.1167	0.0833	0.1	0.15	0.2	0.1	0.1167	0.1166	0.14	0.06
5	Ta(f)	0.1666	0.2333	0.0833	0.2667	0.9833	0.4166	0.2834	0.6167	0.25	0.35	0.1	0.34	0.26
6	Ta(h)	0	0.2833	0.7666	0.7334	1.0833	1.3666	0.15	1.6167	0.95	0.1834	0.15	0.66	0.28
7	ROM(s)	43	52	77	67	66	43	67	50	62	39	49	55.9	12.44
8	ROM(f)	3	5	5	3	3	14	5	4	3	6	5	5.1	3.14
9	ROM(h)	1	1	1	2	1	0	3	1	2	2	2	1.5	0.82
10	V(s)	62	51	105	114	102	36	131	90	94	44	74	82.1	30.9
11	V(f)	10	10	10	14	8	5	10	5	10	5	10	8.8	2.82
12	V(h)	1	1	1	1	1	1	1	1	1	3	3	1.4	0.81
13	V(s)mean	19.5	19.7	23.4	33.4	31.1	13.2	29.2	17.0	22.4	9.6	21.2	21.8	7.31
14	V(f)mean	0.7	2.0	0.9	0.9	0.2	0	0.5	0.1	1.0	0.9	2.9	0.9	0.86
15	V(h)mean	0.01	0	0.1	0.1	0.4	0	0	0.1	0.1	0.7	0.5	0.2	0.24
16	A(s)	301	240	352	551	465	168	612	322	465	197	266	241.6	187.51
17	A(f)	54	43	51	57	31	31	33	11	32	31	31	36.8	13.34
18	A(h)	4	10	10	9	10	8	1	10	10	15	10	8.8	3.63
19	A(s)mean	1.3	4.7	0.5	7.6	22.9	4.7	1.4	3.7	5.0	3.8	4.6	5.5	6.12
20	A(f)mean	3.0	0.3	0.7	5.8	2.1	0.4	4.6	2.3	1.2	2.1	1.6	1.4	1.73
21	A(h)mean	0.3	0	0	0.3	0.5	0.7	0.6	0.8	0.5	0.7	1.1	0.15	0.33

30^0-500mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.2333	0.7167	0.4666	0.3666	0.1666	0.15	0.2666	0.3	0.3667	0.2667	0.4334	0.34	0.16
2	Tv(f)	0.9666	1.35	0.2166	0.8333	0.7666	0	1.1166	0	0	1.0833	0.85	0.65	0.25
3	Tv(h)	0.8333	1.15	1.3833	0.8	0.7666	1.35	0.2	0.5	0.5833	1.05	0.6167	0.84	0.18
4	Ta(s)	0.1166	0.1	0.3	0.1833	0.0833	0.0834	0.15	0.1166	0.1	0.1167	0.0834	0.13	0.06
5	Ta(f)	0.0833	1.2334	0.55	0.7333	0.6833	1.35	1.0333	0.5166	0.7833	0.7333	0.4834	0.74	0.36
6	Ta(h)	0.4833	0.0834	0.0833	0.6166	0.6833	0.4667	0.1333	0.3666	0.45	0.5333	0.55	0.40	0.10
7	ROM(s)	47	61	76	70	71	50	67	46	68	42	56	59.5	11.8
8	ROM(f)	4	3	5	11	8	14	8	8	8	10	11	8.2	3.28
9	ROM(h)	3	1	0	4	4	3	3	6	7	4	4	3.5	1.97
10	V(s)	53	64	104	135	128	46	131	65	65	46	82	83.5	34.81
11	V(f)	15	9	10	14	18	9	28	20	5	14	18	14.5	6.39
12	V(h)	16	2	1	12	7	4	3	11	8	6	6	6.9	4.59
13	V(s)mean	17.5	20.7	21.8	39.9	34.6	17.3	27.0	23.0	24.2	17.1	22.9	24.2	7.26
14	V(f)mean	0.5	1.0	0.2	2.8	3.1	1.0	0.2	3.4	0.7	1.4	1.2	1.3	1.23
15	V(h)mean	1.4	0.3	1.4	2.3	1.9	1.5	0.7	3.2	3.5	1.5	0.8	1.7	1.0
16	A(s)	261	239	363	547	528	182	642	279	268	209	230	376.2	159.2
17	A(f)	76	32	53	60	85	25	122	73	31	59	79	63.2	28.36
18	A(h)	16	10	10	45	20	18	18	50	5	16	32	21.8	14.48
19	A(s)mean	1.8	4.3	1.2	5.1	37	13.0	0.6	7.1	13.2	3.0	21.5	9.8	11.08
20	A(f)mean	1.3	1.7	1.1	2.1	2.2	6.8	1.4	0.8	0.7	2.9	0.3	1.9	1.78
21	A(h)mean	0.2	0.1	0.1	0	0.5	0.1	0.2	0.2	0.4	0.2	0.7	0.2	0.21

60^0-500mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.3	0.4334	0.35	0.3	0.2334	0.1666	0.2833	0.4834	0.2833	0.6	0.4334	0.35	0.12
2	Tv(f)	1.4167	1.4667	0.1667	0.9666	0.7667	0	1.1167	0.9834	0.55	0.0667	0.85	0.76	0.26
3	Tv(h)	0.6667	0.9667	1.0667	0.85	0.8167	0.85	1.15	0.6334	0.8	0.8	0.6167	0.84	0.08
4	Ta(s)	0.1667	0.2834	0.1834	0.1666	0.1167	0.0833	0.1667	0.1	0.1166	0.1834	0.0834	0.15	0.06
5	Ta(f)	0.1	1.35	0.0667	0.9	0.3834	1.3	1.0167	0.55	0.4666	0	0.4834	0.6	0.48
6	Ta(h)	0.5167	0.8667	0.95	0.5	0.7167	0.7333	1.05	0.25	0.4	0.7	0.55	0.66	0.12
7	ROM(s)	51	65	76	71	74	54	70	54	43	47	72	61.5	11.94
8	ROM(f)	5	10	19	14	17	16	13	8	4	13	9	11.6	4.86
9	ROM(h)	4	3	5	4	5	6	4	6	5	4	5	4.6	0.92
10	V(s)	78	86	115	150	145	74	135	85	68	57	82	97.7	32.71
11	V(f)	9	17	24	15	20	14	24	11	8	25	18	16.8	6.05
12	V(h)	8	8	11	8	7	8	10	13	10	7	6	8.7	2.05
13	V(s)mean	22.9	24.5	35.6	36.1	35.1	23.7	30.4	23.3	26.3	15.7	22.9	27.0	6.53
14	V(f)mean	0.5	2.7	6.1	5.9	9.5	2.6	1.2	0	1.2	3.3	1.2	3.1	2.92
15	V(h)mean	2.2	0.7	1.9	2.2	1.5	2.0	0.9	3.5	4.7	1.5	0.8	2.0	1.2
16	A(s)	356	296	480	695	700	354	618	337	292	140	230	408.9	189.0
17	A(f)	41	82	73	76	77	40	109	47	30	108	79	69.3	26.74
18	A(h)	36	36	33	25	26	38	48	36	45	34	32	35.4	6.89
19	A(s)mean	0.3	0.6	2.0	3.7	14.6	11.4	4.6	11.4	14.6	1.8	21.5	7.9	7.15
20	A(f)mean	0.1	0.7	5.8	4.4	4.0	6.3	0.4	4.4	5.2	6.2	0.3	3.4	2.54
21	A(h)mean	0.3	0.1	0.1	0.6	0.4	0.8	0.7	0	0.2	0.7	0.7	0.4	0.29

