RELIABILITY OF ISOVELOCITY DYNAMOMETER TESTING OF FOUR MOVEMENTS OF THE SHOULDER AND SHOULDER GIRDLE IN PERSONS WITH SPINAL CORD INJURY

by

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Abstract

The purpose of this study was to examine the reliability of using an isovelocity dynamometer for the testing of four movements of the shoulder and shoulder girdle in participants having a spinal cord injury. Ten people with quadriplegia and ten people with paraplegia participated in the study. The movements being tested included shoulder girdle (scapular) elevation and depression, and shoulder (glenohumeral joint) flexion and adduction. Scapular movements were tested using a standardised reciprocal protocol at an angular velocity of 30°/s, while glenohumeral movements were tested using a standardised protocol at an angular velocity of 60°/s. Testing was conducted on two days separated by no less than 72 hours, but no more than one week. Data analysis showed high reliability for all tested movements with intraclass correlation values greater than 0.9, and r^2 values greater than 0.76. The protocol and the isovelocity dynamometer used is a reliable means to assess force generated in these movements by participants with spinal cord injuries.

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Introduction

Every year, approximately 900 Canadians suffer a spinal cord injury (SCI); of this number, about half become quadriplegic (CPA, 2000). The incidence of SCI in Canada (35/year/million) is slightly higher than that in the United States (32/year/million). However, due to the greater population in the U.S., approximately 7800 injuries occur each year in that country, with slightly more than half becoming quadriplegic (NSCIA, 2000). The above values do not include individuals who die immediately or soon after injury, individuals who experience a minor SCI but recover fully or have only minor neurological deficit, or those who suffer neurological problems secondary to trauma. The National Spinal Cord Injury Association estimates that an additional 4860 cases each year (20/year/million) die before reaching the hospital. Individuals with SCI who reach the hospital face a long and difficult rehabilitation process, as less than 1% of those affected make a full recovery (NSCIA, 2000). It is during this rehabilitation process that strength of the remaining upper limb musculature is (re)developed in individuals with SCI. Optimal strength development of the muscles of the shoulder and shoulder girdle is imperative, as these muscles are now used for functional activities (Reyes et al., 1995) and for enhancing cardiorespiratory abilities (Hoffman, 1986). To ensure that rehabilitation programs are effective in the process of strength development, reliable methods of testing strength in the shoulder and shoulder girdle muscles must be developed.

Level of Impairment

Injuries that occur higher in the spinal cord will affect more muscle groups and individual muscles, and will result in a loss of control of these muscles. Thus, an individual who experiences a high level spinal cord injury will suffer more functional deficit directly related to this loss of motor control than would an individual with an injury at a lower level. For example, a person who experiences a SCI that results in complete loss of muscular control below the L4 level likely will have control of the muscles about the knee and may be able to walk independently with the use of crutches or canes and supportive devices for the feet. If this same person were to experience complete loss of muscular control below the L2 level, control of the muscles about the knee would be lost and the person would likely require long leg braces and crutches for mobility. Because long leg braces are cumbersome to use, and require increased effort due to the loss of a greater amount of muscular control in the lower limb, the individual is more likely to use a wheelchair for mobility. If the injury occurs above the T10 level, sitting balance will be affected, as the trunk no longer has the support provided by the abdominal and lower back muscles. For injuries that occur in the cervical region of the spinal cord, function is lost dramatically with each increase in the level of injury. For example, complete loss of motor control below the C8 level results in diminished control of the intrinsic muscles of the hand, causing a decrease in power and precision grip strength and hand dexterity; complete loss of motor control below the C7 level impairs control of the extrinsic muscles of the hand, resulting in reduced ability in activities that require prehension. An injury that results in complete loss of motor control below the C6 level effectively renders the hand non-functional unless specialised equipment is provided. In spite of increasing difficulty with loss of function, individuals with no motor control below the C5 level can, after participating in an extensive rehabilitation program, use a manual wheelchair for mobility.

Functional Abilities

Mobility

The overall goal of any rehabilitation program is the productive and self-fulfilling reintegration of the injured person into society (Drayton-Hargrove & Reddy, 1986). Mobility is of paramount importance to individuals who sustain injuries that result in the loss of lower limb function, and success in this aspect of a rehabilitation program requires preparation of the upper limb to accept the tasks that were once carried out by the lower limbs. For example, getting out of bed and walking to the next room become transferring into a wheelchair and wheeling to the next room. This latter task requires great strength, flexibility, and co-ordination of the upper limbs in order to move the body from the mattress to the wheelchair. Once this is accomplished, the upper limbs provide the power to move the wheelchair. Thus, the rehabilitation process required to allow success in this task includes: ensuring appropriate range of motion is maintained at each joint in the upper limb to allow the required movements to occur; development of adequate strength in the necessary muscle groups to perform the task; and development of the appropriate sequencing of muscle activity to complete the task successfully. The rehabilitation program will include specific stretches to maintain range of motion, progressive resistance training regimens to develop muscular strength, practice of component parts of the skill to mastery, followed by combining the component parts to complete the required task.

The shoulder and shoulder girdle muscles receive a great deal of attention during the rehabilitation process of individuals with loss of lower limb function, as individuals with SCI become reliant on their upper extremities for mobility and activities of daily living (Reyes et al., 1995). Effectively, the shoulder performs a role in wheelchair mobility similar to that of the hip in ambulation. The power to move the body, through the use of a wheelchair, is provided by the muscles of the shoulder and shoulder girdle, with the humerus and the scapula serving as the bony links to the trunk. Additionally, a role of the muscles of the shoulder and shoulder girdle has been suggested that include trunk stabilisation during wheelchair activities (Harburn & Spaulding, 1986), and also use of the diaphragm for similar stabilisation purposes (Sinderby et al., 1992). While research in the area of muscle use and activation of muscles during wheelchair mobility has increased (and is discussed in the literature review section), most studies have used participants with paraplegia, or have very small numbers of participants. Consequently, little information could be found regarding this area as it relates to cervical level injuries.

Weight relief and transfers

Another important consideration for individuals who suffer SCI is the ability to perform weight relief manoeuvres and transfers. Frequent lifting of the body to relieve weight from the ischial tuberosities is essential to the prevention of decubitus ulcer formation over this region. Transfers form an integral part of a person's ability to move about, as described previously. Both of these activities require appropriate strength in the muscles that perform the action of shoulder adduction, and shoulder girdle depression, as strength in these muscles is critical to prevent potential impingement of the rotator cuff between the humeral head and the acromion during loading of the arm in these movements (Perry et al., 1996). Electromyographic studies in this area have yielded identification of muscle use and activation during movement, but similar to the mobility research, have provided little information as it relates to higher cervical injuries.

Testing of the muscles that cause movement of the shoulder and shoulder girdle is an important part of determining improvement in strength in response to progressive resistance training programs for these muscles. This is especially true for individuals who have experienced a cervical SCI. As described previously, cervical SCI can result in the loss of voluntary control of a large percentage of the upper limb musculature, leaving only a very small amount of available muscle tissue to participate in strength training. Appropriate training for the remaining muscles is vital to ensure that functions previously carried out by the lower limbs can be done successfully with the upper limbs.

Respiratory Function

It is well known that the muscles of the respiratory system are significantly affected by SCI, and when compared to persons without SCI, those injured suffer from significant decreases in inspiratory and expiratory lung volumes and peak expiratory flow rates (Ohry et al., 1975, Haas et al., 1985, Bluechardt et al., 1992, Hopman et al., 1997). It is becoming increasingly apparent that shoulder and shoulder girdle muscle strength has a significant impact on respiratory function. Well trained and athletic individuals with SCI demonstrate higher values for inspiratory and expiratory mouth pressures (Hopman et al., 1997), and for vital capacity (Hoffman, 1986). Further, Loveridge et al. (1994) demonstrated increases in cardiorespiratory fitness and significantly improved inspiratory mouth pressures using a wheelchair training program in the early phase of rehabilitation. Enhanced inspiratory ability can assist those with SCI in bronchial hygiene by ensuring an adequate volume of air enters the lungs during the inspiratory phase of a cough manoeuvre.

DeTroyer & Estenne (1991, p.359) stated expiratory muscle function "has traditionally been perceived as being totally lost in tetraplegia". The implication of this is the reduced ability to clear bronchial secretions by coughing and maintain bronchial hygiene in persons having sustained this type of injury. This is most likely a contributing factor to the increased morbidity and mortality as a result of pulmonary complications (Cheshire, 1964; Bellamy et al., 1973; DeVivo et al., 1993). Research with persons having quadriplegia has shown that some expiratory function is preserved, and has led to further investigations to identify the muscles responsible. DeTroyer et al. (1986a) determined that one of the muscles of the shoulder girdle, the clavicular head of pectoralis major, played a key role in expiration. The application of this research has been the development of the concept that if this muscle was strengthened, coughing ability (and therefore, bronchial hygiene) would be enhanced, resulting in fewer bronchopulmonary infections (DeTroyer & Estenne, 1991).

The importance of the new role that the muscles of the shoulder and shoulder girdle take on in SCI with regards to function, respiration, and bronchial hygiene, must be recognised and appropriately addressed in the rehabilitation process. This is especially vital for individuals suffering from cervical SCI, where impairments are more severe. It is imperative that rehabilitation programs incorporate effective muscle strengthening regimens in an effort to maximise shoulder and shoulder girdle muscle strength for these purposes. Davis et al. (1981) stated that by maximising the function of the remaining muscle mass, the wheelchair-confined individual is better able to cope with day to day living. Additionally, quality of life is improved, as tasks that are routine for the healthy person are brought within the compass of the disabled individual.

To ensure the effectiveness of these training programs, especially for the individual with cervical SCI, it is necessary to develop reliable and valid testing procedures of shoulder and shoulder girdle muscles.

Review of Literature

Wheelchair mobility

In propulsion of a wheelchair by persons with paraplegia, the scapula serves as a base upon which the humerus describes an arc of motion from a position of extension to a position of flexion; as a consequence of the support function provided by the scapula, some protraction and retraction of that bone does occur (Mulroy et al., 1996). The technique used by persons with quadriplegia (those without hand grip) is more of a pump action, with more elevation and depression of the scapula (Sanderson and Sommer, 1985). Persons without hand grip rely more on a high coefficient of friction between the protective glove on their hand and the pushrim of the wheelchair, and, because they do not grasp the rim, they do not use the same pattern of movement as is seen in wheelchair propulsion by persons with paraplegia (Sanderson and Sommer, 1985). Inman et al. (1944), Sheving and Pauly (1959) and Shevlin et al. (1969) found that the muscle responsible for shoulder flexion in able-bodied subjects is the clavicular head of pectoralis major.

Limited research has been conducted on the muscles responsible for shoulder flexion, and shoulder girdle elevation/depression, in persons with SCI. Harburn and Spaulding (1986) conducted a study using surface electromyography (EMG) to monitor the recruitment of motor units in large muscle groups during a standardised wheelchair ambulation test using three able-bodied, three paraplegic and three quadriplegic participants. Participants were asked to perform an 8 metre long, cadence-controlled wheeling movement, across a firm floor surface. EMG results were compared to videotaped recording of the wheeling test to determine phasing of the push and recovery cycles, and the muscular activity during these phases. Results from this study showed no between-group trends, and wide intersubject variability of muscle recruitment patterns. The authors did note a higher percentage of activation of the muscles used by the quadriplegic participants, and suggested that this may be due to a higher requirement for trunk stabilisation in this group. The authors also noted that the investigation was limited by the small number of subjects tested. Mulroy et al. (1996) also conducted a study using EMG, to identify the phasing and intensity of shoulder muscle activity of 17 paraplegic participants during wheelchair propulsion. These investigators used indwelling electrodes to examine the activity of deep muscles, including those of the rotator cuff, in addition to the more superficial muscles. Participants were asked to wheel at a comfortable pace on a wheelchair ergometer for two bouts of 40 seconds. EMG results were compared to videotaped recording of the wheeling activity to determine phasing and muscle activity. Results from this study showed a consistent pattern of muscular activity: the pectoralis major had the highest peak and average EMG intensity during the push phase, and the trapezius had a dominant role in the recovery phase. The Mulroy et al. study (1996) showed that the muscles used during the flexion movement in persons with paraplegia was similar to those used by able-bodied subjects. Because Harburn and Spaulding (1986) included only three subjects with quadriplegia in their study, little information regarding the muscles used in the flexion movement as it pertains to persons with quadriplegia was found. Mulroy et al. (1996) stated that subjects with paraplegia perform protraction of the scapula through use of the serratus anterior muscle, while retraction is carried out by the trapezius muscle. No information could be found regarding the muscles used to perform these movements in individuals with quadriplegia.

Weight relief

A number of studies have investigated functional activities related to wheelchair mobility and other actions brought about through the use of muscles of the shoulder and shoulder girdle. Anderson et al. (1984), and Reyes et al. (1995) each conducted a study in which the body was lifted off the wheelchair seat and identified the pectoralis major and latissimus dorsi as key muscles; only individuals with paraplegia were examined in these studies. Perry et al. (1996) used EMG to investigate muscle use during a transfer manoeuvre; they identified the key muscles involved in the movement as pectoralis major and latissimus dorsi, with contributions from other muscles of the shoulder and shoulder girdle. Again, only individuals with paraplegia were examined in this study.

Respiratory function

The effects of SCI on the muscles of respiration are well documented. Ohry et al. (1975) tested lung volumes and expiratory flow rates at admission, and again after 6 months. The study found some improvement over the time period, but significantly diminished volumes and flow rates persisted, especially in the quadriplegic group. Haas et al. (1985) further documented this finding and added more information regarding the time course of resolution, remaining deficiencies, and suggested contributory factors for these. They found a relatively rapid increase in lung capacities and inspiratory and expiratory airflow from the acute to post-acute phase. In the chronic phase, a more gradual increase in vital capacity and a gradual decrease in functional residual capacity was noted, with all other ventilatory indices unchanged. It should also be noted that the indices examined were all significantly different (abnormal) from predicted values based on normal subjects. These investigators suggested that early changes resulted from return of respiratory muscle function due to resolution of spinal inflammation and edema above the injury level. Later changes resulted from mechanical changes in chest wall compliance (changing from more compliant to less compliant), such that a stiffer chest wall led to more effective transduction of diaphragmatic displacement, but that expiratory intercostal and abdominal spasticity could limit maximum inspiration. Chen et al. (1990) examined the effect of posture (erect versus supine) on forced vital capacity (FVC) and forced expiratory volume (FEV) in subjects with SCI. They noted that FVC was related to the level of impairment, with quadriplegics being most affected. They also noted that the supine position was most favourable for quadriplegics, with FVC significantly increasing. They suggested this to be due to the effects of gravity on the abdominal contents pushing the diaphragm into a functionally higher position at the start of inspiration when supine. In upright positions the diaphragm is flatter due to the sagging of the abdominal wall, the excursion is reduced on inspiration, and smaller vital capacities are seen. Each of the above investigations noted the significant impact of SCI on respiratory abilities, but especially in quadriplegic subjects. They also noted that this group relies extensively on the diaphragm and the accessory muscles of the neck to perform respiratory movements. In a study that attempted to correlate pulmonary function test (PFT) values with level of injury, Bluechardt et al. (1992) found that lesion level was a good predictor of average PFT values, but was not a good predictor of the potential for individual improvement over the course of rehabilitation. They did note that the subjects who improved most markedly were those who were motivated to improve and engaged in endurance activities such as long distance wheelchair propulsion and arm ergometry. The effect of these kinds of activity on lung volumes and fitness will be discussed later.

Basmajian and De Luca (1985, p.408) stated (in normals), "The muscles usually considered to be the primary muscles of respiration are the diaphragm and the intercostals, but the following have also been implicated as either primary or accessory in respiratory function: the scalenes and the sternomastoid in the neck, the musculature of the shoulder region including pectoral muscles and serratus anterior, and the anterior abdominal muscles." Conceptually, any upper limb, shoulder girdle, and neck muscle which gains attachment to the chest wall may be used to assist in inspiration and/or expiration in people with quadriplegia, especially during exercise. This would be dependent on lesion level, limb/muscle orientation, and appropriate stabilisation of the attachments of the muscle.

Further EMG studies have indicated that the primary muscles of inspiration and their segmental innervation include: the diaphragm (C3,4,5), the scalenes (C4,5,6), and the intercostals (T2-6) (Campbell et al., 1970). Using EMG, DeTroyer et al. (1986b) identified accessory muscles of inspiration used by persons with high cervical SCI. They identified EMG activity in the following muscles during quiet spontaneous breathing, as well as during inspiratory movements that required greater effort: sternocleidomastoid (11th cranial nerve), trapezius (11th cranial nerve, C3, 4), mylohyoid, sternohyoid, and platysma (C1,2,3).

Normal expiration results from the elastic recoil of the lungs (Shaffer et al., 1995). It is during active expiratory manoeuvres, such as coughing, that additional muscle recruitment is required. Muscles of active expiration and their segmental innervation in normal or able bodied subjects include: the intercostals (T2-6), and the abdominals (T7-12) (Campbell et al., 1970). DeTroyer & Estenne, (1991, p.359) state that "expiratory muscle function, in fact, has traditionally been perceived as being totally lost in tetraplegia," and "is totally lost indeed in subjects with transection of the upper cervical cord (C1-C4)." Using EMG analysis, DeTroyer et al. (1986a) identified muscles used by persons with high cervical SCI to assist in active expiration as being the following: pectoralis major (clavicular head) (C5,6); latissimus dorsi (C6,7,8); teres major (C5,6,7). It was postulated by the authors that the latter two serve to stabilise the humerus to limit shortening of the pectoralis major. This facilitated the pull of the pectoralis major muscle on the manubrium, causing depression of the rib cage and assisting in active expiration.