90^0-500mm Kinematic characteristics of the lumbar spine

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tv(s)	0.3	0.4167	0.4	0.4	0.3166	0.1666	0.35	0.5	0.2333	0.5666	0.3	0.36	0.12
2	Tv(f)	0.9333	0.15	0.2667	1.2	0.7833	0.0666	1.0167	1.1667	0.0667	0.0833	0.9333	0.61	0.43
3	Tv(h)	0.8333	1.4333	0.75	0.9167	0.8	0.2666	1.1	1.25	0.8333	0.75	0.8333	0.89	0.15
4	Ta(s)	0.1833	0.2833	0.25	0.2667	0.2	0.0833	0.2334	0.3834	0.1167	0.1	0.1833	0.21	0.09
5	Ta(f)	0.35	0.7667	0.15	0.7167	0.5833	0	0.9167	1.0667	0	0.0166	0.35	0.45	0.39
6	Ta(h)	0.4166	0.0333	0.6667	0.8	0.1833	0.2166	1.05	1.2	0.4833	0.6666	0.4166	0.56	0.18
7	ROM(s)	53	69	73	72	79	60	69	56	50	57	71	64.5	9.55
8	ROM(f)	6	19	23	14	29	24	12	13	4	18	14	16.0	7.56
9	ROM(h)	6	3	6	7	5	6	4	8	6	5	5	5.5	1.37
10	V(s)	105	108	121	139	153	106	114	114	123	63	105	113.7	22.67
11	V(f)	11	32	27	15	36	25	30	30	13	27	11	23.4	9.14
12	V(h)	9	5	11	15	8	8	8	8	11	8	9	9.1	2.55
13	V(s)mean	27.8	36.7	34.2	36.3	40.7	27.0	26.1	26.1	39.8	23.6	28.1	31.5	6.14
14	V(f)mean	1.4	9.1	11.1	7.0	14.8	6.5	3.3	3.3	4.4	7.3	1.4	6.3	4.17
15	V(h)mean	3.1	1.5	1.6	3.7	0.8	2.6	0.6	0.6	4.7	2.1	3.1	2.2	1.35
16	A(s)	493	411	526	596	681	503	493	493	616	197	493	500.2	124.92
17	A(f)	47	151	90	49	114	103	150	150	56	116	47	97.5	42.72
18	A(h)	34	25	44	52	32	24	36	36	42	39	34	36.2	8.08
19	A(s)mean	0.1	3.2	0.6	1.7	3.5	15.3	0.7	0.7	4.9	14.1	0.9	4.2	5.43
20	A(f)mean	0.7	3.2	2.3	0.2	6.7	0.8	0.2	0.2	5.1	4.7	0.7	2.9	2.84
21	A(h)mean	0.1	1.2	0.1	0.1	1.1	0.7	0.8	0.8	1.6	0.6	0.1	0.7	0.52

APPENDIX 3

KINETIC MEASUREMENTS OF ASYMMETRICAL LIFTING

0^{o}-150mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.22	1.33	1.66	1.19	1.56	1.44	1.49	1.44	1.42	1.36	1.17	1.39	0.15
2	Tvmax	0.42	0.52	0.55	0.39	0.41	0.43	0.56	0.50	0.50	0.42	0.33	0.46	0.07
3	Tv(x)	0.72	0.90	1.14	0.79	0.98	0.97	0.90	0.93	0.96	0.92	0.72	0.90	0.12
4	Tv(y)	0.27	0.63	1.11	0.85	1.13	1.00	1.03	0.93	1.02	0.95	0.71	0.88	0.25
5	Tv(z)	0.42	0.51	0.55	0.39	0.41	0.42	0.56	0.50	0.49	0.42	0.33	0.45	0.07
6	Tamax	0.60	0.71	0.45	0.58	0.14	0.24	0.67	0.27	0.28	0.59	0.54	0.46	0.20
7	Ta(x)	0.23	0.73	0.97	0.48	0.79	0.70	0.63	0.64	0.67	0.73	0.42	0.64	0.20
8	Ta(y)	0.18	0.48	0.94	0.70	0.75	0.61	0.68	0.66	0.85	0.76	0.52	0.65	0.20
9	Ta(z)	0.23	0.26	0.45	0.21	0.14	0.24	0.46	0.27	0.28	0.2	0.18	0.27	0.10
10	Tfmax	0.23	0.26	0.45	0.21	0.14	0.24	0.46	0.27	0.28	0.23	0.18	0.27	0.10
11	Tf(x)	0.23	0.73	0.97	0.48	0.79	0.70	0.63	0.64	0.67	0.73	0.42	0.64	0.20
12	Tf(y)	0.18	0.48	0.94	0.70	0.75	0.61	0.74	0.66	0.85	0.76	0.52	0.65	0.21
13	Tf(z)	0.23	0.26	0.45	0.21	0.14	0.24	0.46	0.27	0.28	0.23	0.18	0.27	0.10
14	Tpmax	0.31	0.42	0.46	0.31	0.33	0.34	0.46	0.38	0.28	0.33	0.25	0.35	0.07
15	Vmax	2.29	1.89	2.15	2.40	1.43	1.05	2.11	1.82	1.60	2.09	2.34	1.98	0.32
16	V(x)	1.02	1.08	1.02	1.14	0.79	0.94	0.94	0.90	0.90	1.19	1.11	1.00	0.12
17	V(y)	0.12	0.11	0.31	0.23	0.17	0.28	0.27	0.12	0.13	0.13	0.11	0.18	0.08
18	V(z)	2.29	1.89	2.13	2.39	1.41	1.63	2.08	1.80	1.58	2.07	2.33	1.96	0.33
19	Vmean	0.05	0.91	0.74	1.09	0.73	0.86	0.81	0.79	0.80	0.95	1.04	0.80	0.27
20	V(x)m	0.31	0.37	0.26	0.32	0.31	0.31	0.24	0.33	0.27	0.28	0.35	0.30	0.04
21	V(y)m	0.00	0.02	0.01	0.02	0.02	0.02	0.00	0.00	0.05	0.04	0.02	0.02	0.02
22	V(z)m	0.59	0.54	0.43	0.60	0.46	0.51	0.48	0.50	0.50	0.53	0.61	0.52	0.06
23	Amax	14.23	7.52	10.75	10.04	5.07	5.81	8.84	6.28	5.30	8.61	9.68	8.38	2.77
24	A(x)	7.74	3.46	3.78	4.36	1.62	2.71	3.46	2.10	3.07	4.47	3.73	3.68	1.60
25	A(y)	0.62	0.94	2.50	1.46	1.01	1.29	1.93	0.93	0.94	0.98	1.13	1.25	0.54
26	A(z)	7.74	5.21	10.67	9.37	5.02	5.71	5.46	6.23	5.24	7.15	9.58	7.03	2.03
27	Amean	5.72	3.95	3.31	5.76	2.51	3.36	3.59	3.34	3.27	4.37	5.37	4.05	1.11
28	A(x)m	0.05	0.04	0.01	0.03	0.02	0.07	0.03	0.01	0.01	0.08	0.04	0.04	0.02
39	A(y)m	0.01	0.01	0.00	0.00	0.03	0.01	0.00	0.00	0.00	0.01	0.00	0..01	0.01
30	A(z)m	0.05	0.04	0.01	0.06	0.07	0.06	0.02	0.02	0.03	0.05	0.07	0.04	0.02
31	Fmax	191.82	150.24	205.23	192.17	148.46	155.58	154.75	160.58	150.71	169.8	194.14	170.32	21.32
32	F(x)	77.4	34.6	37.8	43.6	16.2	27.1	34.6	21.0	30.7	44.7	37.3	36.82	16.04
33	F(y)	6.2	9.4	25.0	14.6	10.1	12.9	19.3	9.3	9.4	9.8	11.3	12.48	5.39
34	F(z)	175.5	150.2	204.8	191.8	148.3	155.2	152.7	160.4	150.5	169.6	193.9	168.45	20.29
35	Fmean	115.82	100.94	100.32	104.82	98.47	99.65	101.21	99.45	100.06	102.31	103.51	102.42	4.82
36	F(x)m	0.54	0.41	0.05	0.30	0.18	0.74	0.26	0.14	0.13	0.76	0.40	0.36	0.24
37	F(y)m	0.13	0.07	0.02	0.00	0.28	0.11	0.04	0.00	0.03	0.12	0.06	0.08	0.08
38	F(z)m	97.56	97.69	97.96	97.48	97.42	97.54	97.84	97.88	97.83	97.61	97.40	97.66	0.20
39	Pmax	320.62	228.20	275.93	347.01	158.93	201.06	264.50	233.75	186.32	267.45	334.69	255.32	61.93
40	Pmean	108.42	86.18	71.60	107.74	69.51	82.05	79.11	74.57	75.90	94.18	101.56	86.44	14.34
41	Wk	133.36	115.49	119.57	129.29	109.14	118.97	118.67	108.13	108.53	129.03	119.84	119.09	8.67