The loss of innervation to the intercostal and accessory muscles results in lower reported values in each of the following: total lung capacity; forced vital capacity; expiratory reserve volume; forced inspiratory and expiratory manoeuvres (De Troyer and Heilporn, 1980, Huldtgren et al., 1980, McMichan et al., 1980, Haas et al., 1985, Clough et al., 1986, Mansel and Norman, 1990). These changes result in decreased ability of the person with SCI to take a deep breath and cough. The implication of these changes is a marked increase in morbidity due to pulmonary complications, including decreased clearing of bronchial secretions (Cheshire, 1964, Bellamy et al., 1973). Pneumonia is the leading cause of death for the person with cervical SCI (DeVivo et al., 1993).

Because most individuals who suffer a cervical SCI will experience respiratory complications such as pneumonia (DeVivo et al., 1993), enhanced ability in expiratory manoeuvres may reduce the risk of developing such complications. In a series of investigations, De Troyer et al. (1986a, 1986b), Estenne et al. (1989) and Estenne and De Troyer (1990) showed that subjects with quadriplegia actively contracted the clavicular portion of the pectoralis major muscle during coughing. From these results the investigators postulated that significant improvement in expiratory function as a result of training of this portion of the muscle could assist in more effective bronchial hygiene. Estenne et al. (1989) used horizontal flexion exercise to strengthen the clavicular head of pectoralis major in persons with quadriplegia. Training this muscle in this fashion is in contrast to the findings of Inman et al. (1944), Sheving and Pauly (1959) and Shevlin et al. (1969) who showed that in able-bodied subjects the clavicular head of pectoralis major was the muscle responsible for shoulder flexion. Based on the work of these investigators, the most effective method to train the clavicular portion of pectoralis major would be through shoulder flexion exercises. Thus, it is not clear whether shoulder flexion or horizontal flexion exercises are more effective as a training movement for persons with quadriplegia.

There is a growing amount of research indicating a link between upper limb muscular strength and endurance and respiratory function. Two review articles (Davis et al., 1981, Hoffman, 1986) compared prior studies relating to cardiorespiratory fitness. Davis et al. (1981, p.163) stated that "disabled athletes are superior to their non-athletic counterparts on measures of cardio-respiratory fitness and muscle function." Hoffman

(1986, p. 316) echoed this and stated, "When sedentary and athletic spinal cord injured subjects are compared, the athletic subjects tend to have higher values for vital capacity maximum ventilation during exercise, and 12 or 15 second maximum breathing capacity at rest." Loveridge et al. (1994) conducted a study using wheelchair training in the first six months of rehabilitation to assess the effects on multiple facets of respiratory and strength parameters. A preliminary analysis of data indicated that in early rehabilitation training maximum oxygen uptake increased by 27 per cent in both the participant groups, the paraplegic group within 2 months, and the quadriplegic group within 4 months. Also, significant increases in mean maximal inspiratory mouth pressures were found in the quadriplegic group between 2 and 6 months. Hopman et al. (1997) examined respiratory muscle strength and endurance in individuals with quadriplegia, and found a similar trend in maximum mouth pressures and training. Those individuals who participated in training activities for more than 2 hours per week were noted to have higher values in maximal inspiratory, expiratory pressures, and also in endurance indices.

The reliance of people with quadriplegia on the accessory respiratory muscles to assist the diaphragm in respiration, and the link between exercise training of the upper limb muscles and improvement in pulmonary function tests and cardiorespiratory fitness, underscores the need for appropriate shoulder and shoulder girdle testing and training methods in a rehabilitation program.

Shoulder Girdle (Scapular) motion

Elevation of the scapula, as defined by Kapandji (1982), is the upward movement of the lateral end of the clavicle towards the ear, having a range of 10 cm of movement vertically; depression of the scapula is the downward movement of the lateral end of the clavicle away from the ear, having a range of 3 cm of movement. However, the use of units such as centimetres for describing range of movement is problematic, as such range will be dependent on the length of the clavicle - a longer clavicle will have a greater vertical range. As clavicular length may vary in individuals, a more useful description of the motion is given by Rasch and Burke (1989): upward and downward translation of the scapula in the frontal plane. Rasch and Burke (1989) also noted that while movement of the shoulder girdle is defined by scapular excursion, all movements are constrained by the motion available at the sternoclavicular joint. Thus, they described the motion of scapular elevation and depression as a function of the sternoclavicular joint, having a range of 45° upward, and 5° downward, respectively. Kapandji (1982) noted that the axis of rotation for elevation of the scapula, mediated through the clavicle, is located superior to the costoclavicular ligament; the axis is oriented horizontally and slightly obliquely anteriorly and laterally. The muscle considered to be responsible for elevation of the scapula is the upper trapezius (Yamshon and Bierman, 1948, Wiedenbauer and Mortensen, 1952). The muscles considered to be responsible for depression of the scapula are: pectoralis minor through direct action from the thorax to the coracoid process (Moseley et al., 1992), and pectoralis major and latissimus dorsi through indirect action on the scapula via the humerus (Reyes et al., 1995).

Shoulder (Glenohumeral joint) motion

Flexion

Shoulder (glenohumeral) flexion is defined by Kapandji (1982) as anterior movement of the humerus away from the body in a sagittal plane about a frontal axis. The range of glenohumeral flexion is 180° (Warwick and Williams, 1973). Glenohumeral flexion is actually the resultant of movement of the head of the humerus on the glenoid fossa of the scapula, with some lateral movement of the inferior angle of the scapula over the thorax (Inman et al., 1944). Inman et al. (1944) found that the ratio of movement of the humerus in relation to the scapula is 2:1 - that is, 5° of scapular movement occurs for every 10° of humeral movement, commencing after approximately 30° of humeral flexion has occurred. The muscles responsible for glenohumeral flexion have been shown to be the clavicular head of pectoralis major and the anterior fibres of deltoid (Inman et al., 1944; Sheving & Pauly, 1959; Shevlin et al., 1969).

Adduction

Kapandji (1982) noted that shoulder (glenohumeral) adduction in the frontal plane, starting from any position of abduction, is always possible up to the position of reference (anatomical neutral). This is called relative adduction. It is mechanically impossible to attain greater adduction, in the frontal plane, due to the presence of the trunk, and further adduction can only be achieved when combined with glenohumeral extension or flexion. Thus, if his definition of abduction is accepted as being lateral movement of the humerus away from the body in a frontal plane about a sagittal axis, having a range of 180° (Warwick and Williams, 1973), then it can be stated that glenohumeral adduction from full abduction to anatomical neutral occurs in the frontal plane about a sagittal axis, and has a range of 180°. Similar to glenohumeral flexion, abduction/adduction is also the resultant of movement of the head of the humerus on the glenoid fossa, and lateral/medial movement of the inferior angle of the scapula over the thorax. Poppen and Walker (1976), describing abduction, noted that the contribution of scapular motion to this combined movement occurs after 30° of abduction, with a large lateral displacement of the inferior tip of the scapula occurring after 60°, resulting in upward positioning of the glenoid fossa. Bagg and Forrest (1988) further indicated that the majority of scapular contribution occurs after approximately 82° of abduction. The muscles responsible for glenohumeral adduction have been shown to be pectoralis major and latissimus dorsi (Reyes et al., 1995, Perry et al., 1996).

Muscle Testing Techniques

Different modes of testing have been used clinically to determine muscular performance. These include isometric, isotonic, and, most recently, isovelocity or isokinetic testing. The simplest of these is isometric testing, in which the clinician places the limb in a specified position and, without allowing the limb to move, "feels" the effort expended by the participant. This form of isometric testing is called manual muscle testing and is currently used by therapists to assess strength, using a classification scale ranging from zero (no contraction elicited) to grade 5 (normal strength) (Daniels and Worthingham, 1980). With respect to manual strength testing of shoulder girdle muscles, Frese et al. (1987) found that interrater reliability was low during testing for the trapezius muscle. This can be attributed to the subjective nature of the procedure; "normal" muscle strength is determined by each therapist based on patient anthropometry, age, and the therapist's experience. In an effort to decrease the subjective nature of this form of testing, myometers have been developed that have been found to be valid and reliable (Marino et al., 1982; Hyde et al., 1983a, 1983b). A myometer is a device that uses a strain gauge to sense force input applied to the sensing surface of the unit. The device is positioned on the limb of a subject, held in place by the therapist, and the subject is directed to attempt to move the limb. The therapist resists the movement, and the device records the force of the subject's muscular effort. Used in this fashion, these devices provide information regarding a subject's ability to generate a muscular force at one specific point in the entire joint range. Perrin (1993) argued that, because isometric testing is not a dynamic test method, it has limited application as it provides only a minimal amount of information regarding muscle performance over the entire joint range.

Another form of muscle testing is isotonic testing. Fleck and Kraemer (1997) stated that isotonic testing is traditionally defined as a muscular contraction in which the muscle exerts a constant tension while performing a movement. Fleck and Kraemer (1997) further explained that a major drawback in using this definition results from the arrangement of the lever systems in the body. Using the elbow joint as an example, the insertion of the brachialis muscle into the ulna creates a situation where the joint has more mechanical advantage in the mid range of flexion, and less mechanical advantage at the extremes of the joint range. Thus, the tension developed in the muscle will vary with the changes in mechanical advantage of the joint as the forearm moves through the range of motion. Consequently, the maximum amount of weight that can be moved through the entire range is that which can be moved through the weakest part of the range - the extremes. A more workable term to describe this type of testing and training, where the external resistance or weight does not change through the range of motion is dynamic constant external resistance (DCER) (Fleck & Kraemer, 1997). Dynamic constant external resistance testing involves apparatus to determine muscular performance, such as free weights or specially designed equipment using cables and pulleys. It is important to note that while muscle training devices have been developed to target specific muscle groups, the resistance provided by equipment that uses cables and pulleys does not change throughout the range of movement. Resistance provided through the use of free weights also does not change throughout the range of movement. Some equipment manufacturers (e.g. Nautilus®) have incorporated cams in the design of muscle training devices in an effort to match the mechanical advantage of the body's lever system. Fleck and Kraemer (1997) argue that while this type of equipment provides variable resistance through the range of motion, to date, matching of the body's mechanical advantage has not occurred. They further argue that matching the body's mechanical advantage for all individuals is not possible due to individual differences in limb length, tendon insertion points, and body size. Perrin (1993) argued that in spite of the dynamic nature of DCER testing, it provides a less than accurate representation of muscle performance through the entire range. In addition, there is no equipment available that is specifically designed to test shoulder elevation/depression. Free weights are not an option for individuals who lack hand grip.

Isovelocity testing (also called isokinetic testing) refers to a muscular action performed during a constant angular limb velocity, and requires the use of specialised equipment that has been specifically designed for this purpose (Perrin, 1993, Fleck & Kraemer, 1997). A function of the equipment design process has been the ability to control a number of parameters during the test procedure through the use of computer software. In this mode of testing, the angular velocity of the testing machine is precisely controlled, and upon reaching the desired test velocity, it does not change throughout the test movement. Consequently, the muscular force applied to the sensing apparatus is met with an equal and opposite reaction force. As a result, the muscle can exert a dynamic, continuous, maximal force throughout the movement's full test range (Perrin, 1993, Fleck & Kraemer, 1997). Perrin (1993) adds that this type of equipment has another distinct advantage over either of the previously described testing modes: the computerisation of the equipment sensing systems allow for very accurate recording of force exerted, and thus can provide greater information over the entire test range. Finally, the attachments that are provided with these machines are designed such that hand grip is not required for testing, and, with some modification, are suitable for shoulder elevation/depression testing.

Computer controlled isovelocity dynamometers are particularly suited for testing muscle performance and have been adopted as the definitive mode of testing since their introduction in the late 1960's (Dvir, 1995). There are many different systems available. The systems are comprised of electro/mechanical or hydraulic/mechanical devices that include a computer controlled feed back system.

The Kin-Com® 500H (Chattecx Corp., Hixon, TN) is an example of a hydraulic/mechanical system.

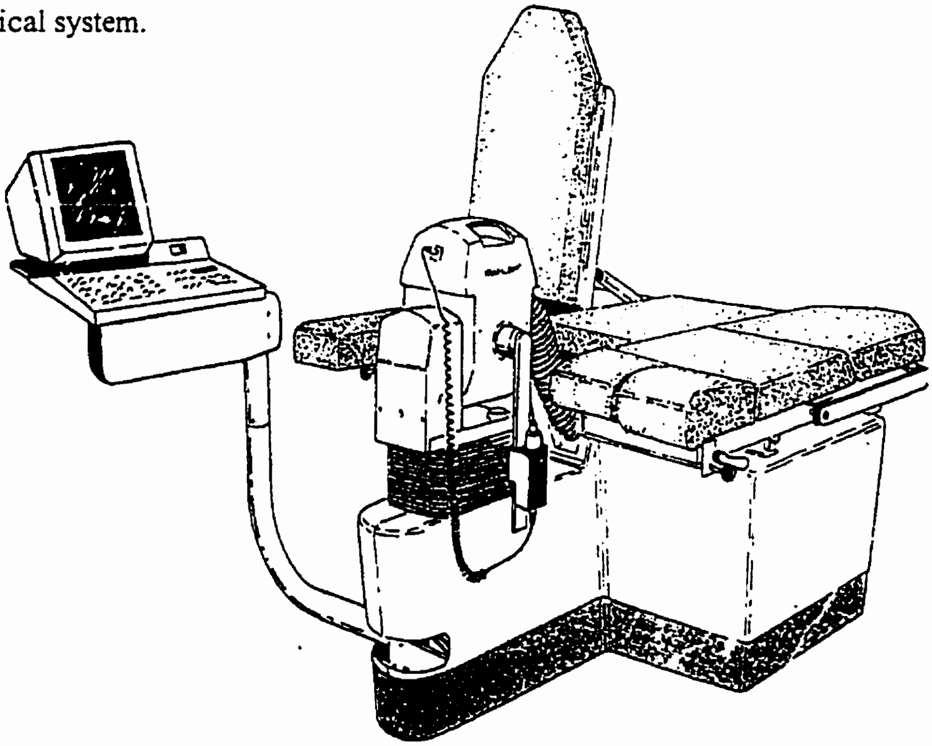


Figure 1 - Diagram of the KinCom® 500H Dynamometer.

The computer monitors input from multiple sensors and activates controlling mechanisms based on desired test parameters selected by the operator, in response to input from the sensors. The hydraulic/mechanical system is comprised of an actuator and actuator arm, a motor-pump unit, and a servo valve. The sensors are comprised of a load cell, which houses a strain gauge bridge to detect direction and amount of force applied to the load cell; a potentiometer that allows measurement of the exact position of the actuator arm; and a tachometer that measures the rotational speed of the actuator arm. The load cell is coupled to the actuator arm such that when a force is applied to the load cell the actuator arm is rotated by the hydraulic actuator at the velocity selected by the operator for the test. The actuator receives its power from the motor-pump unit via hydraulic lines; the

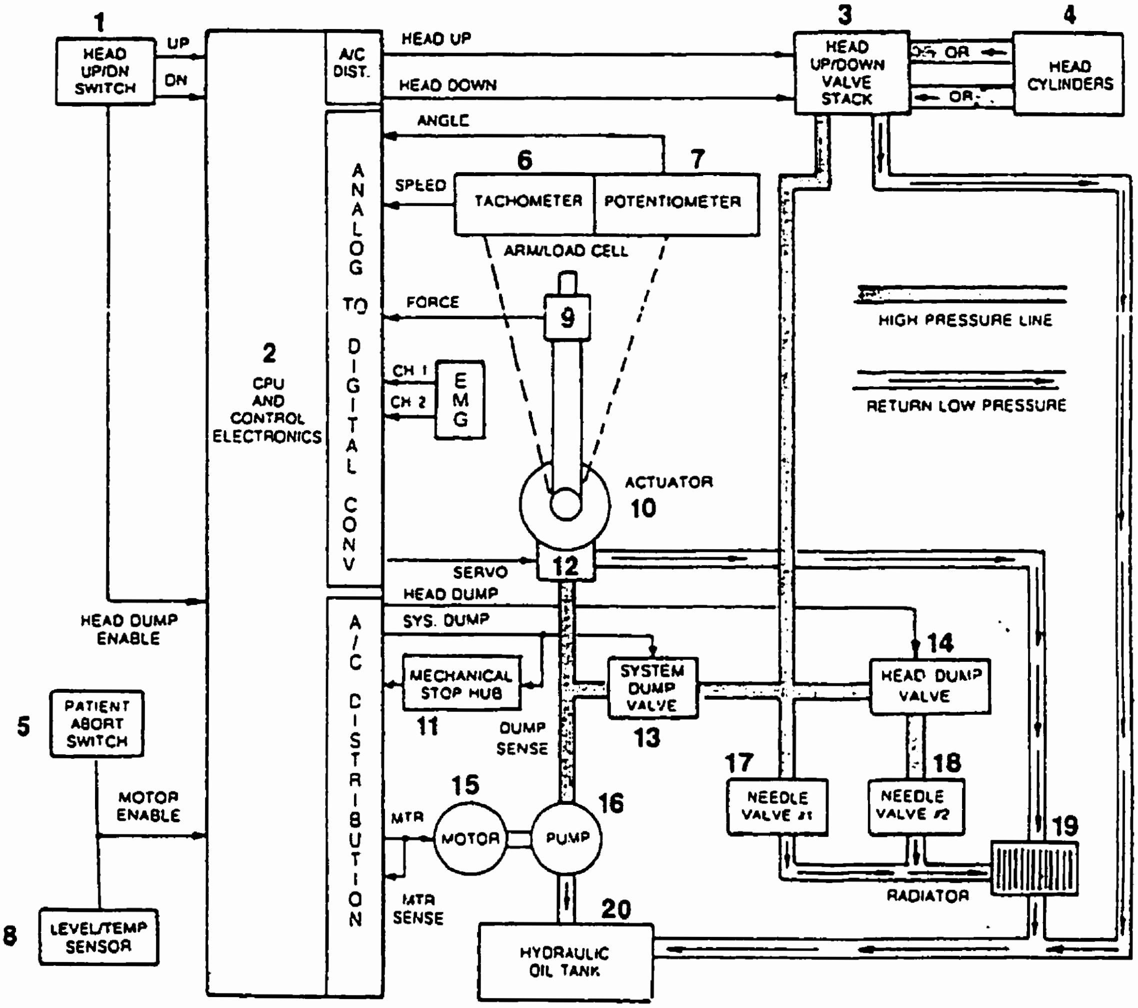


Figure 2 - Schematic diagram of the control system and feedback mechanism of the KinCom® 500H Dynamometer.

servo valve in the hydraulic power circuit controls the angular velocity and direction of motion of the actuator. An electrical signal from the computer controls the servo valve, and therefore ultimately controls the actuator arm motion. The feedback parameters to the computer are the level of force exerted by the participant (monitored by the load cell), the actuator arm position (monitored by the potentiometer), and the actuator arm velocity (monitored by the tachometer). With the above feedback mechanisms established, any change in force or velocity will initiate a response by the computer to adjust the servo valve accordingly. For example, in constant velocity mode, if a participant exerts greater force in the mid range of the movement being tested, the computer will adjust the servo valve to increase the hydraulic pressure to ensure that the test velocity selected by the operator is maintained throughout the test range. The computer controls the dynamometer and also collects, stores, and processes the data (Chattecx, 1992; Snow, 1991).

Computer controlled isovelocity dynamometers have been shown to be reliable and valid instruments (Farrell & Richards, 1986) for objective and accurate quantification of changes in strength during glenohumeral flexion/extension, ab/adduction, and internal/external rotation in able-bodied subjects (Perrin, 1986; Greenfield et al., 1990; Hellwig & Perrin, 1991). However, no studies were found that described reliable and reproducible techniques for testing scapular elevation/depression. Further, no studies were found that tested concentric muscle activity for the movement of glenohumeral flexion in persons with SCI.

The Kin-Com® 500H dynamometer was chosen for use in this study because of previously demonstrated reliability in testing shoulder movements, and ease of adaptability for testing shoulder girdle movements.

Reliability

Portney and Watkins (1993) conceptualised reliability as consistency, dependability, and predictability, and identified three main contributors to reliability as being the participant, the tester, and the measurement device. If a participant is consistent in their response to testing, and the tester dependable in taking measurements with a device that is also dependable, then it could be predicted that in the absence of any real change in the components, the results of two test sessions would be identical. This rarely, if ever, happens in real life. In an effort to minimise the amount of error that is inherent in any of the components of a measurement system, the expected sources of error can be identified, and strict procedures or protocols can be developed. In general, measurement errors can be attributed to the measuring instrument, the tester, or the variability of the characteristics being tested (Portney & Watkins, 1993).

The operation of the measurement device, the Kin-Com® Dynamometer, has been previously assessed to be valid and reliable. Farrell and Richards (1986) used external and independent devices to mirror the Kin-Com's data acquisition systems in both static and dynamic modes of operation. These devices included standardised weights to test the load sensing system, spirit levels and protractors to assess reference angle position measurements, and an external microcomputer controlled electronics system to monitor lever arm position, lever arm speed, and force applied to the load sensing system during dynamic system measurements. They concluded that the unit tested appeared to produce valid and reliable measurement of the condition of the lever arm and strain gauge systems. In addition, specific calibration procedures (Appendix A), and diagnostic tests were conducted as part of this study to ensure accuracy of the measurement device during the test period.

If the same tester is involved in both the test and the retest, a source of bias may be introduced as the tester may be influenced by their memory of the first test scoring (Portney & Watkins, 1993). Tester bias is reduced through the use of the computer software in the measurement system. Reporting of a participant's performance is through a computer generated image (a graphical pattern), and as such, does not have a "score". Because the pattern generated in the test is not immediately available during the retest, no bias can be introduced in this respect.

Test-retest intervals are an additional area of concern, and should be far enough apart to avoid fatigue, learning, or memory effects, but close enough to avoid genuine changes in the variable being assessed (Portney & Watkins, 1993). To address this threat to reliability, test sessions were scheduled to be no less than 72 hours apart, and no more than one week apart. This is consistent with previous isokinetic reliability studies (Perrin, 1986, Harding et al., 1988).

People participating in tests can improve significantly in retests simply as a result of their participation in the test;. in effect, learning how to improve their score for subsequent retests (Portney & Watkins, 1993). These practice or carry over effects can be reduced or neutralised through the use of familiarisation and practice sessions prior to the collection of data (Harding et al., 1988, Portney & Watkins, 1993). A specific protocol utilising a familiarisation and practice session was developed for this study based on Harding et al. (1988). In spite of conducting familiarisation and practice sessions as part of a specific protocol, due to the unpredictability of the environment and human responses to testing, it is doubtful that participants can produce identical results in each test session (Portney & Watkins, 1993). This problem is further compounded when the test protocol requires repeated trials of the same action during each test session to be used for data analysis. Trial one may be lower than trial two, and trial two may be higher than trial three. If only one of these scores was to be selected and reported, the tester is faced with the dilemma of selecting the most representative trial. An alternative to this dilemma is to report the mean of the three trials as being most representative, as theoretically, the mean of the three scores is the most representative score (Portney & Watkins, 1993).

Purpose

The purpose of this study was to determine the feasibility of using isovelocity dynamometry for testing the movements of shoulder girdle (scapular) elevation/depression, and shoulder (glenohumeral) flexion and adduction in individuals with SCI. These movements were chosen for testing because individuals with SCI use the muscles associated with scapular elevation/depression and glenohumeral flexion and adduction extensively for functional, respiratory, and bronchial hygiene activities.

Objectives

The primary objective of this study was to establish the reliability of four tests of strength using isovelocity dynamometry - scapular elevation, scapular depression, glenohumeral flexion, and glenohumeral adduction - in each of two participant groups - persons with quadriplegia and persons with paraplegia. Reliability was assessed through an analysis of total work measures. A secondary objective was to compare the strength parameters between the groups. A final objective was to conduct a subset analysis to investigate variables related specifically to available muscle mass (active versus non-active sternal head of pectoralis major), and recreational exercise participation.

Clinical Relevance

The establishment of a reliable testing mechanism for muscles of the shoulder and shoulder girdle will help to ensure that the outcome of upper limb training programs for individuals with SCI can be accurately evaluated. Such training programs are vital to maximise function in activities of daily living, respiratory function, general exercise performance, and to enhance quality of life for people with SCI. Furthermore, in higher SCI lesions, the gains during rehabilitation may appear smaller. It is important to have measurement tools which can accurately measure improvement in individual muscles because it is through this method that exercise training strategies can be modified to maximise gains in muscular strength and endurance in the most cost effective manner.

Hypotheses

For Objective 1:

Hypothesis 1 - When testing isovelocity scapular elevation/depression, there will be no difference in force production between two test periods separated by no less than 72 hours, but no more than 1 week.

Hypothesis 2 - When testing isovelocity glenohumeral flexion, there will be no difference in force production between two test periods separated by no less than 72 hours, but no more than 1 week.

Hypothesis 3 - When testing isovelocity glenohumeral adduction, there will be no difference in force production between two test periods separated by no less than 72 hours, but no more than 1 week.

For Objective 2:

Hypothesis 4 - There will be no difference in the test-retest coefficient between individuals with quadriplegia versus individuals with paraplegia.

For subset investigations:

Hypothesis 5 – There will be no difference in the reliability of force production between two test periods separated by no less than 72 hours, but no more than 1 week, in the group of participants with quadriplegia who also have an active sternal head of pectoralis major, when compared to the group of participants who do not have an active sternal head of pectoralis major.

Hypothesis 6 – There will be no difference in the reliability of force production between two test periods separated by no less than 72 hours, but no more than 1 week, in the group of participants who participate in recreational exercise more than two times per week, when compared to participants who participate in recreational exercise less than two times per week.

Limitations

The size of the participant groups was small as a result of the small pool of potential participants.

Delimitations

The test velocity was delimited to 30°/s for scapular elevation/depression, as it was found in pilot testing that reliable results were unobtainable at greater velocities. The test velocity was delimited to 60°/s for glenohumeral flexion and adduction, as this velocity has been used in previous studies of isovelocity testing of the glenohumeral joint (Ivey et al. (1985), Walsh et al. (1985), Perrin (1986), Pawlowski & Perrin (1989), Reid et al. (1989), Hellwig & Perrin (1991), Nicholas et al. (1992)). The test range was delimited to 25° for scapular elevation/depression, as the force sensing apparatus impinges upon the participant's ear with a larger test range. The test range was delimited to 90° for glenohumeral flexion, and 80° for glenohumeral adduction to limit the amount of scapular movement associated with greater movement ranges at this joint.

Assumptions

While injuries to the spinal cord are classified as being at a specific lesion level, individuals with the same level of injury may have varying muscular function. It was assumed that individuals with the same lesion level had similar abilities with regard to performing the tests of scapular elevation/depression, and glenohumeral flexion and adduction. It was also assumed that those who participated in this study were representative of the general SCI population regarding their abilities in the test movements.

For purposes of comparison between the group with quadriplegia and the group with paraplegia, it was further assumed that all individuals in the group with quadriplegia had similar abilities, and all individuals with paraplegia had similar abilities.

For purposes of the subset analysis, it was assumed that participants without an active sternal head of pectoralis major had similar abilities in the test movements, and those with an active sternal head of pectoralis major had similar abilities. Recreational exercise participation consisted of playing wheelchair basketball or rugby, or freewheeling on the street or on a wheelchair ergometer, for periods of greater than 40 minutes per session, more than two times per week. It was assumed that these different forms of training provided similar training effects. In the absence of formal cardiorespiratory fitness testing, it was assumed that this subset of participants had similar and higher levels of general physical fitness, as compared to those who did not participate in comparable recreational exercise activities.

Testing was conducted on the right upper limb only. It was assumed that limb strength was equivalent bilaterally.

Methodology

Participants

Twenty people participated in this study: 10 participants with traumatic quadriplegia as a result of SCI lesions between C5 and C8, and 10 participants with traumatic paraplegia as a result of SCI lesions below T11. One paraplegic participant withdrew from testing due to sustaining a shoulder injury between the test sessions. Below the C4 level was chosen because innervation of the clavicular head of pectoralis major stems from the C5 and C6 spinal segment level (Warwick and Williams, 1973). The group with paraplegia was included as they would have normal innervation to all shoulder and shoulder girdle muscles, and therefore, normal activation of these muscles for force generation. The group with quadriplegia would have varying innervation to the muscles of the shoulder and shoulder girdle based on lesion level, and therefore, reduced activation during force generation.

Inclusion Criteria

Participants were between the ages of 18 and 55, and in self reported good health. Previous studies used subjects within this age range when testing individuals with SCI (Harburn & Spaulding, 1986, Powers et al., 1994, Reyes et al., 1995, Mulroy et al. 1996, Perry et al., 1996). Participants had complete motor loss below the lesion level, but may have had incomplete sensory loss below the lesion level, and had experienced their injury more than one year prior to enrolment in this study. This time frame allowed participants to have reached a level of post injury rehabilitation in which they were considered stable in their recovery, and had been discharged from hospital. All participants used a manual wheelchair for mobility purposes.

Exclusion Criteria

Participants were excluded if they had any current shoulder pathology, or previous pathology that had been surgically corrected. Participants were to be pain free during activity without the use of analgesics.

Recruitment

A list of potential participants who met the above criteria was generated from information provided by the Outpatient Spinal Cord Injury Clinic at the Winnipeg Health Sciences Centre. Individuals were contacted by telephone, the study purpose and procedure were explained, and a request was made for participation. Upon agreement to participate in the study, appointment times were established for participants to attend a familiarisation session and the test sessions; participants were provided with directions to the test facility.

Informed Consent

Prior to the familiarisation session, the study purpose and procedure were explained a second time, and written informed consent was obtained (Appendix B). This study was approved by the University of Manitoba Faculty of Medicine Committee on the Use of Human Subjects in Research.

Study Design

A repeated measures design was used (Portney & Watkins, 1993); there were two test sessions.

Instrumentation

Isovelocity Dynamometer

The Kin-Com® 500H isovelocity dynamometer (Chattecx Corp., Hixon, TN) was used for this study. The instrument has been described previously. Calibration and diagnostic procedures were completed prior to the commencement of the study and a check was made for discrepancies upon completion of the study to ensure accuracy of recorded data. No discrepancies in calibration were discovered. The Kin-Com® system has a software package that includes a diagnostic mode. The diagnostic program was activated prior to commencement of the study to determine any identifiable areas of concern; none were found. The diagnostic mode was activated prior to each test session to allow the system to identify problems that may have resulted from internal component faults or failures. No internal faults or failures occurred during the testing period. Calibration of the hydraulic system, which is not user serviceable, was performed by a service technician upon initial set up of the system. No hydraulic problems were detected through use of the diagnostic software prior to commencement of or during the study.

The reliability of the Kin-Com® dynamometer has been comprehensively assessed and the device is reported to have less than 3% discrepancy from values obtained by external testing devices (Farrell and Richards, 1986).

Support and Linking Devices

All participants were required to wear a supportive splint during testing of glenohumeral flexion and adduction. The splint held the elbow in 90° flexion, and also supported the forearm in a neutral position. A splint was necessary because some of the participants with quadriplegia did not have complete control of their elbow movements, and most did not have control of wrist movement. With the elbow and forearm secured, comfort and safety for quadriplegic participants was ensured. To ensure that bias was not introduced into the procedure, this device was also worn by the participants who did have elbow and wrist control. The splint was constructed of a thermoplastic material, and was designed and manufactured by a certified prosthetist.

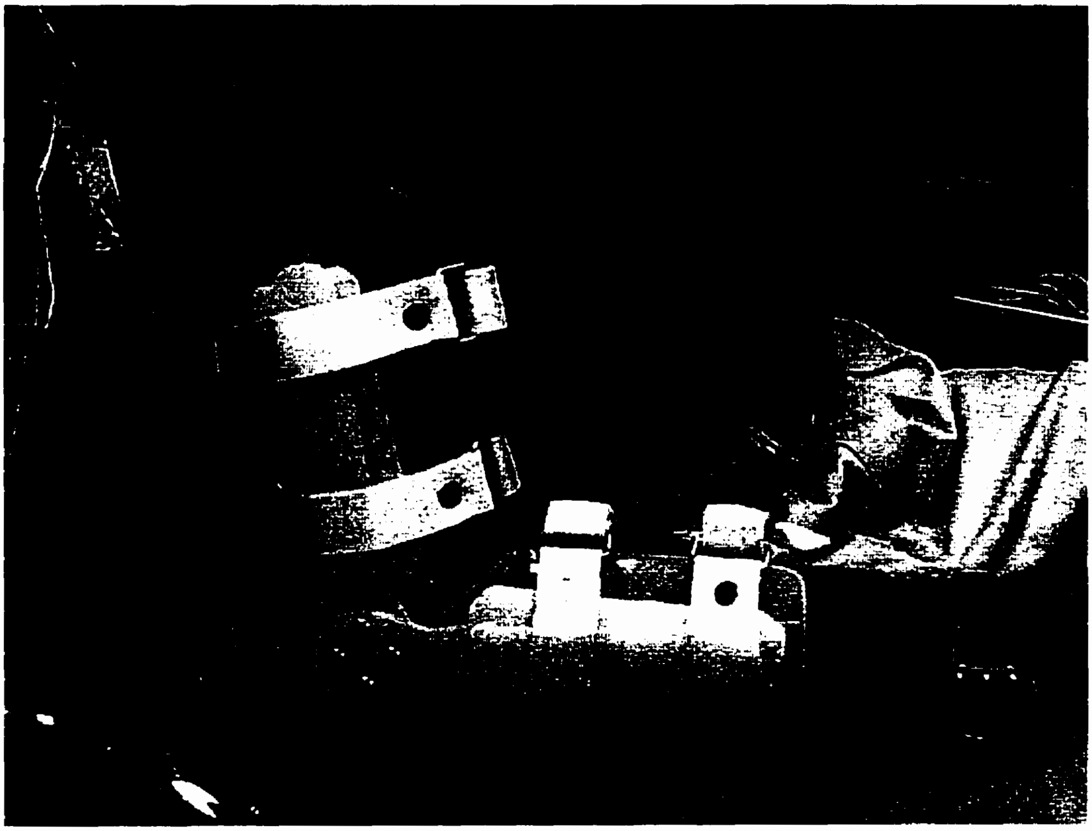


Figure 3 - Image of elbow splint.

A sling device was constructed in order to couple the upper limb to the force sensing pad of the dynamometer, specifically for the purpose of measuring scapular depression. The sling was constructed of heavy weight cotton; a 5 centimetre wide Velcro strap was used to fasten the sling to the force sensing pad. A tape measure was sewn into the Velcro strap to ensure consistent application and tension in each test session. The sling was constructed such that when applied, the Velcro strap passed under the olecranon process, and maintained the elbow in 90° flexion.

Figure 4 – Representative image of sling coupling upper limb to force sensing pad, in starting position for testing of shoulder girdle elevation and depression.

Procedure

A review of available patient records from the Health Sciences Centre Outpatient Spinal Cord Injury Clinic was conducted for selection of individuals who met the inclusion criteria. Demographic data were collected and accuracy was confirmed with the participants. These individuals were also questioned regarding the type of activities done during employment hours (active or sedentary), and the type and frequency of participation in recreational activities, as it was hypothesised that activity level might have an effect on the outcome of the results.

Participants were asked to attend a familiarisation session prior to the first test session; this included an introduction to the testing equipment and an opportunity to practice the movements that were to be tested. The first test session followed within 48 hours, with the second test session conducted not less than 72 hours, but not more than one week following the first test. The sequence of movement testing was randomised to ensure that a sequencing effect did not occur (Appendix C). All tests were conducted on the right upper extremity due to mechanical restrictions when using a wheelchair with the dynamometer. The axes of rotation of the dynamometer and the segment being tested were aligned prior to each test type.

Isovelocity was chosen as the testing mode because it allows the selected muscle group to be tested to its maximum potential throughout the entire available range of motion in a dynamic movement pattern (Perrin, 1993).