30^{0}-150mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.41	1.41	1.52	1.16	1.23	1.62	1.51	1.11	1.46	1.30	1.15	1.35	0.17
2	Tvmax	0.42	0.45	0.20	0.39	0.20	0.45	0.15	0.39	0.48	0.60	0.36	0.40	0.12
3	Tv(x)	0.79	0.87	0.87	0.75	0.68	1.01	0.94	0.68	0.93	0.84	0.72	0.73	0.26
4	Tv(y)	0.74	0.39	0.20	0.72	0.75	0.81	0.96	0.31	0.87	0.57	0.57	0.63	0.24
5	Tv(z)	0.42	0.45	0.20	0.39	0.20	0.45	0.94	0.36	0.48	0.60	0.35	0.44	0.20
6	Tamax	0.69	0.64	0.10	0.59	0.10	0.35	0.68	0.53	0.70	0.50	0.56	0.49	0.22
7	Ta(x)	0.64	0.74	0.65	0.45	0.25	0.56	0.60	0.46	0.66	0.55	0.46	0.55	0.14
8	Ta(y)	0.57	0.19	0.10	0.43	0.31	0.16	0.74	0.11	0.66	0.50	0.46	0.38	0.23
9	Ta(z)	0.21	0.29	0.10	0.022	0.10	0.35	0.31	0.20	0.23	0.50	0.19	0.25	0.11
10	Tfmax	0.21	0.29	0.20	0.22	0.10	0.35	0.31	0.20	0.24	0.50	0.19	0.26	0.11
11	Tf(x)	0.64	0.74	0.65	0.45	0.25	0.56	0.60	0.46	0.66	0.55	0.46	0.55	0.14
12	Tf(y)	0.57	0.19	0.10	0.43	0.31	0.16	0.74	0.11	0.66	0.50	0.46	0.38	0.23
13	Tf(z)	0.21	0.29	0.20	0.25	0.10	0.36	0.31	0.20	0.21	0.50	0.19	0.26	0.11
14	Tpmax	0.34	0.38	0.20	0.31	0.20	0.36	0.40	0.30	0.39	0.50	0.27	0.33	0.09
15	Vmax	1.71	2.19	2.19	2.32	2.54	2.32	2.01	2.33	1.65	3.04	2.26	2.23	0.38
16	V(x)	1.16	1.01	0.95	1.13	0.91	0.94	1.06	1.10	0.97	0.88	1.17	1.03	0.10
17	V(y)	0.71	0.37	0.47	0.61	0.73	0.42	0.58	0.38	0.32	0.45	0.30	0.49	0.15
18	V(z)	1.70	2.16	2.10	2.31	2.47	2.32	1.98	2.29	1.65	2.96	2.25	2.20	0.36
19	Vmean	0.87	0.91	0.84	1.09	1.01	0.75	0.86	1.06	0.85	0.93	1.04	0.93	0.11
20	V(x)m	0.31	0.36	0.26	0.38	0.36	0.30	0.27	0.39	0.30	0.28	0.40	0.33	0.05
21	V(y)m	0.16	0.11	0.15	0.15	0.14	0.13	0.13	0.20	0.11	0.12	013	0.14	0.03
22	V(z)m	0.51	0.51	0.48	0.62	0.59	0.45	0.48	0.65	0.50	0.56	0.63	0.54	0.07
23	Amax	8.34	8.65	11.20	9.99	12.94	9.87	8.26	10.06	5.98	15.14	9.25	9.97	2.47
24	A(x)	4.44	2.83	2.89	3.23	3.48	2.89	4.05	4.12	3.11	3.98	3.48	3.41	0.65
25	A(y)	3.31	1.57	2.40	3.12	3.16	1.23	2.61	1.79	1.13	2.18	0.80	2.12	0.88
26	A(z)	6.15	7.32	10.74	8.66	12.57	9.86	6.13	9.18	5.58	14.7	9.13	9.09	2.83
27	Amean	3.99	4.06	3.20	5.69	4.46	3.18	3.81	5.50	3.50	5.42	5.23	4.31	1.00
28	A(x)m	0.00	0.01	0.07	0.05	0.08	0.00	0.02	0.04	0.01	0.14	0.02	0.04	0.04
39	A(y)m	0.01	0.00	0.06	0.01	0.12	0.04	0.00	0.04	0.01	0.05	0.03	0.03	0.04
30	A(z)m	0.03	0.03	0.27	0.06	0.48	0.05	0.02	0.07	0.03	0.07	0.07	0.11	0.14
31	Fmax	160.83	171.89	121.02	185.29	225.91	196.76	160.14	190.34	154.01	247.75	189.45	182.5	34.84
32	F(x)	44.4	28.3	28.9	32.3	34.8	28.9	40.5	41.2	31.1	39.8	34.8	35.0	5.67
33	F(y)	33.1	15.7	24.0	31.2	31.6	12.3	26.1	17.9	11.3	21.8	8.0	21.18	8.80
34	F(z)	159.6	171.3	120.01	184.7	223.8	196.7	159.9	189.9	153.9	245.1	103.21	173.42	41.44
35	Fmean	103.32	100.85	108.36	104.95	99.39	99.62	102.70	103.76	100.36	106.48	103.21	103.0	2.85
36	F(x)m	0.05	0.12	0.73	0.48	0.80	0.02	0.20	0.43	0.10	1.45	0.24	0.42	0.43
37	F(y)m	0.12	0.02	0.58	0.08	1.17	0.42	0.01	0.37	0.11	0.48	0.26	0.33	0.34
38	F(z)m	97.83	97.80	106.47	97.54	94.34	97.62	97.89	97.36	97.83	97.35	97.38	98.13	2.94
39	Pmax	212.25	282.97	265.1	316.06	333.0	324.95	244.34	332.91	192.14	644.04	320.72	342.59	205.49
40	Pmean	84.01	87.49	93.53	106.11	99.99	71.99	85.46	102.35	80.38	97.49	100.15	91.72	10.63
41	Wk	119.30	124.23	143.11	124.15	123.99	117.34	129.90	114.63	118.16	127.71	116.17	123.52	8.13