Landmark Selection

The spinous process of the second thoracic vertebra (T2) was selected as the bony landmark for actuator alignment during scapular elevation/depression as it most closely approximates the location of the axis of rotation of the shoulder girdle when it is moved through the test range. The only connection of the scapula to the axial skeleton occurs at the sternoclavicular joint. Moore (1985) stated that the superior aspect of the jugular notch lies at the level of the junction of the second and third thoracic (T3) vertebrae, and also indicated that the spinous process of T2 is nearly horizontal in its orientation. The sternoclavicular joint lies immediately lateral to the jugular notch, and the tip of the spinous process of T2 lies at the level of the intervertebral disc between the T2 and T3 vertebral bodies. Taking into consideration Kapandji's (1982) orientation of the axis of rotation of the sternoclavicular joint, the closest approximation for a stable, easily identifiable bony landmark on the posterior aspect of the trunk is the spinous process of T2. In addition, in order to test isolated scapular elevation/depression, a convenient position for the participant is seated comfortably in a chair, with the trunk strapped to the backrest. Securing the participant in this fashion greatly reduces the contribution of the trunk muscles in performing the scapular elevation/depression movement, as would be the case if testing were done in the standing position. When using the seated position, it is impossible to use the sternoclavicular joint as the axis of rotation, because the participant must sit with his/her back to the dynamometer in order to apply the force sensing mechanism to the shoulder girdle.

The landmark for the axis for glenohumeral flexion was located approximately 4 centimetres inferior to the lateral border of the acromion process, midway between the anterior and posterior borders of the deltoid muscle. This was the closest approximation to the axis of rotation identified by Kaltenborn (1980) and Kapandji (1982) as being midway between the anterior and posterior borders of the humeral head.

The landmark for the movement of glenohumeral adduction was located approximately 4 centimetres inferior to the lateral border of the acromion process, midway between the lateral border and the coracoid process. This was the closest approximation of the axis of rotation of the humerus, radiologically identified by Poppen and Walker, (1976), and described by Kapandji (1982), as lying in the centroid of the head of the humerus, inferior to the acromion process.

Testing Protocol

Scapular elevation/depression

Testing scapular elevation required the participants to be seated in their own wheelchair with their back to the dynamometer; the axis of rotation of the dynamometer's actuator arm was aligned with the T2 spinous process. In the event that the backrest height of the participant's wheelchair was too great to allow alignment with the T2 spinous process, the participant was transferred to a wheelchair with a lower seat height. The trunk was stabilised in the chair by a Velcro strap approximately 10 centimetres wide placed across the chest, immediately under the axillae, and secured to the wheelchair backrest at the highest level possible. The force sensing apparatus was placed so that the lateral edge of its padding was aligned with the lateral border of the acromion. The sling device was used to secure the arm to the force sensing apparatus and to maintain the elbow in 90° flexion. The participant was instructed to keep the elbow as close as possible to the side of the body throughout the movements of scapular elevation/depression.

The movement pattern for testing was explained to the participant. A standardised reciprocal movement testing protocol was used that included the same explanation and instructions, familiarisation period, warm-up period, and testing period for each participant (Appendix D). This protocol was based on that used by Harding et al. (1988) which showed high reliability coefficients for inter-repetition and inter-occasion measures. Angular velocity for this test was set to 30°/s, as it was determined through pilot testing of the scapular elevation movement carried out by a quadriplegic participant that reliable results would be produced at this velocity; higher velocities gave rise to unreliable results for scapular elevation. The range of motion for testing was from the neutral resting position (0°) to 25° elevation, and the reciprocal 25° returning from the elevated position to neutral. A limit of 25° elevation was necessary as the force sensing mechanism impinged on the participants' ear when a greater range was used.

Glenohumeral flexion and adduction

Testing for each of glenohumeral flexion and adduction required the participants to be seated in their own wheelchair. The trunk was stabilised in the wheelchair by a Velcro strap approximately 10 centimetres wide placed across the chest, immediately under the axillae, and secured to the wheelchair backrest at the highest level possible. For testing glenohumeral flexion, the axis of rotation described earlier was aligned with the axis of rotation of the dynamometer. The upper limb was positioned in 90° elbow flexion and was maintained in this position through the use of a lightweight thermoplastic splint. The force sensing apparatus was positioned on the volar aspect of the distal humerus such that the padding was 1 centimetre proximal to the forearm when the elbow was held in the 90° flexed position. For testing glenohumeral adduction, the axis of rotation described previously was aligned with the axis of rotation of the dynamometer, and the upper limb positioned as described above. The force sensing apparatus was positioned on the lateral aspect of the distal humerus such that the padding was 1 centimetre proximal to the lateral epicondyle when the elbow was held in the 90° flexed position.

The pattern for testing each movement was explained to the participant. A standardised reciprocal movement testing protocol, involving only concentric muscle action, was used that included the same explanation and instructions, familiarisation period, warm-up period, and testing period for each participant (Appendix D). Angular velocity for testing both movements was set to 60°/s, as this velocity had been used previously with athletic, non-athletic, and paraplegic participants (Ivey et al., 1985; Pawlowski & Perrin, 1989; Nicholas et al., 1989; Burnham et al., 1993). The range of motion for testing flexion was from the neutral resting position (0°) into 90° glenohumeral flexion; that for adduction was from 80° to 0°. These ranges were selected in order to limit the amount of scapular movement associated with glenohumeral flexion and adduction.

Participants were asked to perform 3 maximal contractions for each muscle group during each test session. This number of contractions has been used in testing by previous investigators (MacDonald et al., 1988, Estenne et al., 1989) .

Data Collection, Reduction and Analysis

Data collection was performed through the use of the software package provided by the manufacturer of the dynamometer. Force, angular displacement, angular velocity and duration of force application were recorded. Force application was sampled at a fixed rate of 100 Hz, by the software provided by the manufacturer. Winter (1990) described the sampling theorem and specified that to reconstruct the input signal it must be sampled at a "frequency at least twice as high as the highest frequency present in the signal itself". A characteristic force tracing ramps up, peaks, and ramps down. Because the scapular elevation/depression movements required approximately one second to complete (25 degrees of movement at 30°/s), and the glenohumeral flexion and adduction movements required approximately 1.5 seconds to complete (90/80 degrees of movement at 60°/s), the sampling rate of 100 Hz was more than adequate to accurately capture the force tracings. All sensors in the system provide analogue input signals, which were converted to digital signals by the computer's A/D converter.

Using a custom designed computer program (Convert), the recorded information was converted into a format that could be exported to other computer programs more efficiently than is allowed using the manufacturer's software, and transferred via floppy disk to another computer. The Convert program created two files: an ASCII text file

(*.dat) that could be reviewed directly by the operator, and a batch file (*.bai) that could be exported to other computer software applications for further analysis. Other software used was Quattro Pro (Version 6, Novell Corporation) for spreadsheet analysis, SlideWrite Plus (Version 2.1, Advanced Graphics Software) for graph generation, and SPSS (Release 8.0, SPSS Inc.) for statistical analysis.

When conducting isovelocity testing, a force versus angular position graph was generated. One method to represent muscular strength when using isovelocity testing is to report the peak force (measured in Newton metres - Nm) recorded during the test session. Perrin (1993) determined that peak force varies in repeated contractions during the same test session, and suggested reporting an average of the peak values obtained. Another method of representing muscular strength is to report the total work done by the muscle group. Dvir (1995) explained that in this method the recorded force generated is multiplied by the angular displacement, with the unit of measure being Joules. Effectively, this is the area under the curve of the force versus angular position graph (AUC). Perrin (1993) indicated that this method of reporting can reveal deficits in overall force production that would not be apparent if only peak force values are reported. Therefore, in this study area under the curve values were determined from force data provided by the three contractions generated during each test session. The three curves generated were averaged to provide one curve for each participant. The averaged curves from each participant were combined to calculate the composite graphs.

Statistical Analysis

The statistical tests used to analyse the data obtained were the intraclass correlation coefficient (ICC) and a linear regression analysis. Shrout and Fleiss (1979) and Portney and Watkins (1993) indicate that the ICC provides a single index that overcomes the limitations of using other correlation measures to express reliability. From the models described by Shrout and Fleiss (1979), the model 3, k type (two-way mixed effect layout) was used because there is a single rater for this study, and mean data is used for the analysis. To complete the analysis, the AUC values from each repetition were averaged to provide a mean AUC value for each participant (individual AUC mean) in the four tested movements. The individual AUC means from each test session were compared using the ICC to determine reliability. Significance value for the ICC was determined a priori to be $p \leq 0.05$.

The data from the participant group with quadriplegia was compared to the data from the participant group with paraplegia through a comparison of the 95 percent confidence interval of the ICC . This revealed the extent and magnitude of any differences between the groups.

A further analysis of the data was done through a linear regression to determine the significance and magnitude of any shift of the data from test one to test two. This analysis provided the r^2 value, intercept and slope of the regression line in order to assess the predictive ability of the test protocol for further research to be conducted in the future. Significance value for the linear regression was determined a priori to be $p \leq 0.05$.

The statistical tests above were used to test the hypotheses of this study, as well as some other variables pertinent to the outcome. Tables 1, 2, and 3 include a summary of the variables that were analysed, and the statistical tests used for the analysis as they pertain to the study hypotheses. The detailed results of the statistical tests are presented in appendices F through H.

Table 1 - Variables and statistical tests for objective one.

Hypothesis	Independent variable	Dependent variable	Statistical test
(Hypotheses 1 – 3) No change in isovelocity moment generated when test 1 is compared to test 2 in each test movement	Test 1 area under the curve (auc) value	Test 2 auc value	ICC and Linear regression

Table 2 - Variables and statistical tests for objective two.

Hypothesis	Independent variable	Dependent variable	Statistical test
(Hypothesis 4) No difference between test-retest coefficient between groups	Test-retest coefficient for group with quadriplegia	Test-retest coefficient for group with paraplegia	Compare ICC confidence interval values

Table 3 - Variables and statistical tests for subset analysis objective.

Hypothesis	Independent variable	Dependent variable	Statistical test
(Hypothesis 5) No difference between values of group with quadriplegia re: active sternal head	AUC values for group with non-active sternal head	AUC values for group with active sternal head	Compare ICC confidence interval values
(Hypothesis 6) No difference between values of groups re: recreation	AUC values for non-recreation group	AUC values for recreation group	Compare ICC confidence interval values
No difference between age of groups	Average of ages of group with quadriplegia	Average of ages of group with paraplegia	Independent t-test

Results

Demographic data

Demographic data of study participants is presented in Table 4. The majority of participants (94.7%) were male. Mean age for the group of participants with quadriplegia was 33.2 years, with a standard deviation of ±8.4 years. Mean age of the group of participants with paraplegia was 38.1 years, with a standard deviation of ±8.8 years. There was no significant difference in age between the two groups. Of the group with quadriplegia, five had lesions at the C6 spinal cord level; these participants also did not have an active sternal head of the pectoralis major muscle. Eight of the participants with quadriplegia were either employed or were students, with occupations that would be classified as sedentary. Only three of the participants with quadriplegia participated in recreational activities more than two times per week. Of the group with paraplegia, seven had T12 lesions, one lesion was at L1, and two at L2. Eight of the ten were employed or were students, in occupations that would be classified as sedentary. Only two of the participants with paraplegia participated in recreational activities more than two times per week.

Table 4 - Demographic data for study participants.

PARTICIPANTS WITH QUADRIPLEGIA

PARTICIPANT NUMBER	SEX	AGE	INJURY LEVEL	ACTIVE P.M. STERNAL HEAD	EMPLOYED	RECREATION > 2 times/week
Q1	M	40	C6	NO	YES	YES
Q2	M	30	C7	YES	NO	NO
Q3	M	25	C6	NO	YES	YES
Q4	M	24	C7	YES	STUDENT	NO
Q5	M	24	C6	NO	YES	YES
Q6	M	41	C7	YES	YES	NO
Q7	M	46	C6	NO	NO	NO
Q8	M	26	C6	NO	STUDENT	NO
Q9	M	41	C7	YES	YES	NO
Q10	F	35	C7	YES	YES	NO

PARTICIPANTS WITH PARAPLEGIA

PARTICIPANT NUMBER	SEX	AGE	INJURY LEVEL	ACTIVE P.M. STERNAL HEAD	EMPLOYED	RECREATION > 2 times/week
P1	M	31	T12	YES	YES	NO
P2	M	43	T12	YES	NO	NO
P3	M	31	L2	YES	YES	NO
P4	M	52	T12	YES	SEMI	YES
P5	M	21	T12	YES	STUDENT	YES
P6	M	42	L2	YES	YES	NO
P7	M	41	T12	YES	YES	NO
P8	M	35	T12	YES	YES	NO
P9	M	45	L1	YES	NO	NO
p10*	M	40	T12	YES	STUDENT	NO

P.M. = Pectoralis Major

*withdrew due to shoulder injury between test sessions

Test data

Representative data for each test condition are presented graphically in figures 5 to 16. These graphs provide a visual representation of the moments generated by the participants at specific angles in the tested movements. For each test, a graph depicting the standard deviation is used in order to show the variability of the individual participant's raw data. The combined data for the groups is also presented, specifically in figures 8, 11, 14, and 17, to show the overall results. The combined curves were derived by taking the mean of the three data curves generated by each participant during the first and second test sessions. These individual means were combined and the average of all participants for each test session is presented, with standard error bars depicting variability in patterns and moments generated. Graphs depicting the results from each participant are included in Appendix E.

Scapular Elevation

Data representative of the moment / angle relationship in participants with quadriplegia is presented in Figure 5. There is a significant relationship between Test 1 values and Test 2 values ($r^2 = .994$, $p \leq 0.01$). This indicates there is no difference between the test values for this participant.

In general, these participants showed a uniform movement pattern in which moment generation peaked within the first 5 degrees of the test movement and decreased progressively until the end of the test movement.

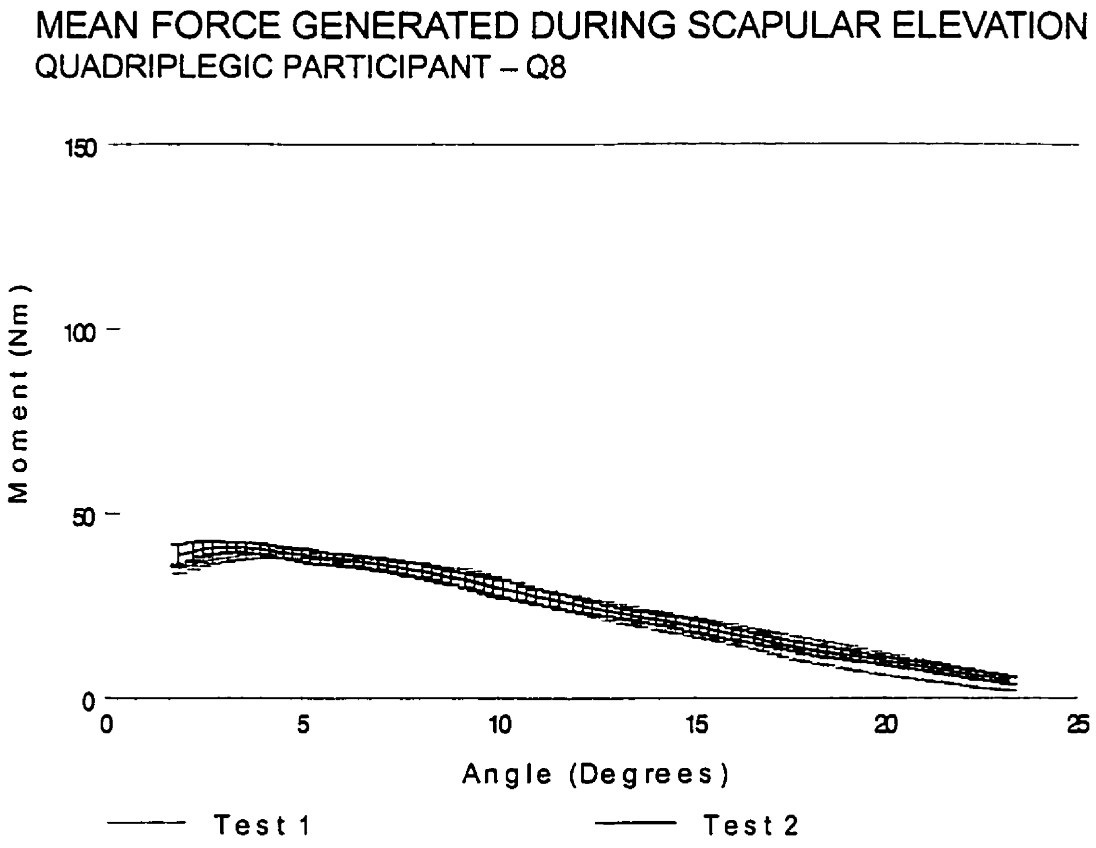
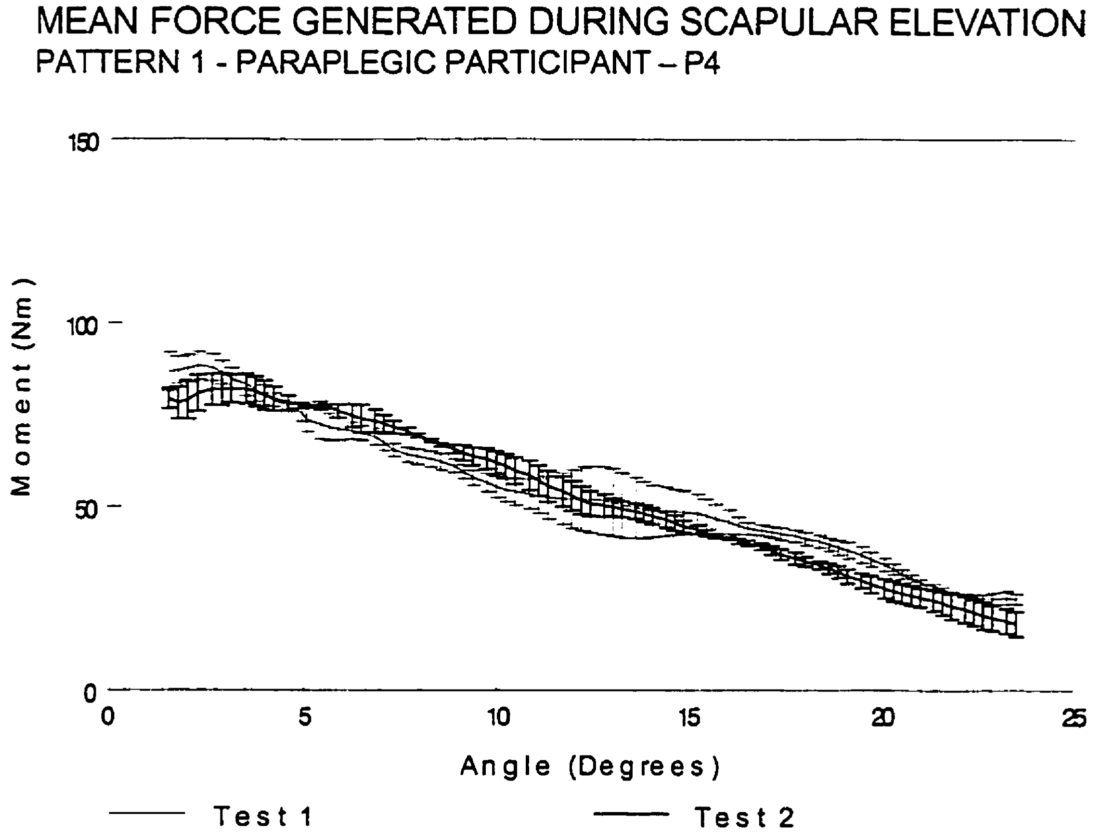


Figure 5 - Mean force generated during scapular elevation for participant Q8.



MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
PATTERN 1 - PARAPLEGIC PARTICIPANT – P4

Figure 6 - Mean force generated during scapular elevation for participant P4.

Data representative of the moment / angle relationship in participants with paraplegia is presented in Figures 6 and 7. These participants showed two distinct movement patterns, one in which moment generation peaked within the first 5 degrees of movement (n=5), and the other in which it peaked between 5 and 10 degrees of movement (n=5). A significant relationship exists between Test 1 values and Test 2 values for both patterns (Pattern 1 – $r^2 = 0.951$, $p \leq 0.01$; Pattern 2 – $r^2 = 0.954$, $p \leq 0.01$), indicating that there is no significance difference between the test values for the two distinct patterns shown by these participants.

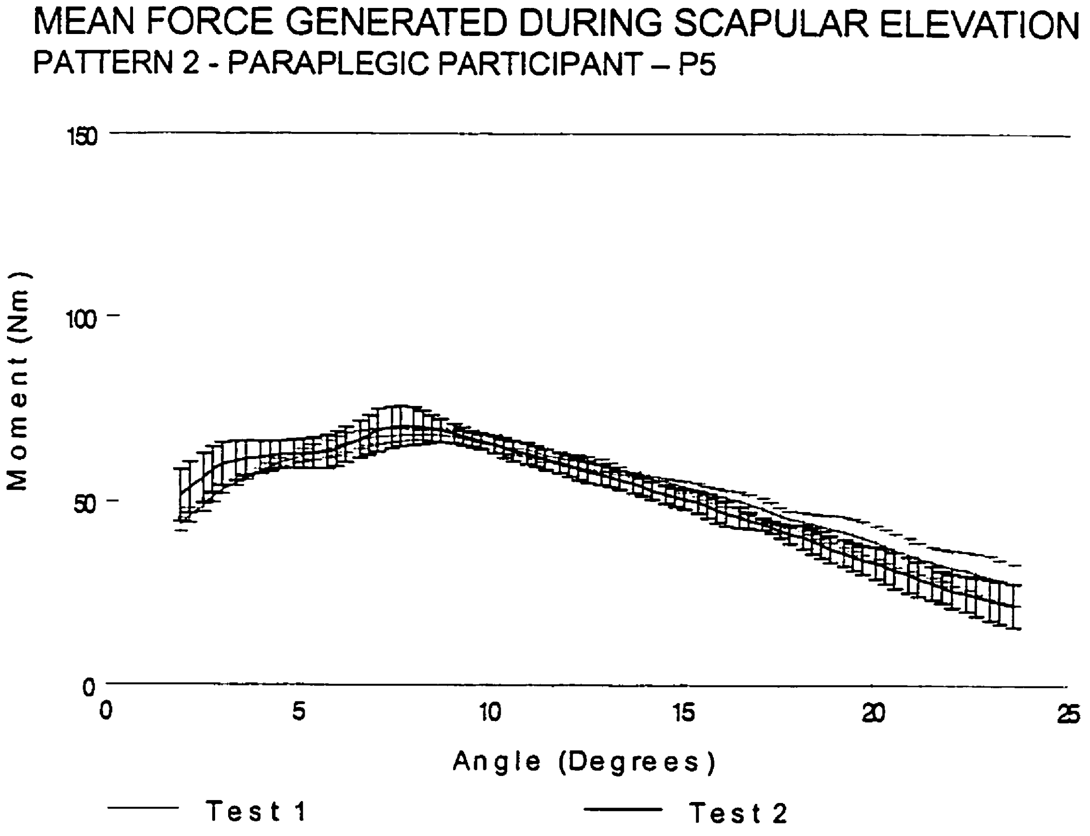


Figure 7 - Mean force generated during scapular elevation for participant P5.

Figure 8 is a combined group representation of the movement patterns for both test sessions.

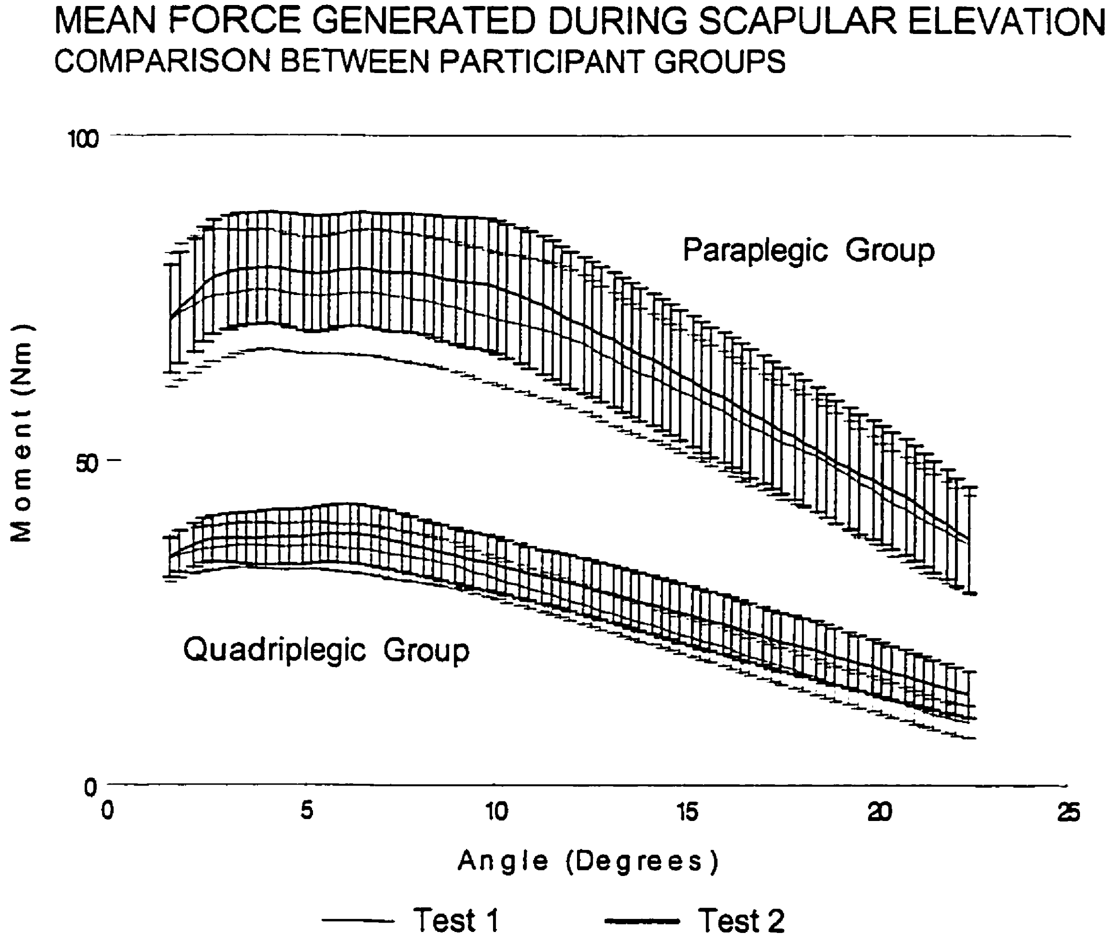


Figure 8 - Mean force generated during scapular elevation for both participant groups.

From the above figure, it is clear that participants with paraplegia produced greater moments throughout the entire test range than did the participants with quadriplegia, but that both participant groups had similar moment generation patterns. The patterns for the second test session closely overlap the first test session, with only a slight increase in moment generation. There is a significant relationship between the second test curve and the first test curve for both groups, with the paraplegic participant group having an intraclass correlation coefficient (ICC) of 0.984, and a linear regression (LR) r^2 value of 0.94 ($p \leq 0.01$), while the quadriplegic participant group had an ICC value of 0.903, and LR r^2 value of 0.76 ($p \leq 0.01$).

Scapular Depression

Data representative of the moment / angle relationship in participants with quadriplegia is presented in Figure 9. The most convenient testing method for this movement was to have the participants engage in the test motion that was the return from the scapular elevation end point, i.e. the 25 degree mark, generating moments through to the end point, the zero mark. Thus, descriptions of peak moment generation are relative to the start point of 25 degrees. The dynamometer was calibrated to record the elevation moments as positive, and the depression moments as negative. This explains the representation of depression moments as negative in figure 9. There is a significant relationship between Test 1 values and Test 2 values ($r^2 = .950$, $p \leq 0.01$). This indicates there is no difference between the test values for this participant. In general, the participants with quadriplegia showed a movement pattern in which moment generation peaked within the first 10 degrees of movement, and decreased until the end of the movement.

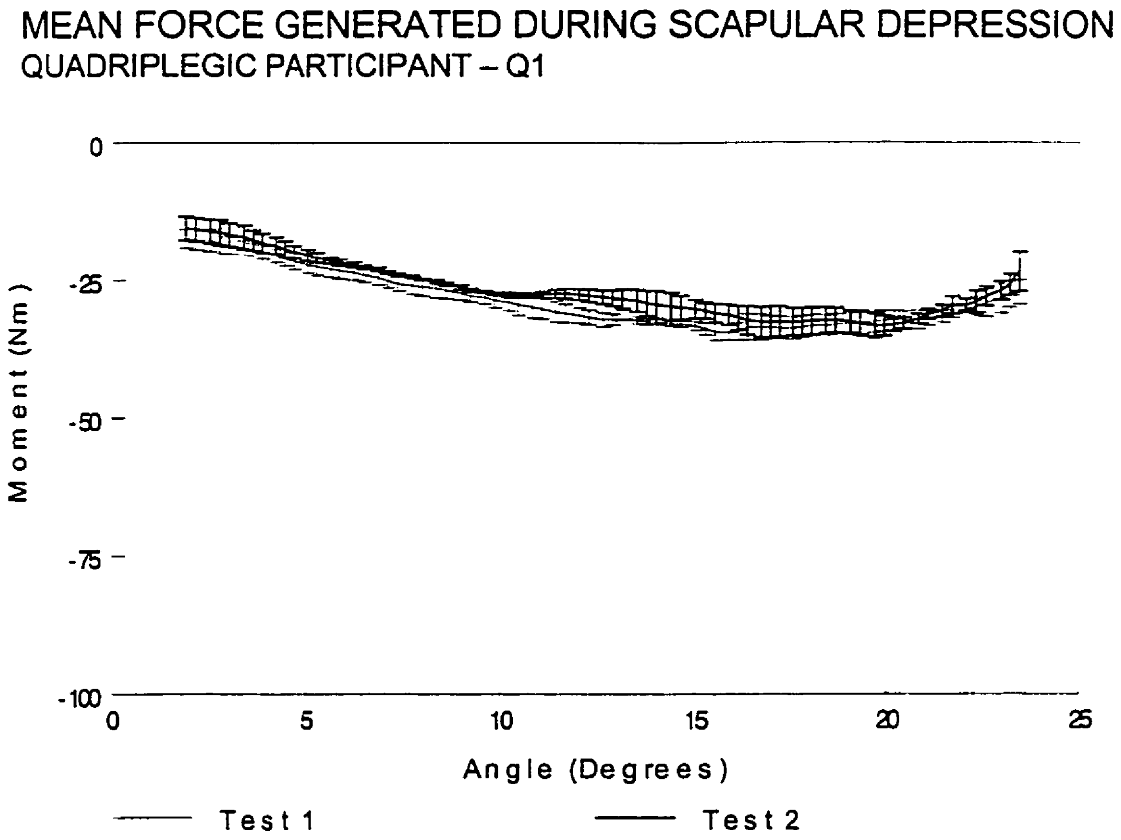


Figure 9 - Mean force generated during scapular depression for participant Q1.

Figure 10 shows data representative of the moment / angle relationship in participants with paraplegia. There was a slight increase in moment generation throughout the test range during the second test for this participant, but a significant relationship does exist ($r^2 = 0.939$, $p \leq 0.01$).

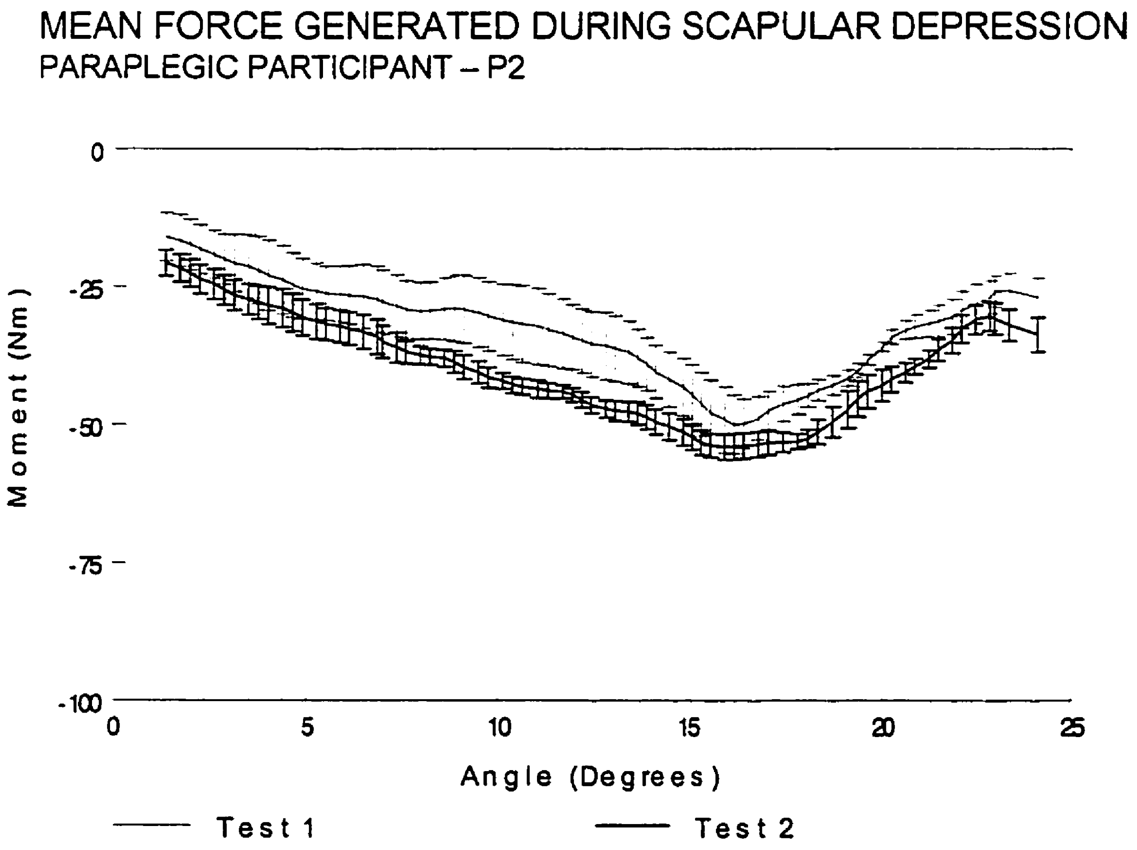


Figure 10 - Mean force generated during scapular depression for participant P2.

Figure 11 is a combined group representation of the movement patterns for both test sessions. In general, with depression, participants with paraplegia showed variability in moment generation; peak moment generation in participants with paraplegia occurred slightly later in the movement than that for participants with quadriplegia.