60^0-150mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.32	1.44	1.28	1.10	1.40	1.13	1.50	1.27	1.57	1.42	1.26	1.35	0.15
2	Tvmax	0.46	0.34	0.20	0.24	0.47	0.20	0.49	0.44	0.56	0.67	0.48	0.41	0.15
3	Tv(x)	0.78	0.83	0.66	0.70	0.80	0.56	0.91	0.76	0.99	0.95	0.71	0.79	0.13
4	Tv(y)	0.72	0.36	0.20	0.62	0.85	0.20	0.85	0.56	1.48	0.67	0.48	0.64	0.36
5	Tv(z)	0.42	0.33	0.20	0.24	0.47	0.20	0.49	0.40	0.54	0.67	0.48	0.40	0.15
6	Tamax	0.59	0.10	0.10	0.24	0.37	0.10	0.66	0.57	0.24	0.57	0.59	0.38	0.23
7	Ta(x)	0.55	0.70	0.35	0.36	0.44	0.10	0.64	0.48	0.67	0.68	0.39	0.49	0.18
8	Ta(y)	0.48	0.11	0.10	0.34	0.37	0.10	0.73	0.11	0.53	0.57	0.13	0.32	0.23
9	Ta(z)	0.22	0.10	0.10	0.24	0.37	0.10	0.31	0.21	0.24	0.57	0.22	0.24	0.14
10	Tfmax	0.22	0.10	0.1	0.24	0.37	0.10	0.31	0.21	0.24	0.57	0.22	0.24	0.14
11	Tf(x)	0.55	0.70	0.35	0.36	0.44	0.10	0.64	0.48	0.67	0.68	0.39	0.49	0.18
12	Tf(y)	0.48	0.11	0.10	0.34	0.37	0.10	0.73	0.11	0.53	0.57	0.13	0.32	0.23
13	Tf(z)	0.22	0.10	0.10	0.24	0.37	0.10	0.31	0.21	0.24	0.57	0.22	0.24	0.14
14	Tpmax	0.35	0.20	0.10	0.24	0.37	0.10	0.40	0.34	0.39	0.57	0.38	0.31	0.14
15	Vmax	2.02	2.04	3.01	8.91	2.81	3.47	2.02	2.21	1.57	3.15	2.49	3.06	0.90
16	V(x)	1.03	1.33	0.97	1.15	0.83	1.08	0.99	1.11	1.07	1.38	1.15	1.10	0.16
17	V(y)	0.84	0.55	0.87	0.89	0.049	0.92	0.60	0.59	0.01	0.74	0.61	0.65	0.26
18	V(z)	1.97	1.97	2.88	8.91	2.78	3.30	2.0	2.09	1.53	3.05	2.33	2.48	0.80
19	Vmean	0.98	0.98	0.98	4.67	0.85	1.15	0.88	1.00	0.94	0.98	1.03	0.98	0.07
20	V(x)m	0.44	0.46	0.40	0.55	0.36	0.057	0.37	0.44	0.39	0.42	0.46	0.44	0.07
21	V(y)m	0.29	0.20	0.29	0.34	0.23	0.33	0.25	0.32	0.39	0.22	0.27	0.28	0.06
22	V(z)m	0.55	0.50	0.58	0.07	0.53	0.64	0.49	0.57	0.46	0.51	0.58	0.56	0.15
23	Amax	7.94	9.67	15.23	9.39	14.10	17.59	7.43	8.67	5.15	15.8	10.78	11.07	4.0
24	A(x)	2.64	3.60	3.4	3.01	3.98	3.02	2.62	3.19	2.23	4.86	3.27	3.26	0.72
25	A(y)	3.14	2.25	4.43	3.98	1.38	4.70	2.11	2.51	0.81	3.71	1.78	2.80	1.28
26	A(z)	6.85	9.27	14.56	8.91	14.02	16.68	7.00	6.76	5.09	15.0	8.64	10.28	4.07
27	Amean	4.35	4.12	4.44	5.90	4.19	5.48	3.60	4.48	3.11	6.13	4.95	4.61	0.93
28	A(x)m	0.02	0.04	0.06	0.03	0.05	0.07	0.02	0.00	0.01	0.04	0.00	0.03	0.02
39	A(y)m	0.01	0.04	0.09	0.01	0.01	0.21	0.00	0.06	0.01	0.03	0.03	0.05	0.06
30	A(z)m	0.05	0.17	0.49	0.05	0.07	0.82	0.02	0.07	0.03	0.04	0.04	0.17	0.25
31	Fmax	166.65	192.77	247.76	189.56	238.78	270.73	168.12	166.44	149.20	254.18	185.38	202.69	42.25
32	F(x)	26.4	36.0	34.0	30.1	39.8	30.2	26.2	31.9	22.3	48.6	32.7	32.56	7.21
33	F(y)	31.4	22.5	44.3	39.8	13.8	47.0	21.1	25.1	8.1	37.1	17.8	28.0	12.84
34	F(z)	166.6	190.8	243.7	187.2	238.3	264.9	168.1	165.7	149.0	251.1	184.5	200.9	40.73
35	Fmean	101.83	100.4	97.48	103.32	101.36	97.46	100.6	101.22	99.67	111.46	102.31	103.2	3.76
36	F(x)m	0.21	0.44	0.64	0.26	0.48	0.72	0.20	0.01	0.15	0.43	0.02	0.32	0.24
37	F(y)m	0.09	0.36	0.93	0.09	0.10	2.09	0.04	0.59	0.08	0.26	0.25	0.44	0.61
38	F(z)m	97.65	96.36	93.23	97.55	97.35	89.93	97.92	97.40	97.80	97.73	97.69	96.42	2.54
39	Pmax	253.55	290.0	565.14	1689.7	541.93	762.33	262.18	276.38	174.15	762.87	348.91	475.22	448.85
40	Pmean	91.83	93.58	96.11	455.16	84.50	114.40	83.69	94.45	85.7	109.51	96.47	127.74	109.01
41	Wk	122.13	135.68	123.99	505.22	119.15	130.42	126.36	120.9	135.41	156.6	122.52	163.49	113.84

90^0-150mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.37	1.70	1.36	1.21	1.24	1.12	1.37	1.09	1.57	1.00	1.55	1.39	0.22
2	Tvmax	0.41	0.39	0.37	0.46	0.20	0.20	0.28	0.20	0.56	0.20	0.20	0.32	0.13
3	Tv(x)	0.70	1.05	0.27	0.72	0.20	0.20	0.73	0.20	0.99	020	0.20	0.50	0.34
4	Tv(y)	0.72	0.61	0.42	0.62	0.20	0.20	0.65	0.41	1.48	0.20	0.20	0.52	0.38
5	Tv(z)	0.41	0.20	0.32	0.45	0.20	0.20	0.28	0.20	0.54	0.20	0.20	0.29	0.12
6	Tamax	0.31	0.10	0.26	0.63	0.10	0.10	0.10	0.10	0.24	0.10	0.10	0.19	0.16
7	Ta(x)	0.31	0.87	0.17	0.26	0.10	0.10	0.10	0.10	0.67	0.10	0.10	0.26	0.27
8	Ta(y)	0.52	0.15	0.37	0.39	0.10	0.10	0.18	0.10	0.53	0.10	0.10	0.24	0.18
9	Ta(z)	0.31	0.10	0.26	0.27	0.20	0.10	0.10	0.10	0.24	0.10	0.10	0.17	0.09
10	Tfmax	0.31	0.10	0.26	0.28	0.10	0.10	0.10	0.10	0.24	0.10	0.10	0.16	0.09
11	Tf(x)	0.31	0.87	0.17	0.26	0.10	0.10	0.10	0.10	0.67	0.10	0.10	0.26	0.27
12	Tf(y)	0.52	0.15	0.37	0.39	0.10	0.10	0.18	0.10	0.53	0.10	0.10	0.24	0.18
13	Tf(z)	0.31	0.10	0.26	0.27	0.20	0.10	0.10	0.10	0.24	0.10	0.10	0.17	0.09
14	Tpmax	0.32	0.20	0.37	0.37	0.20	0.10	0.19	0.10	0.39	0.10	0.11	0.22	0.12
15	Vmax	2.71	1.79	2.05	2.78	2.89	4.15	2.25	3.63	1.57	4.38	3.09	3.13	0.92
16	V(x)	1.26	1.19	0.89	1.32	1.15	1.64	1.08	1.51	1.07	1.35	1.06	1.23	0.20
17	V(y)	1.02	0.52	0.60	0.98	0.60	1.64	0.64	0.88	0.01	1.13	1.26	0.84	0.44
18	V(z)	2.53	1.73	1.94	2.47	2.58	2.44	2.17	3.20	1.53	4.01	2.61	2.56	0.75
19	Vmean	1.03	0.88	0.97	1.21	1.05	1.23	1.02	1.27	0.94	1.45	0.97	1.09	0.17
20	V(x)m	0.59	0.45	0.56	0.69	0.60	0.71	0.60	0.74	0.39	0.82	0.54	0.61	0.13
21	V(y)m	0.31	0.18	0.26	0.36	0.32	0.38	0.25	0.41	0.39	0.33	0.32	0.31	0.07
22	V(z)m	0.53	0.45	0.53	0.60	0.61	0.65	0.53	0.66	0.46	0.73	0.47	0.57	0.09
23	Amax	12.9	8.90	8.90	11.37	6.82	20.92	11.20	18.46	5.15	21.97	15.66	12.93	5.65
24	A(x)	4.21	3.13	4.94	4.02	5.85	8.28	4.63	7.70	2.23	6.84	5.38	5.20	1.86
25	A(y)	4.03	2.50	3.23	3.81	3.04	8.25	1.90	4.15	0.81	5.67	6.37	3.92	2.13
26	A(z)	12.19	8.78	8.74	9.10	2.58	17.35	10.19	16.26	5.09	20.09	13.25	11.24	5.26
27	Amean	4.81	3.17	3.82	5.99	2.18	6.38	3.90	5.91	3.11	7.90	3.88	4.64	1.72
28	A(x)m	0.01	0.01	0.02	0.00	0.20	0.43	0.14	0.33	0.01	0.30	0.16	0.15	0.15
29	A(y)m	0.01	0.05	0.02	0.02	0.13	0.43	0.03	0.14	0.01	0.35	0.15	0.12	0.14
30	A(z)m	0.01	0.18	0.08	0.03	0.61	0.89	0.19	0.60	0.03	1.38	0.31	0.39	0.43
31	Fmax	224.0	186.5	186.3	195.0	133.2	295.7	205.3	275.0	149.2	311.9	245.2	209.1	56.0
32	F(x)	42.1	31.3	49.4	40.2	58.5	82.8	46.3	77.0	22.3	68.4	53.8	52.0	18.6
33	F(y)	40.3	25.0	32.3	38.1	30.4	82.5	19.0	41.5	8.1	56.7	63.7	39.78	21.2
34	F(z)	220.0	185.9	185.5	189.1	123.9	271.6	200.0	260.7	149.0	299.3	230.6	210.5	52.6
35	Fmean	103.4	98.5	99.6	105.7	106.7	107.7	98.7	98.4	99.7	118.6	98.4	103.2	6.23
36	F(x)m	0.1	0.06	0.15	0.03	1.97	4.27	1.38	3.31	0.15	2.99	1.59	1.45	1.52
37	F(y)m	0.13	0.45	0.16	0.17	1.33	4.33	0.35	1.44	0.08	3.52	1.54	1.23	1.46
38	F(z)m	97.96	96.31	97.25	97.84	104.18	89.16	96.16	92.11	97.8	84.33	95.04	95.29	5.22
39	Pmax	371.3	236.4	328.9	399.3	361.8	1035.7	363.3	667.9	174.1	1268.9	468.4	516.0	342.6
40	Pmean	96.27	83.96	93.58	112.57	117.8	154.2	96.61	118.77	85.70	201.71	94.71	114.18	35.31
41	Wk	132.86	143.58	128.20	137.34	147.25	174.25	133.32	130.65	135.41	203.78	147.75	172.5	22.83