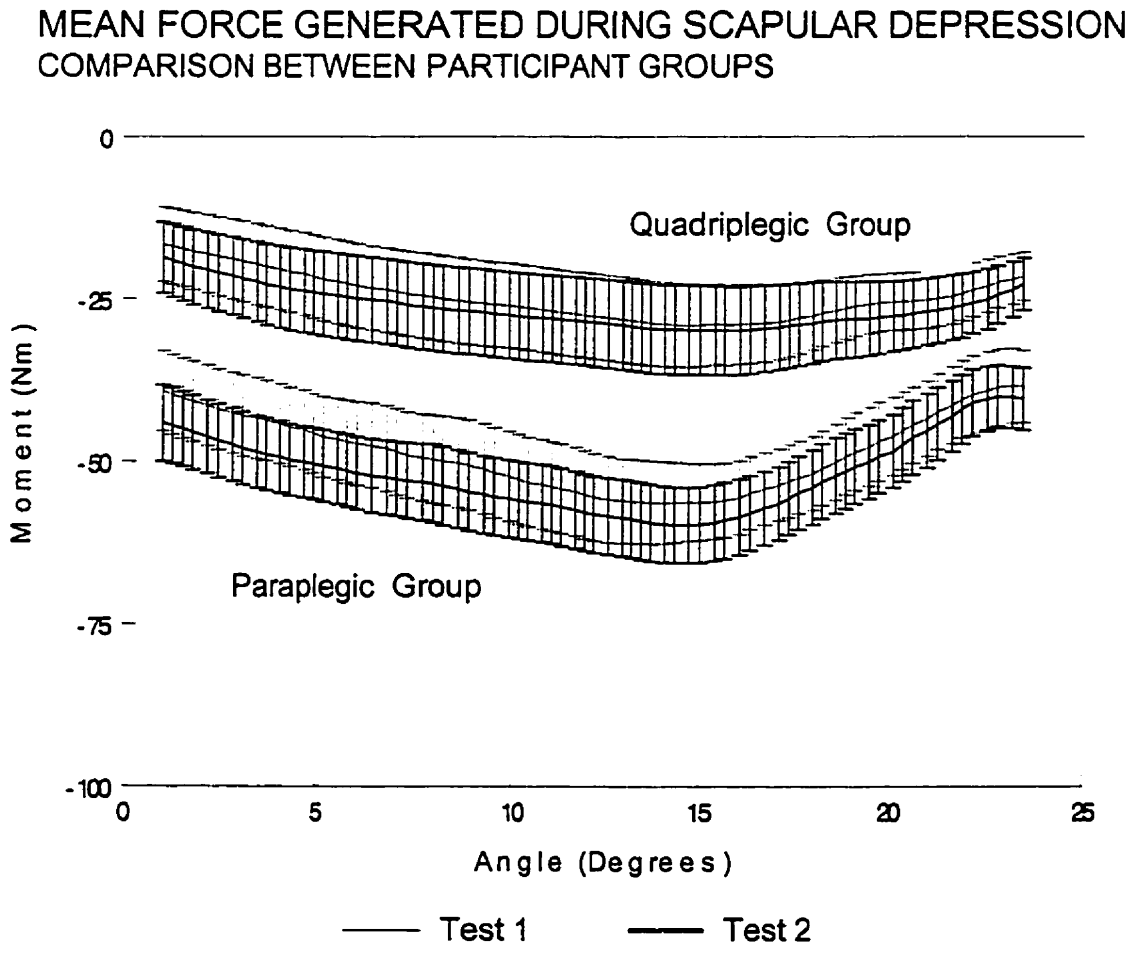
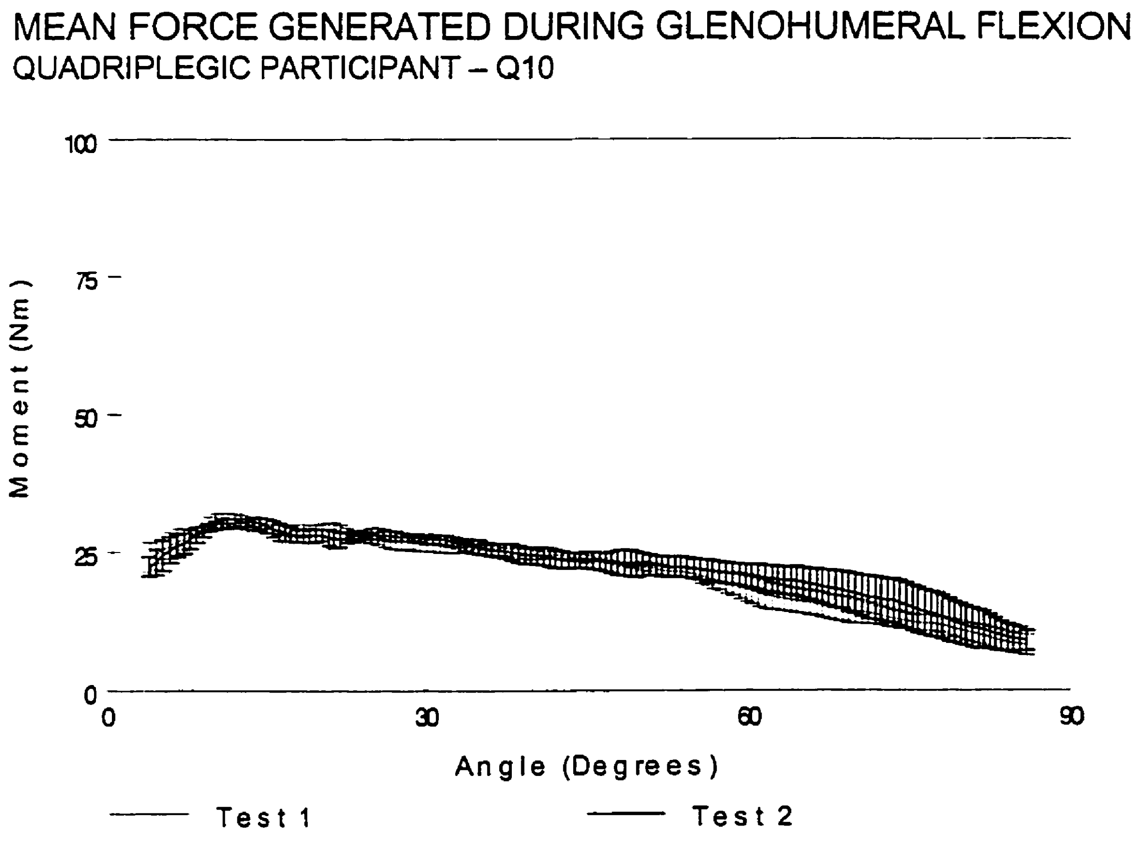


Figure 11 - Mean force generated during scapular depression for both participant groups.

From the above figure, it is clear that participants with paraplegia had the ability to produce greater moments throughout the entire test range than did those with quadriplegia, with peak moment generation occurring slightly later than the participants with quadriplegia. The patterns for the second test session closely overlapped the first test session, with only a slight increase in moment generation. There is a significant relationship between the second test curve and the first test curve for both groups, with the paraplegic participant groups having an ICC of 0.981, and a LR r^2 value of 0.94 ($p \leq 0.01$), while the quadriplegic participant group had an ICC value of 0.994, and LR r^2 value of 0.98 ($p \leq 0.01$).

Glenohumeral Flexion

Data representative of the moment / angle relationship in participants with quadriplegia is presented in Figure 12. The second test tracing very nearly overlaps the first test tracing, with some increased variability and moment generation in the last half of the second test tracing. A significant relationship between the first and second test values exists with an r^2 value of 0.976 ($p \leq 0.01$). Effectively, there is no significant difference between the second test and the first test for this participant. In general, the participants with quadriplegia showed a uniform pattern of moment generation which peaked within the first 10 degrees of movement.



MEAN FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT – Q10

Figure 12 - Mean force generated during glenohumeral flexion for participant Q10.

Data representative of the moment / angle relationship in participants with paraplegia is presented in Figure 13. The tracings for this participant show a slight decrease in moment generation in the first half of the second test, with increased variability and a drop off of moment generation at the end of the test range. A significant relationship does exist between the two test sessions, with an r^2 value of 0.973 ($p \leq 0.01$). Therefore, this participant demonstrated no significant difference between test one and test two. In general, these participants also showed a uniform movement pattern.

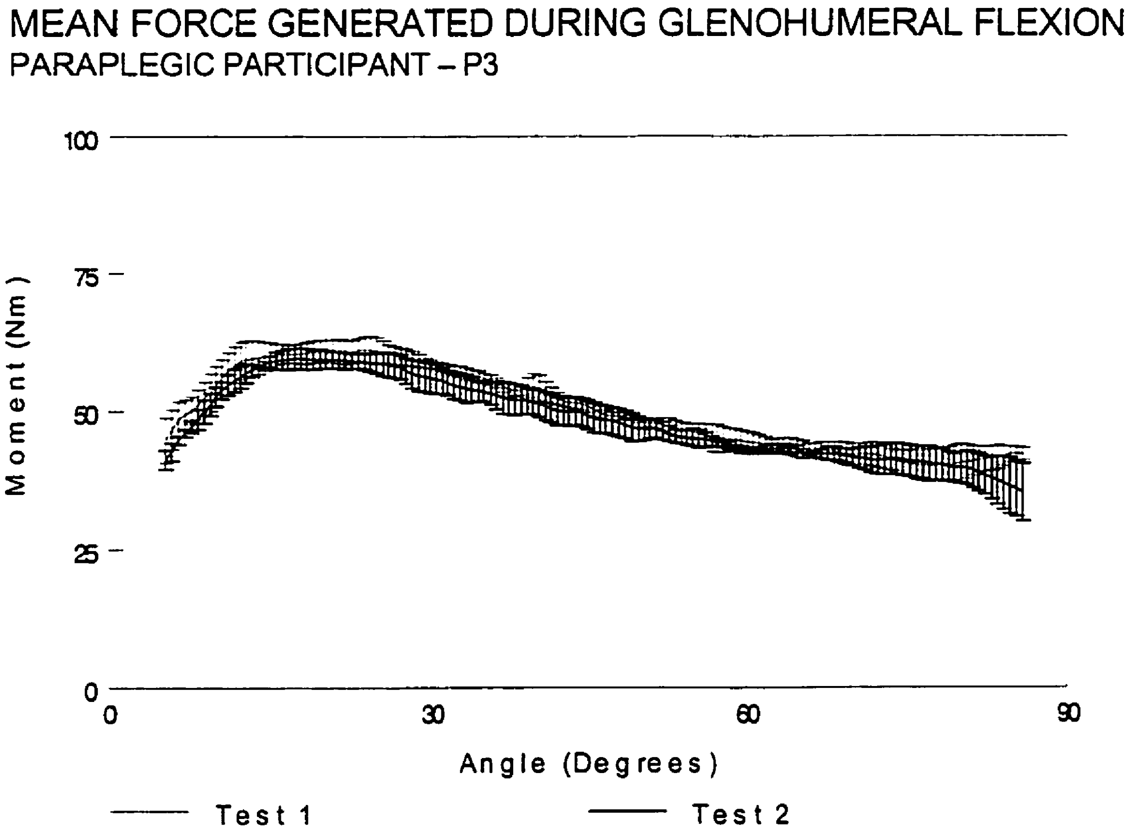


Figure 13 - Mean force generated during glenohumeral flexion for participant P3.

Figure 14 is a combined group representation of the movement patterns for both test sessions.

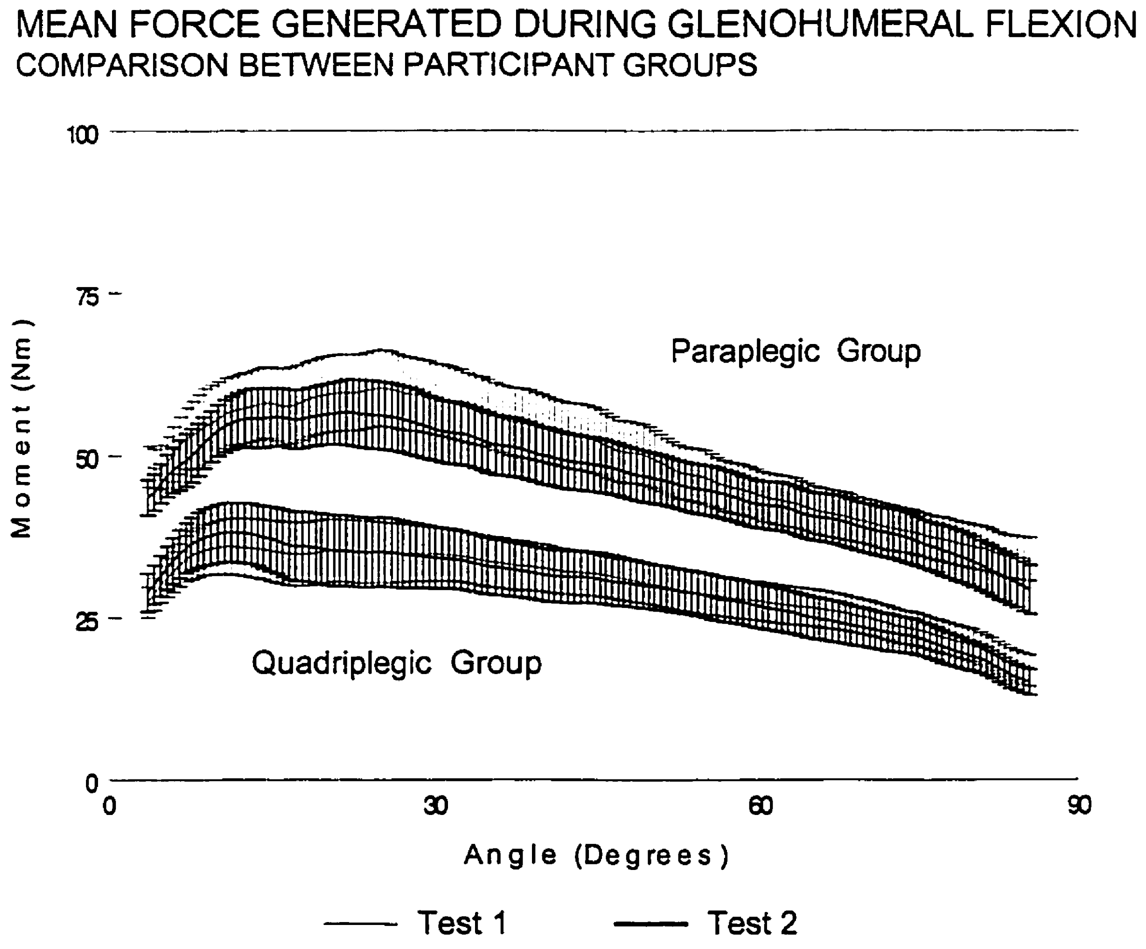
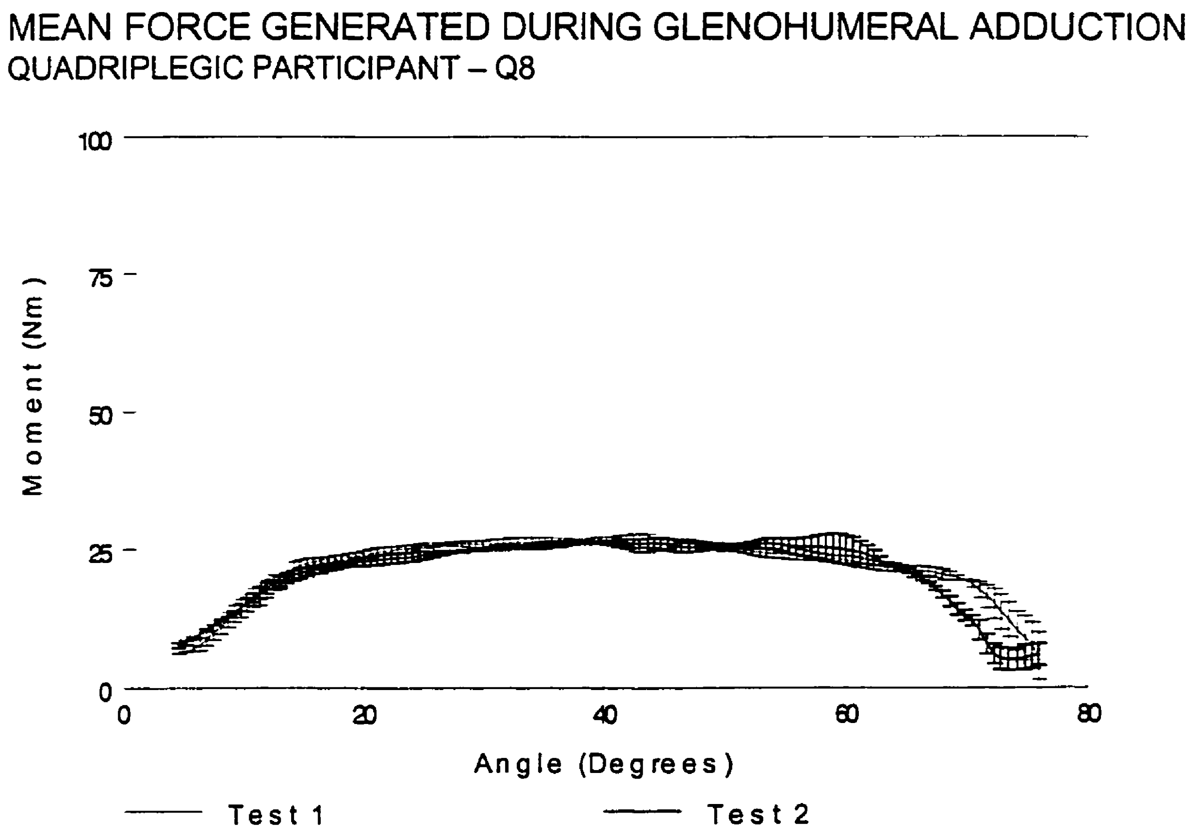


Figure 14 - Mean force generated during glenohumeral flexion for both participant groups.

From the above figure, it is clear that participants with paraplegia had the ability to produce greater moments throughout the entire test range than those with quadriplegia. Moment generation peaked later than the participants with quadriplegia, but still within the first 30 degrees of movement. The patterns for the second test session closely overlapped the first test session. It is interesting to note that in this movement, the group with paraplegia showed a slight decrease in moment generation throughout the entire test range during the second test session, while the group with quadriplegia showed nearly uniform moment generation between the two test sessions. In spite of the differences that appear between the tracings for the two test sessions, there is a significant relationship between the second test curve and the first test curve for both groups, with the paraplegic participant groups having an ICC of 0.952, and a LR r^2 value of 0.83 ($p \leq 0.01$), while the quadriplegic participant group had an ICC value of 0.986, and LR r^2 value of 0.95 ($p \leq 0.01$).

Glenohumeral Adduction

Data representative of the moment / angle relationship in participants with quadriplegia is presented in Figure 15. It should be noted that all descriptions of peak moment generation for adduction are relative to the testing start point of 80 degrees. Thus, these graphs are read from right to left. This participant with quadriplegia showed a uniform pattern of moment generation which peaked approximately in the midrange of the movement. During the second test, there was a decrease in the initial moment generation, followed by an overshoot of the first tracing in the next segment, with continued mismatch of the tracings in the remainder of the movement. In spite of this, this participant demonstrated a significant relationship between the first and second test sessions, with an r^2 value of 0.868 ($p \leq 0.01$). In general, participants with quadriplegia showed a uniform movement pattern.



MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT – Q8

Figure 15 - Mean force generated during glenohumeral adduction for participant Q8.

Data representative of the moment / angle relationship in participants with paraplegia is presented in Figure 16.

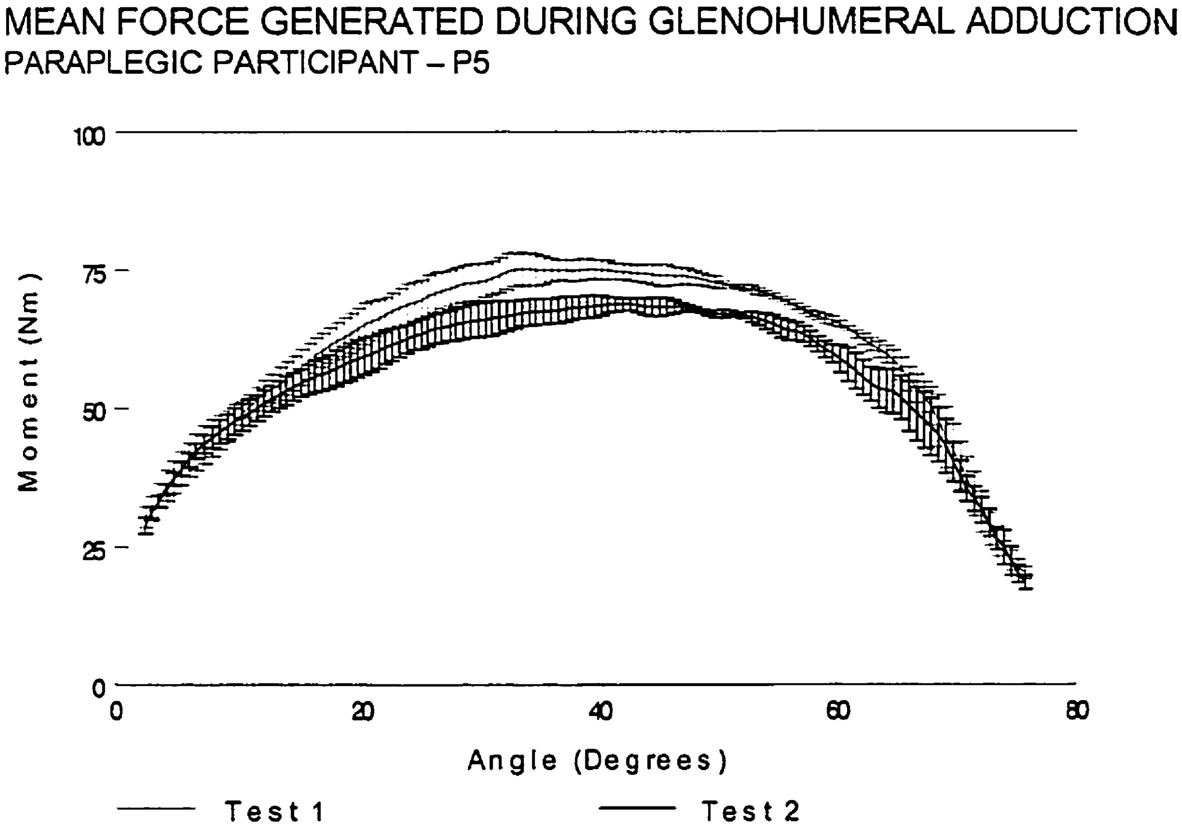

MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT – P5

Figure 16 - Mean force generated during glenohumeral adduction for participant P5.

This participant demonstrated an overall decrease in moment generation throughout nearly all of the test range between the two sessions, but areas of increased variability are very closely matched between sessions. This participant demonstrated a significant relationship between the two test sessions with an r^2 value of 0.992, ($p \leq 0.01$). In general, these participants also showed a uniform movement pattern, but moment generation peaked later than the participants with quadriplegia.

Figure 17 is a combined group representation of the movement patterns for both test sessions.

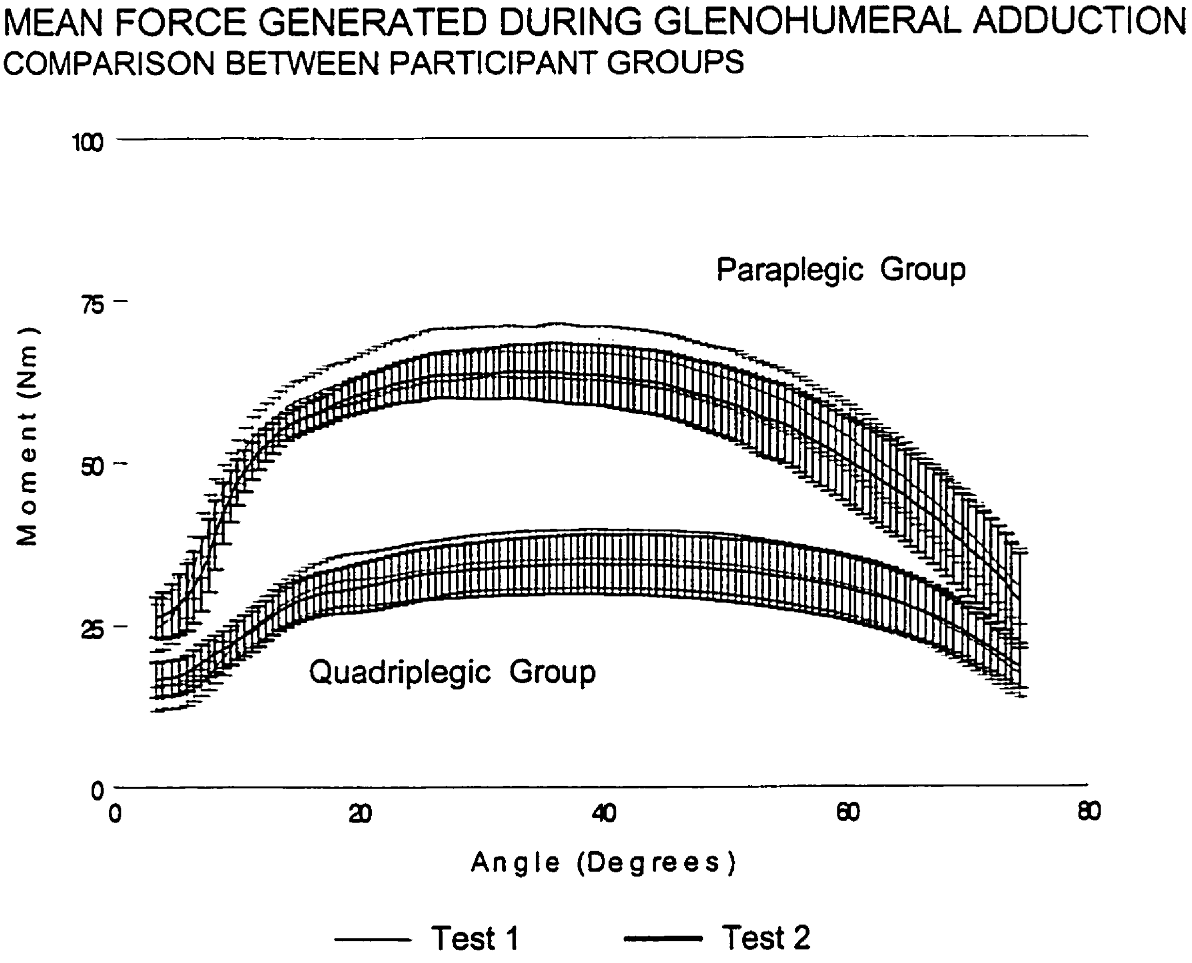


Figure 17 - Mean force generated during glenohumeral adduction for both participant groups.

From the above figure, as with other tested movements, it is clear that participants with paraplegia have the ability to produce greater moments throughout the entire test range than those with quadriplegia, with peak moment generation occurring later in the movement when compared to the participants with quadriplegia. The patterns for the second test session closely overlap the first test session. Similar to the flexion movement, it is interesting to note that in this movement, the group with paraplegia showed a slight decrease in moment generation throughout nearly the entire test range during the second test session, while the group with quadriplegia showed nearly uniform moment generation between the two test sessions. In spite of the differences that appear between the tracings for the two test sessions, there is a significant relationship between the second test curve and the first test curve for both groups, with the paraplegic participant groups having an ICC of 0.960, and a LR r^2 value of 0.86 ($p \leq 0.01$), while the quadriplegic participant group had an ICC value of 0.996, and LR r^2 value of 0.98 ($p \leq 0.01$).

Table 5 is a summary of the intraclass correlation coefficients (ICC) and r^2 values from the linear regression analysis for each movement tested. It can be seen from the statistical analysis that there is little variability and hence high reliability between the values obtained during the two test sessions. The linear regression results show good to excellent correlations between the two test sessions (Portney & Watkins, 1993).

Table 5 - Intraclass correlation coefficients and linear regression values for hypothesis four.

Hypothesis 4 No difference between test-retest coefficient between groups	Quadriplegic Participants		Paraplegic Participants	
	ICC	Linear Regression r^2 value*	ICC	Linear Regression r^2 value*
Scapular Elevation	.903	.76	.984	.94
Scapular Depression	.994	.98	.981	.94
Glenohumeral Flexion	.986	.95	.952	.83
Glenohumeral Adduction	.996	.98	.960	.86

* $p \leq 0.01$

Table 6 is a summary of the subset analysis comparing the participants with quadriplegia having an active sternal head of pectoralis major (ASH) with those having a non-active head of pectoralis major (NASH). High reliability and correlations exist for these groups for the two test sessions (Portney & Watkins, 1993).

Table 6 - Intraclass correlation coefficients and linear regression values for hypothesis five.

Hypothesis 5 No difference between test-retest AUC values between group with active sternal head (ASH) of pectoralis major, and group with non-active sternal head (NASH) of pectoralis major	ASH Participants (n=5)		NASH Participants (n=5)	
	ICC	Linear Regression r^2 value*	ICC	Linear Regression r^2 value*
Scapular Elevation	.893	.82	.984	.97
Scapular Depression	.992	.99	.956	.96
Glenohumeral Flexion	.993	.99	.946	.92
Glenohumeral Adduction	.996	.99	.979	.96

* $p \leq 0.034$

Table 7 is a summary of the subset analysis comparing the group of participants who engaged in recreational activities more than two times per week, with those who engaged in recreational activities two times or less during the week. As with the previous grouping of participants, there is high reliability and correlations for these groups for the two test sessions (Portney & Watkins, 1993).

Table 7 - Intraclass correlation coefficients and linear regression values for hypothesis six.

Hypothesis 6 No difference between test-retest AUC values between group with greater than two days (2+) of recreational activity in one week, and group with 2 or fewer days (2-) of recreational activity in one week	2+ Participants (n=5)		2- Participants (n=14)	
	ICC	Linear Regression r^2 value*	ICC	Linear Regression r^2 value*
Scapular Elevation	.997	1.00	.977	.95
Scapular Depression	.992	.98	.982	.96
Glenohumeral Flexion	.959	.92	.966	.94
Glenohumeral Adduction	.975	.98	.980	.96

* $p \leq 0.01$

Summary of Results

From the above results, it can be seen that for all test movements participants with paraplegia were able to generate greater moments than participants with quadriplegia, but that similar moment generation patterns are seen. Statistical analyses show good to excellent reliability in all movements tested and also in subset analysis of varied groupings of participants based on considerations other than injury level.

Discussion

The results support the hypotheses which predicted that there would be no change in isovelocity moments generated during two test sessions separated by not less than 72 hours, but not more than one week, in all test conditions, and all subset analyses. In an examination of the combined data graphs in figures 8, 11, 14, and 17, one can see higher moment generating capabilities of the paraplegic group, similar movement patterns for both groups, and a similar amount of variability about the mean for each tested movement, in each test session. Reliability values contained in tables 5, 6, and 7, indicating good to high reliability ((Portney & Watkins, 1993) further support the hypotheses.

The higher moment generating capabilities of the paraplegic group compares favourably to previous work done by Powers et al. (1994), who used isometric testing for subjects with quadriplegia, paraplegia, and an able bodied group, and found reduced moment generating ability for the quadriplegic group when compared to the paraplegic and able bodied groups. That is, the quadriplegic group had significantly less moment generating ability when compared to the other groups. The paraplegic group showed lower moment generating ability than the able bodied group, but the difference was not significant. The most reasonable explanation for lower moment generation in the group with quadriplegia is the decreased available muscle mass. Participants with cervical cord lesions have decreased innervation to many of the muscles activated in the tested movements as compared to the group with paraplegia. Decreased activation of muscle groups and specific motor units implies less available muscle mass to participate in moment generation and results in further diminished maximal torque [moment] (Powers et al., 1994).

Muscular stabilisation of the trunk might allow the muscles involved in the testing process an advantage in moment generation, but this is less likely in the current study. In order to be able to generate force, the base (origin) from which the muscle works must be stable and secure, or the origin will be pulled toward the insertion. Using a weight relief transfer as an example, the arm is stabilised through the armrests of the chair, and the muscle's origin and insertion are conceptually reversed, allowing the body to be lifted off the wheelchair seat (Reyes et al., 1995). The group with paraplegia had the ability to stabilise the trunk, which serves as the base for the muscles that attach to the humerus. The group with quadriplegia had a disadvantage in moment generation due to a decreased ability to stiffen the trunk. This is due to the inability to activate the abdominal and lower back muscles, as does the group with paraplegia. Sinderby et al. (1992) suggested that the diaphragm has a postural function in people with cervical SCI. It was noted during testing that participants with a cervical SCI would inhale prior to the commencement of the test movement. This would induce an increased pressure in the abdomen, as was found by Sinderby et al. (1992), in an effort to stiffen the trunk during testing movements. Klefbeck et al.(1996) assessed trunk support in the performance of arm ergometry in a group of subjects with cervical SCI. They indicated that trunk stabilisation improved performance during this activity. Thus, as used in this study, securing the trunk to the wheelchair would be an additive measure to the postural stiffening provided by the diaphragm. Review of the patterns of moment generation in figures 8, 11, 14, and 17, support the concept that the group with quadriplegia were well supported in the testing process, and therefore were most likely able to stiffen the trunk reasonably well. The dynamometer records only the amount of force that is generated at the force sensor attached to the arm, and reflects the ability of the entire system (i.e. the muscles of the arm and the base through which they attach to the body) to generate this force. The figures show strikingly similar moment generation patterns for both groups. Differences in movement of the base of support would have been indicated by differences in the appearance in the movement pattern at the initiation of the movement. The similarity of the moment curves for both groups in all movements supports the concept that the reduced moment generation capabilities in the group with quadriplegia is more dependent on the reduced amount of available muscle mass to activate during testing, rather than shifting of an unstable base.

The combined data graphs in figures 8, 11, 14, and 17, show similar amounts of variability in each test session for each movement. Participants in each group have different moment generating abilities ranging from high to low, for any given movement. The consistency in the variability in each test session provides evidence that this range is the most likely source. The greatest amount of variability is seen in the paraplegic group in the scapular elevation movement. This is most likely caused by the two distinct patterns of moment generation in the paraplegic group for this movement, as shown in figures 6 and 7. The use and application of the coupling strap for scapular elevation and depression movements might have been an additional source of variability. This was effectively limited through the use of the wide velcro strap, providing high tensile strength through secure contact between the hook and the loop portions of the velcro. The inclusion of a flexible tape measure (Figure 4) also allowed for accurate re-application of the sling during the second test session.

A thorough search of the available literature yielded very few studies that reported reliability values for movements tested about the shoulder, and no studies were found that used isovelocity testing for measurement of movements of the shoulder girdle. The results of this study compare favourably to reliability values reported by Perrin (1986) in an analysis of shoulder flexion. Perrin (1986) indicated reliability coefficients for glenohumeral flexion as being 0.84 for the left shoulder, and 0.91 for the right shoulder, using normal subjects. The current study demonstrated reliability coefficients of 0.94 to 0.99 for the glenohumeral flexion movement.

The results of this study compare favourably with previous work related to moment generation in normal subjects, as well as athletic normal subjects. Powers et al. (1994) indicated no significant difference in isometric moment generation between a group with paraplegia, and an able bodied control group. Other investigators (Ivey et al., 1985, Walsh et al., 1985, Alderink & Kuck, 1986, Perrin, 1986, Pawlowski & Perrin, 1989) have reported mean peak shoulder flexion moments between 58.5 (Alderink & Kuck, 1986) and 73 (Walsh et al., 1985) Newton metres (Nm). The current study demonstrated a mean peak isovelocity value of 60.4 Nm. Investigators examining moment generation in the adduction movement (Ivey et al., 1985, Walsh et al., 1985, Alderink & Kuck, 1986, Reid et al, 1989) have reported mean peak values between 62.7 (Walsh et al., 1985) and 118.6 (Alderink & Kuck, 1986) Nm., with higher moments being generated by well trained athletic normals, and college aged baseball pitchers. The current study demonstrated a mean peak isovelocity value of 67.32 Nm. The moments generated in this study during glenohumeral flexion and adduction are consistent with those generated by normal able bodied subjects from the above studies.

A subset analysis of the participant group with quadriplegia was conducted to compare the results of the movements of scapular elevation/depression, and glenohumeral flexion and adduction, between the group with an active sternal head (ASH) of pectoralis major, and those of the group with a non-active sternal head (NASH). Analysis of these data supports the hypothesis that there is no difference in reliability of moment generation between test periods for each of the movements tested when comparing the two groups. The small number of participants in each group does weaken the validity, but in spite of this, one would also expect that regardless of the participant grouping there would be no significant difference in the results from test one to test two. This is most likely due to the application of the testing protocol. Each participant was given an opportunity to become familiar with the testing procedure, and each of the required movement patterns. The use of pre-test warm-up trials also helped to ensure the participant was comfortable and proficient in the movement pattern, and is consistent with the approach used in the reliability study conducted by Perrin (1986). This warm-up and practice allowed maximal effort to be exerted in both test sessions, with minimal practice or carry-over effects experienced in the second test session (Portney & Watkins, 1993).