0^{0}-500mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.18	1.14	1.43	0.97	1.25	1.15	1.07	1.09	1.80	1.15	0.98	1.20	0.24
2	Tvmax	0.32	0.33	0.50	0.32	0.32	0.39	0.32	0.30	0.52	0.29	0.28	0.35	0.08
3	Tv(x)	0.63	0.72	0.96	0.58	0.80	0.71	0.68	0.59	1.13	0.67	0.56	0.73	0.17
4	Tv(y)	0.23	0.75	0.05	0.22	0.38	0.57	0.60	0.64	0.81	0.34	0.22	0.44	0.25
5	Tv(z)	0.32	0.33	0.53	0.31	0.32	0.37	0.32	0.30	0.51	0.29	0.28	0.35	0.09
6	Tamax	0.51	0.64	0.37	0.49	0.17	0.55	0.17	0.45	0.31	0.45	0.43	0.41	0.15
7	Ta(x)	0.40	0.41	0.61	0.38	0.62	0.37	0.43	0.45	0.13	0.51	0.39	0.39	0.15
8	Ta(y)	0.12	0.60	0.37	0.91	0.27	0.32	0.50	0.38	0.59	0.67	0.12	0.44	0.24
9	Ta(z)	0.17	0.15	0.37	0.16	0.17	0.18	0.17	0.19	0.31	0.17	0.15	0.20	0.07
10	Tfmax	0.17	0.16	0.37	0.16	0.17	0.18	0.17	0.19	0.31	0.17	0.15	0.20	0.07
11	Tf(x)	0.40	0.41	0.61	0.38	0.62	0.37	0.43	0.45	0.13	0.51	0.39	0.43	0.13
12	Tf(y)	0.12	0.60	0.37	0.96	0.27	0.32	0.50	0.38	0.59	0.67	0.12	0.45	0.25
13	Tf(z)	0.17	0.15	0.37	0.16	0.17	0.18	0.17	0.19	0.31	0.17	0.15	0.20	0.07
14	Tpmax	0.26	0.27	0.48	0.25	0.25	0.30	0.26	0.25	0.43	0.24	0.23	0.29	0.08
15	Vmax	1.77	1.35	1.46	2.09	1.40	1.49	1.56	1.92	1.65	1.76	1.98	1.68	0.25
16	V(x)	1.07	0.99	0.86	1.05	0.92	0.86	1.07	0.95	0.85	1.05	1.20	0.99	0.11
17	V(y)	0.16	0.04	0.15	0.09	0.08	0.08	0.11	0.07	0.39	0.16	0.05	0.13	0.09
18	V(z)	1.76	1.34	1.44	2.08	1.40	1.48	1.53	1.92	1.55	1.72	1.97	1.65	0.25
19	Vmean	0.79	0.75	0.65	1.00	0.68	0.74	0.82	0.80	0.86	0.76	0.90	0.80	0.10
20	V(x)m	0.29	0.27	0.20	0.37	0.25	0.29	0.30	0.28	0.49	0.26	0.34	0.30	0.08
21	V(y)m	0.01	0.01	0.02	0.02	0.01	0.00	0.01	0.02	0.19	0.02	0.01	0.03	0.05
22	V(z)m	0.40	0.75	0.33	0.48	0.37	0.41	0.44	0.43	0.40	0.41	0.48	0.45	0.11
23	Amax	8.48	6.31	7.28	11.15	6.19	6.41	6.76	9.37	4.58	8.43	10.42	7.76	2.00
24	A(x)	4.56	3.73	3.25	4.62	3.05	3.53	4.65	4.08	2.38	5.26	5.55	4.06	0.97
25	A(y)	0.72	0.48	2.00	0.65	0.67	0.46	0.93	1.03	1.08	1.53	0.25	0.80	0.47
26	A(z)	7.50	5.57	6.42	9.14	6.16	5.91	6.58	8.71	4.56	8.07	9.14	7.07	1.54
27	Amean	4.48	3.91	3.17	6.13	3.28	3.74	4.41	4.50	2.69	4.18	5.47	4.18	1.00
28	A(x)m	0.00	0.01	0.08	0.06	0.05	0.05	0.04	0.09	0.04	0.05	0.05	0.05	0.03
39	A(y)m	0.01	0.00	0.03	0.01	0.00	0.01	0.04	0.01	0.01	0.00	0.00	0.01	0.01
30	A(z)m	0.06	0.06	0.05	0.12	0.05	0.08	0.07	0.07	0.02	0.05	0.10	0.06	0.03
31	Fmax	173.44	154.43	165.91	189.58	159.83	157.27	164.65	185.52	143.75	179.72	189.75	169.44	15.36
32	F(x)	45.6	37.3	32.5	46.2	30.5	35.3	46.5	40.8	23.8	52.6	55.5	40.6	9.73
33	F(y)	7.2	4.8	20.0	6.5	6.7	4.6	9.3	10.3	10.8	15.3	2.5	8.91	5.10
34	F(z)	173.1	153.8	162.3	189.5	159.7	157.2	163.9	185.2	143.7	178.8	189.5	168.79	15.44
35	Fmean	102.38	100.96	100.03	105.15	99.88	99.59	102.61	102.39	98.75	103.00	106.22	101.91	2.35
36	F(x)m	0.04	0.10	0.84	0.64	0.48	0.53	0.42	0.94	0.44	0.47	0.48	0.49	0.27
37	F(y)m	0.06	0.05	0.33	0.08	0.01	0.12	0.36	0.08	0.10	0.01	0.05	0.11	0.12
38	F(z)m	97.52	97.54	97.59	96.85	97.55	97.28	97.40	97.42	97.89	97.62	97.14	97.44	0.27
39	Pmax	240.15	167.92	181.56	298.25	181.63	182.80	201.54	265.28	193.41	239.73	280.43	221.16	45.50
40	Pmean	75.18	70.10	62.39	96.36	65.88	69.42	76.64	78.05	80.34	74.85	89.65	76.26	9.96
41	Wk	89.46	80.61	89.84	94.42	83.01	80.53	84.93	85.87	145.41	86.83	88.75	91.79	18.26

30^0-500mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.48	1.39	1.33	0.93	1.30	1.14	1.33	1.13	1.16	1.10	1.07	1.21	0.16
2	Tvmax	0.40	0.43	0.45	0.30	0.36	0.39	0.33	0.31	0.37	0.28	0.31	0.36	0.06
3	Tv(x)	0.65	0.90	0.83	0.55	0.66	0.66	0.78	0.57	0.61	0.68	0.56	0.68	0.11
4	Tv(y)	0.65	0.51	0.49	0.48	0.63	0.46	0.76	0.46	0.62	0.22	0.57	0.53	0.14
5	Tv(z)	0.36	0.43	0.45	0.30	0.32	0.35	0.33	0.31	0.36	0.28	0.29	0.34	0.05
6	Tamax	0.55	0.18	0.73	0.45	0.18	0.57	0.48	0.18	0.20	0.14	0.46	0.43	0.23
7	Ta(x)	0.42	0.48	0.61	0.34	0.31	0.34	0.47	0.34	0.40	0.41	0.39	0.41	0.09
8	Ta(y)	0.47	0.22	0.23	0.35	0.45	0.12	0.51	0.11	0.50	0.11	0.10	0.29	0.17
9	Ta(z)	0.20	0.19	0.25	0.16	0.15	0.20	0.18	0.17	0.20	0.15	0.16	0.18	0.03
10	Tfmax	0.21	0.19	0.25	0.16	0.15	0.20	0.18	0.18	0.20	0.14	0.16	0.19	0.04
11	Tf(x)	0.42	0.48	0.61	0.34	0.31	0.34	0.47	0.34	0.40	0.41	0.39	0.41	0.09
12	Tf(y)	0.47	0.22	0.23	0.35	0.45	0.12	0.51	0.11	0.50	0.11	0.10	0.29	0.17
13	Tf(z)	0.20	0.19	0.25	0.16	0.15	0.20	0.18	0.18	0.20	0.15	0.16	0.18	0.03
14	Tpmax	0.30	0.29	0.42	0.24	0.26	0.30	0.28	0.25	0.30	0.22	0.23	0.28	0.05
15	Vmax	1.69	1.14	1.39	2.25	1.55	1.63	1.57	2.00	1.72	1.62	1.98	1.69	0.30
16	V(x)	1.01	0.84	0.88	1.26	1.00	0.98	1.13	1.14	0.93	1.28	1.22	1.06	0.15
17	V(y)	0.48	0.23	0.32	0.54	0.56	0.30	0.36	0.40	0.36	0.29	0.33	0.38	0.11
18	V(z)	1.68	1.12	1.34	2.21	1.46	1.52	1.55	1.93	1.69	1.56	1.95	1.64	0.31
19	Vmean	0.70	0.68	0.71	1.07	0.74	0.80	0.78	0.87	0.87	0.93	0.94	0.83	0.12
20	V(x)m	0.28	0.23	0.31	0.51	0.36	0.41	0.33	0.45	0.41	0.39	0.46	0.38	0.08
21	V(y)m	0.13	0.12	0.16	0.18	0.15	0.17	0.12	0.20	0.15	0.17	0.17	0.16	0.03
22	V(z)m	0.32	0.34	0.36	0.51	0.36	0.42	0.36	0.42	0.41	0.43	0.44	0.40	0.05
23	Amax	7.73	5.66	5.74	11.95	6.42	6.55	6.64	8.86	6.85	7.40	9.78	7.60	1.91
24	A(x)	5.63	3.64	2.38	5.26	3.95	3.98	3.73	4.21	3.80	5.29	4.44	4.21	0.92
25	A(y)	1.65	1.06	0.92	2.86	2.42	0.90	1.73	1.33	1.39	1.39	1.14	1.53	0.62
26	A(z)	6.87	4.42	4.65	10.14	6.36	6.24	6.44	8.59	6.82	7.10	8.87	6.95	1.71
27	Amean	3.60	3.04	3.21	6.66	3.61	3.97	3.65	4.62	4.35	4.70	5.17	4.23	1.04
28	A(x)m	0.01	0.01	0.03	0.04	0.00	0.03	0.02	0.02	0.03	0.09	0.04	0.03	0.02
29	A(y)m	0.00	0.00	0.00	0.02	0.01	0.01	0.00	0.02	0.00	0.02	0.03	0.01	0.06
30	A(z)m	0.03	0.03	0.02	0.10	0.05	0.05	0.04	0.07	0.06	0.08	0.08	0.01	0.03
31	Fmax	166.83	143.44	145.02	199.89	161.90	160.87	162.85	185.27	166.43	170.28	187.11	168.17	17.13
32	F(x)	56.3	36.4	23.8	52.6	39.5	39.8	37.3	42.1	38.0	52.9	44.4	42.1	9.24
33	F(y)	16.5	10.6	9.2	28.6	24.2	9.0	17.3	13.3	13.9	13.9	11.4	15.26	6.19
34	F(z)	166.8	142.3	144.6	199.5	161.7	160.5	162.5	184.0	166.1	169.1	186.8	166.74	17.70
35	Fmean	101.54	99.83	99.59	109.02	100.75	100.22	101.62	101.86	100.33	103.94	103.65	102.07	2.69
36	F(x)m	0.15	0.05	0.33	0.36	0.01	0.28	0.25	0.19	0.33	0.93	0.38	0.30	0.24
37	F(y)m	0.00	0.00	0.04	0.17	0.06	0.09	0.04	0.16	0.05	0.15	0.26	0.09	0.08
38	F(z)m	97.77	97.83	97.94	97.01	97.60	97.63	97.71	97.37	97.53	97.34	92.26	97.09	1.62
39	Pmax	222.03	131.68	155.30	332.36	191.43	197.71	197.76	270.49	217.98	213.39	282.17	219.3	57.34
40	Pmean	66.1	63.74	65.78	104.46	69.06	73.29	74.81	80.60	79.13	89.95	88.95	77.81	12.56
41	Wk	98.49	89.24	88.14	98.19	90.47	84.28	100.25	91.88	92.58	99.85	96.06	93.58	5.33

60^{0}-500mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	1.19	1.47	1.42	1.09	1.20	1.48	1.24	1.19	1.11	1.24	1.07	1.25	0.15
2	Tvmax	0.38	0.34	0.45	0.32	0.40	0.37	0.32	0.34	0.36	0.45	0.31	0.37	0.05
3	Tv(x)	0.62	0.91	0.78	0.64	0.49	1.01	0.75	0.55	0.59	0.75	0.57	0.70	0.16
4	Tv(y)	0.61	0.74	0.43	0.56	0.63	0.39	0.65	0.42	0.60	0.30	0.26	0.51	0.16
5	Tv(z)	0.36	0.33	0.43	0.31	0.34	0.37	0.31	0.30	0.34	0.30	0.28	0.33	0.04
6	Tamax	0.51	0.18	0.10	0.46	0.21	0.18	0.19	0.17	0.23	0.19	0.16	0.23	0.13
7	Ta(x)	0.37	0.24	0.21	0.30	0.31	0.71	0.41	0.33	0.41	0.35	0.36	0.36	0.13
8	Ta(y)	0.44	0.16	0.11	0.27	0.47	0.10	0.10	0.11	0.14	0.19	0.12	0.20	0.14
9	Ta(z)	0.21	0.18	0.10	0.17	0.20	0.18	0.19	0.17	0.20	0.19	0.16	0.18	0.03
10	Tfmax	0.21	0.18	0.10	0.17	0.20	0.18	0.19	0.17	0.23	0.19	0.16	0.18	0.03
11	Tf(x)	0.37	0.26	0.21	0.30	0.31	0.71	0.41	0.33	0.41	0.35	0.36	0.37	0.13
12	Tf(y)	0.44	0.16	0.11	00.27	0.47	0.10	0.10	0.11	0.14	0.19	0.12	0.20	0.14
13	Tf(z)	0.21	0.18	0.10	0.17	0.20	0.18	0.19	0.17	0.20	0.19	0.16	0.18	0.03
14	Tpmax	0.28	0.28	0.35	0.24	0.31	0.29	0.28	0.26	0.30	0.20	0.23	0.28	0.04
15	Vmax	1.93	1.46	1.30	2.08	1.65	1.24	1.70	2.11	1.88	1.76	2.17	1.75	0.32
16	V(x)	1.14	1.02	1.01	1.09	0.99	1.07	1.13	1.23	1.16	1.29	1.27	1.13	0.10
17	V(y)	0.87	0.42	0.55	0.76	0.75	0.66	0.51	0.74	0.63	0.77	0.76	0.67	0.13
18	V(z)	1.84	1.42	1.07	1.97	1.45	1.06	1.67	1.85	1.74	1.58	1.96	1.60	0.32
19	Vmean	0.92	0.78	0.72	1.00	0.85	0.76	0.88	0.93	0.95	0.92	1.02	0.88	0.10
20	V(x)m	0.44	0.39	0.40	0.54	0.48	0.37	0.46	0.53	0.55	0.48	0.57	0.47	0.07
21	V(y)m	0.31	0.22	0.25	0.33	0.28	0.25	0.29	0.33	0.30	0.28	0.33	0.29	0.04
22	V(z)m	0.40	0.34	0.34	0.44	0.40	0.33	0.39	0.40	0.43	0.39	0.45	0.39	0.04
23	Amax	8.67	6.07	5.13	9.12	6.40	5.05	7.37	8.82	7.53	7.60	9.65	8.43	1.59
24	A(x)	4.75	2.23	2.22	2.97	3.75	2.93	3.04	4.32	3.71	4.20	4.53	3.51	0.89
25	A(y)	3.31	1.64	1.73	2.46	3.03	2.77	1.27	2.94	1.30	3.25	3.50	2.47	0.84
26	A(z)	7.4	5.67	4.84	8.7	5.99	4.46	7.28	8.17	7.11	6.87	9.01	6.86	1.49
27	Amean	4.77	3.07	2.67	5.05	3.88	2.85	3.87	4.66	4.62	4.20	5.34	4.08	0.91
28	A(x)m	0.02	0.01	0.03	0.02	0.02	0.06	0.04	0.01	0.01	0.02	0.00	0.02	0.02
39	A(y)m	0.01	0.00	0.02	0.01	0.01	0.05	0.02	0.05	0.02	0.02	0.04	0.02	0.02
30	A(z)m	0.05	0.04	0.07	0.07	0.05	0.04	0.06	0.06	0.06	0.05	0.08	0.06	0.01
31	Fmax	172.62	156.32	147.49	186.53	159.53	144.66	171.28	182.85	170.87	169.94	191.37	173.2	52.4
32	F(x)	47.5	22.3	22.2	29.7	37.5	29.3	30.4	43.2	37.1	42.0	45.3	35.14	8.92
33	F(y)	33.1	16.4	17.3	24.6	30.3	27.7	12.7	29.4	13.0	32.5	35.0	24.73	8.39
34	F(z)	172.1	154.8	146.5	185.1	158.0	142.7	170.9	179.8	169.2	166.8	188.2	166.74	14.92
35	Fmean	103.48	99.69	99.33	102.34	101.07	100.18	100.8	102.32	101.5	102.46	103.31	101.5	1.14
36	F(x)m	0.19	0.08	0.26	0.21	0.24	0.58	0.44	0.07	0.09	0.22	0.05	0.20	0.17
37	F(y)m	0.14	0.03	0.24	0.12	0.07	0.49	0.19	0.54	0.19	0.24	0.45	0.25	0.17
38	F(z)m	97.64	97.74	97.39	97.37	97.65	97.74	97.55	97.52	97.54	97.55	97.28	97.54	0.15
39	Pmax	248.8	181.51	129.26	285.33	192.65	144.48	213.81	285379	233.42	209.7	310.56	221.39	58.39
40	Pmean	84.57	73.84	68.23	91.54	78.12	72.82	82.03	85.19	85.41	87.96	95.34	82.19	8.47
41	Wk	101.49	109.28	96.14	100.7	94.53	108.49	102.54	102.66	95.66	109.95	102.96	102.18	5.40

90^0-500mm Kinetic Measurements of Asymmetrical Lifting

NO	VAR.	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	MEAN	S.D.
1	Tmax	2.04	1.55	1.38	1.0	1.25	1.08	1.04	1.30	1.14	1.06	1.04	1.26	0.31
2	Tvmax	1.22	0.17	0.43	0.31	0.46	0.20	0.20	0.42	0.48	0.44	0.20	0.29	0.36
3	Tv(x)	1.35	0.92	0.43	0.61	0.59	0.60	0.51	0.60	0.62	0.44	0.43	0.26	0.27
4	Tv(y)	1.41	0.46	0.43	0.61	0.63	0.20	0.50	0.42	0.56	0.44	0.20	0.26	0.27
5	Tv(z)	1.15	0.17	0.43	0.29	0.35	0.20	0.20	0.40	0.41	0.44	0.20	0.34	0.43
6	Tamax	1.32	0.07	0.33	0.12	0.24	0.10	0.10	0.16	0.21	0.34	0.10	0.29	0.36
7	Ta(x)	1.21	0.27	0.33	0.11	0.24	0.10	0.10	0.26	0.28	0.34	0.10	0.30	0.31
8	Ta(y)	1.28	0.07	0.33	0.46	0.01	0.10	0.25	0.12	0.14	0.34	0.10	0.33	0.29
9	Ta(z)	1.02	0.07	0.33	0.13	0.35	0.10	0.10	0.16	0.21	0.34	0.10	0.30	0.27
10	Tfmax	1.03	0.07	0.33	0.13	0.25	0.10	0.10	0.16	0.21	0.34	0.10	0.26	0.32
11	Tf(x)	1.61	0.27	0.11	0.24	0.10	0.10	0.26	0.28	0.34	0.10	0.10	0.34	0.27
12	Tf(y)	1.28	0.07	0.33	0.46	0.01	0.10	0.25	0.12	0.14	0.34	0.10	0.28	0.36
13	Tf(z)	1.02	0.07	0.33	0.13	0.35	0.10	0.10	0.16	0.21	0.34	0.10	0.26	0.27
14	Tpmax	1.12	0.17	0.34	0.22	0.35	0.10	0.10	0.30	0.38	0.34	0.20	0.33	0.28
15	Vmax	2.08	1.32	2.15	1.87	1.81	2.32	2.62	1.80	1.92	2.88	2.42	2.11	0.44
16	V(x)	1.41	1.13	1.03	1.47	1.27	1.13	1.33	1.22	1.26	1.58	1.31	1.29	0.16
17	V(y)	0.92	0.7	0.77	0.83	0.66	0.94	0.70	0.73	0.64	0.79	1.12	0.8	0.14
18	V(z)	1.84	1.18	1.72	1.61	1.40	1.82	2.32	1.37	1.43	2.27	1.84	1.54	0.36
19	Vmean	0.64	0.82	0.84	1.18	0.88	1.02	1.17	0.89	1.03	1.07	1.18	0.97	0.18
20	V(x)m	0.36	0.49	0.55	0.80	0.61	0.73	0.78	0.61	0.72	0.74	0.79	0.65	0.14
21	V(y)m	0.22	0.21	0.31	0.36	0.33	0.32	0.34	0.35	0.35	0.34	0.44	0.32	0.06
22	V(z)m	0.24	0.33	0.35	0.48	0.39	0.42	0.46	0.37	0.42	0.45	0.47	0.38	0.07
23	Amax	8.45	8.96	11.32	8.14	4.81	11.66	13.28	6.29	6.18	13.44	12.36	9.48	3.14
24	A(x)	3.91	4.07	5.25	3.88	4.54	5.43	5.93	3.42	3.34	7.76	5.70	4.83	1.33
25	A(y)	3.13	3.23	4.65	2.46	1.87	4.75	2.07	2.16	1.65	3.6	5.66	3.20	1.33
26	A(z)	7.62	7.30	8.89	7.19	1.4	9.16	11.73	4.97	5.08	10.37	9.4	7.56	2.90
27	Amean	3.31	2.91	3.31	5.38	2.42	4.24	5.03	3.45	4.23	5.62	5.26	4.11	1.10
28	A(x)m	0.01	0.07	0.03	0.01	0.05	0.21	0.18	0.04	0.01	0.02	0.14	0.07	0.07
39	A(y)m	0.03	0.02	0.02	0.03	0.03	0.18	0.05	0.04	0.01	0.00	0.020	0.05	0.07
30	A(z)m	0.02	0.06	0.01	0.11	0.39	0.46	0.54	0.03	0.05	0.06	0.23	0.18	0.20
31	Fmax	176.5	178.84	199.72	174.19	117.55	202.96	224.2	152.75	152.99	219.18	208.22	182.46	32.49
32	F(x)	41.8	40.7	52.5	38.8	45.4	54.3	59.3	34.2	33.4	77.6	57.0	48.64	13.14
33	F(y)	31.3	32.3	46.5	24.6	18.7	47.5	20.7	21.6	16.5	36.0	56.6	32.03	13.34
34	F(z)	174.3	171.1	187.0	170.0	112.1	189.7	215.4	147.8	148.9	201.8	192.1	193.66	28.98
35	Fmean	101.8	100.11	99.91	103.52	105.22	98.39	97.64	100.66	101.03	104.62	100.86	101.25	2.4
36	F(x)m	0.09	0.73	0.26	0.15	0.51	2.08	1.81	0.39	0.11	0.17	1.39	0.7	0.72
37	F(y)m	0.26	0.22	0.24	0.26	0.35	1.83	0.51	0.44	0.06	0.04	1.96	0.56	0.67
38	F(z)m	97.9	97.52	97.95	96.98	101.96	93.52	92.74	97.85	97.63	97.52	95.78	97.03	2.44
39	Pmax	263.94	183.71	283.88	248.18	209.32	369.52	411.89	202.45	214.51	482.47	348.77	292.6	97.59
40	Pmean	59.07	77.46	78.31	106.11	96.03	98.98	105.71	82.36	92.99	97.56	110.01	91.33	15.43
41	Wk	121.09	120.84	108.85	107.18	121.0	107.89	111.0	107.89	106.94	104.39	115.52	112.05	6.37