When comparing the group of participants who engaged in recreational activity more than 2 times per week with those who participated in recreational activity less than 2 times per week, no difference in reliability of moment generation between the two test periods was observed. The small number of participants in the more active group does weaken the validity, but the effect of warm-up and practice to become proficient in test movements most likely had a similar effect here as well. It is consistent that no matter what grouping of participant subsets, reliability and correlations would show no significant difference between the two test sessions.

Age was not a significant contributing factor to the difference between the groups in moment generation.

The major weaknesses of this study are twofold: the limited number of participants, and the single tester used for data collection. The number of participants was limited by the number of available persons with spinal cord injuries who met the inclusion criteria and lived within close proximity to the testing area. In spite of the small sample size, the variability observed is small, and the reliability of the testing protocol high.

This study does not address the differences that may occur if more than one person is used to conduct the test from one session to the next. While the use of multiple testing centres and multiple testers will no doubt increase the variability of test results, it is expected that with appropriate initial instruction to the testing protocol, and careful attention to the details required during the testing procedure, it is possible that many different testers conducting the protocol would generate similar findings, as they relate to reliability. Further, in most centres the number of personnel involved in dynamometry testing is very low.

Conclusion

The results of this study support the overall hypothesis that there is no change in moment generated during one test session and a second test session separated by no less than 72 hours, but no greater than one week, in four test movements about the shoulder and shoulder girdle in participants with paraplegia or quadriplegia. Therefore, the results support the application of the test protocols used in this study as reliable methods to assess moment generation for the movements of scapular elevation and depression, and glenohumeral flexion and adduction, in participants with either paraplegia or quadriplegia. It should be noted that this testing was conducted using the Kin-Com® dynamometer, and as such are valid only for this device. Other dynamometers may show different results.

Future considerations

With the above testing protocol being established as a reliable means for measuring moment generation in participants with SCI, further testing with other groups having special needs (i.e. those who require the use of a wheelchair) can be conducted to determine reliability in these groups. Additionally, this method of testing can be used to establish objectively the efficacy of strength training programs for the muscles that effect movement about the shoulder and shoulder girdle.

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Appendix A - Kin-Com calibration procedure

1. Check voltage output of internal potentiometers, adjust to 0 volts as necessary.
2. Position actuator spline to mechanical center position, check with spirit level and adjust to horizontal.
3. Install actuator arm, use spirit level to assess vertical positioning, and adjust using potentiometer to align to top dead center.
4. Rotate actuator arm to horizontal position, assess with spirit level for true horizontal positioning, adjust angle span with potentiometer.
5. Assess tachometer zero speed setting. Adjust with potentiometer to zero motion, if necessary.
6. Assess servo zero/gain and speed gain. Adjust with potentiometer if necessary.
7. Final speed check. Assess for actuator rotation in stop position. If rotation exists, repeat steps 5 and 6 until zero rotation occurs in stop position.
8. Position actuator arm in the top dead center position. Assess load cell reading for zero output, adjust with potentiometer if necessary. Position actuator arm in horizontal position, apply known weight, adjust output reading to match known weight value using potentiometer if necessary.

Appendix B - Paraphrase and Informed Consent Form

"RELIABILITY OF ISOVELOCITY DYNAMOMETER TESTING OF FOUR MOVEMENTS OF THE SHOULDER AND SHOULDER GIRDLE IN PERSONS WITH SPINAL CORD INJURY"
University of Manitoba 1998

PARAPHRASE

The strength of the muscles around the shoulder and shoulder blade is important to persons with high level spinal cord injuries. These muscles are used in various activities, but most importantly for all functions of daily living, the propulsion of wheelchairs, and with coughing. Training programs can improve the strength of these muscles, but reliable testing procedures are needed to measure improvement accurately. Current methods of testing some of the shoulder movements have not been adequately validated. This study is aimed at establishing a reliable method of testing strength of the muscles around the shoulder and shoulder blade, particularly the movements of elevation and depression (shoulder shrug), shoulder flexion (forward movement of the arm), and shoulder adduction (movement of the arm toward the side of the body), using a sensitive piece of equipment called a dynamometer. Once accurate testing methods have been determined, effectiveness of training programs can be more accurately assessed.

Procedure

As a participant in this study, you will be asked to perform a test on a special device that measures the strength of the muscles around your shoulder and shoulder blade. You will be asked to attend a familiarisation session to show you the equipment used in testing, and to instruct you in the movements required for the test. The test session will follow within the next 48 hours. You will be asked to perform a warm up consisting of light movement of the arms, and some stretching exercises. Prior to the test, you will be asked to perform 2 or 3 movements of moderate effort to familiarise you to the movement of the testing device, and as a warm-up specific to the test movement. This warm-up will be repeated a second time, and you will be asked to provide maximal effort. During the test, you will be asked to provide 3 maximal voluntary efforts in a series of movements, which can be performed while seated in a wheelchair. The entire testing procedure will require approximately one hour to complete. Three to seven days after the test, you will be asked to return to perform the test a second time. Again, the test will require approximately one hour.

Risks

The risks associated with the performance of the test are minimal. A certain amount of muscle discomfort may be felt during the test (as with any form of exercise). However, if obvious pain arises at any point during the test, the test will be stopped. After maximal exertion of the muscles surrounding the shoulder you may experience some discomfort which may last up to 72 hours after the test. However, as only 3 maximal movements for each test are required, this discomfort should be minimal. It is a normal consequence of exercise and will resolve on its own.

There have not been any published reports of muscle or tendon injury during these tests. Similar tests have been performed on athletes, individuals with and without shoulder problems, and individuals who have spinal cord injury, without documented damage to the muscles.

You will not be identified in any published report of the results of this study. Your participation in this study is voluntary, and you are free to withdraw at any time without prejudice.

You will not receive reimbursement for participation in this study, nor will you be responsible for any costs directly related to this study.

If you do not understand any aspect of this form, please contact:

Lloyd Tymchyshyn, School of Medical Rehabilitation
University of Manitoba
787-4363

Paraphrase and Informed Consent Form
"RELIABILITY OF ISOVELOCITY DYNAMOMETER TESTING OF TWO MOVEMENTS OF THE SHOULDER AND SHOULDER GIRDLE IN PERSONS WITH SPINAL CORD INJURY"
University of Manitoba 1997

CONSENT FORM

I have read the paraphrase and understand the nature of the study including the potential benefits and risks. I am satisfied that any questions I may have had with respect to this study have been answered. I hereby consent to participate in the research study and abide by the procedural requirements.

I understand that I may withdraw from the study at any time.

________________________ ____________
Signature of Participant/date

________________________ ____________
Signature of Witness/date

________________________ ____________
Signature of Investigator /date

Appendix C - Tables for sequence of movement testing

Participants with quadriplegia.

Participant Number	Elevation/ depression flexion adduction	Participant Number	Flexion, elevation/ depression adduction	Participant Number	Adduction flexion elevation/ depression
Q4		Q5		Q2	
Q7		Q3		Q6	
Q9		Q8		Q10	
Q1					

Participants with paraplegia.

Participant Number	Elevation/ depression flexion adduction	Participant Number	Flexion, elevation/ depression adduction	Participant Number	Adduction flexion, elevation/ depression
P3		P6		P2	
P1		P9		P7	
P8		P5		P10	
P4					

Appendix D - Standardised protocol for movement testing

Instructions to participants following axial alignment of the dynamometer to the spinous process of T2.

"In a moment you will be asked to perform a maximal voluntary contraction of your trapezius muscle (the muscle that causes you to shrug your shoulder), but first, in order for you to warm-up, and get used to the movement of the machine, you need to do some practice movements. Please wait until I say, "Ready, and go." before moving. I will ask that you go through the movement now with a light effort. First I will ask you to "shrug" your shoulder, trying to touch your shoulder to your ear without moving your head. There will be a five second pause when you reach the end point, and I will ask you to push down, again with only a light effort. I will ask you to do these movements three times, and there will be a 30 second break between each repetition. Do you have any questions? Ready, and go."

2 minute break to go through 2nd warm-up instructions

Repeat the above instructions, substituting "light effort" with "moderate effort".

2 minute break to go through final testing instructions

"Now that you have completed the warm-up, we can move on to the actual test. Please wait until I say, "Ready, and go." before moving. I will ask you to push up as hard and as fast as you possibly can. As in the warm-up, there will be a five second pause when you reach the end point, and I will ask you to push down as hard and fast as you possibly can. I will ask you to do these movements three times, and there will be a 30 second break between each repetition. Do you have any questions? Ready, and go."

Instructions to participants following axial alignment of the dynamometer to the glenohumeral joint, with the elbow held in 90° flexion.

"In a moment you will be asked to perform a maximal voluntary contraction of your pectoralis major and anterior deltoid muscles (the muscles that causes you to move your arm up and forward), but first, in order for you to warm-up, and get used to the movement of the machine, you need to do some practice movements. Please wait until I say, "Ready, and go." before moving. I will ask that you go through the movement now with a light effort. First I will ask you to try to "punch" the ceiling. When you reach the end point, the machine will return your arm to the starting position. I will ask you to do this movement three times, and there will be a 30 second break between each repetition. Do you have any questions? Ready, and go."

2 minute break to go through 2nd warm-up instructions

Repeat the above instructions, substituting "light effort" with "moderate effort".

2 minute break to go through final testing instructions

"Now that you have completed the warm-up, we can move on to the actual test. Please wait until I say, "Ready, and go." before moving. I will ask you to push up as hard and as fast as you possibly can. As in the warm-up, when you reach the end point the machine will return your arm to the starting position. I will ask you to do these movements three times, and there will be a 30 second break between each repetition. Do you have any questions? Ready, and go."

Instructions to participants following axial alignment of the dynamometer to the glenohumeral joint, with the elbow held in 90° flexion, for adduction.

"In a moment you will be asked to perform a maximal voluntary contraction of your pectoralis major and latissimus dorsi muscles (the muscles that causes you to move your arm in toward your side), but first, in order for you to warm-up, and get used to the movement of the machine, you need to do some practice movements. Please wait until I say, "Ready, and go." before moving. I will ask that you go through the movement now with a light effort. First I will ask you to try to pull your arm to your side. When you reach the end point, the machine will return your arm to the starting position. I will ask you to do this movement three times, and there will be a 30 second break between each repetition. Do you have any questions? Ready, and go."

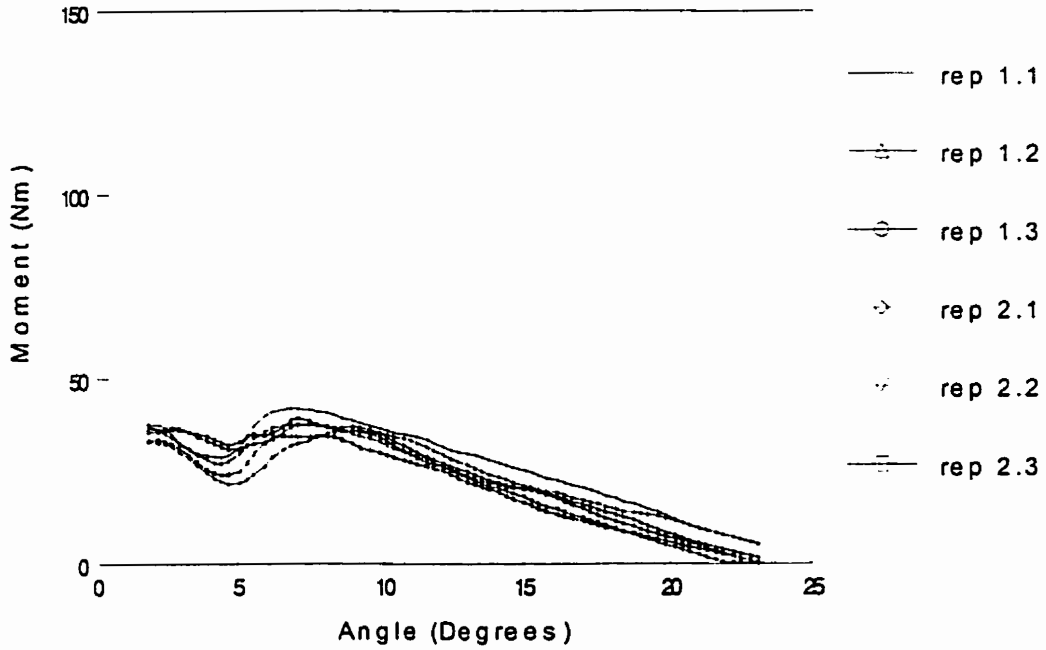
2 minute break to go through 2nd warm-up instructions

Repeat the above instructions, substituting "light effort" with "moderate effort".

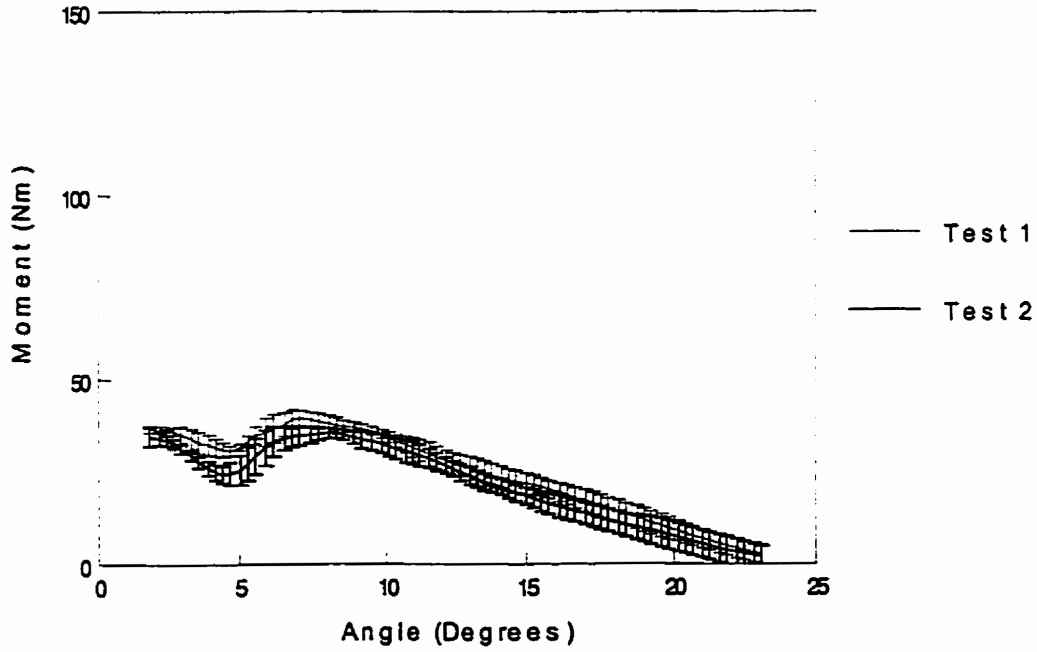
2 minute break to go through final testing instructions

"Now that you have completed the warm-up, we can move on to the actual test. Please wait until I say, "Ready, and go." before moving. I will ask you to pull your arm to your side as hard and as fast as you possibly can. As in the warm-up, when you reach the end point the machine will return your arm to the starting position. I will ask you to do these movements three times, and there will be a 30 second break between each repetition. Do you have any questions? Ready, and go."

Appendix E - Moment / Angle Graphs for Individual Participants

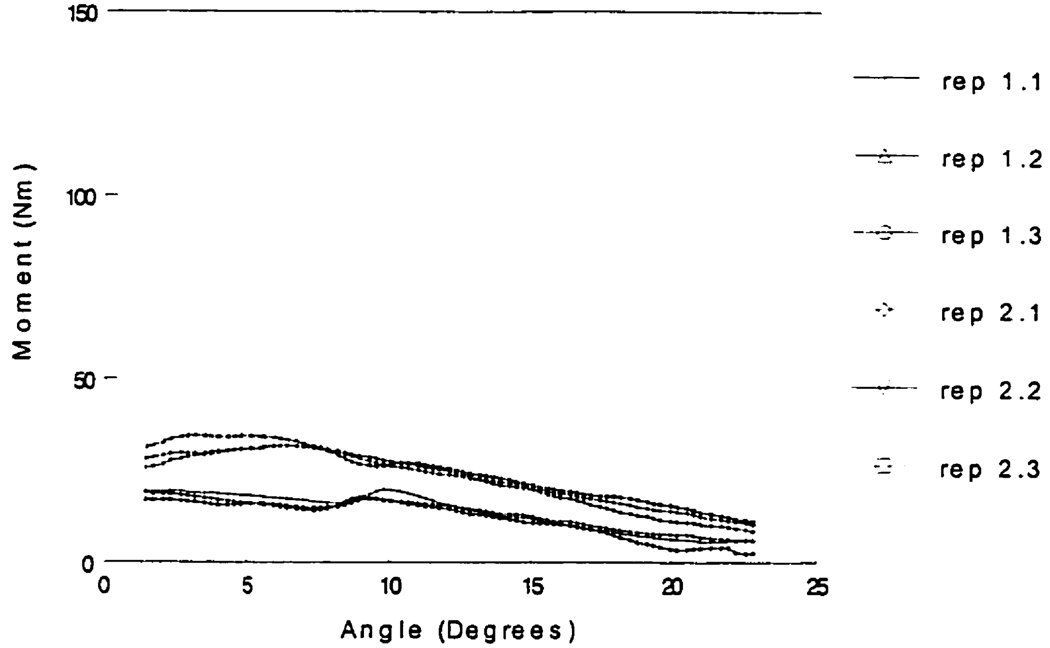


FORCE GENERATED DURING SCAPULAR ELEVATION
QUADRIPLEGIC PARTICIPANT Q1 – RAW DATA



MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q1 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants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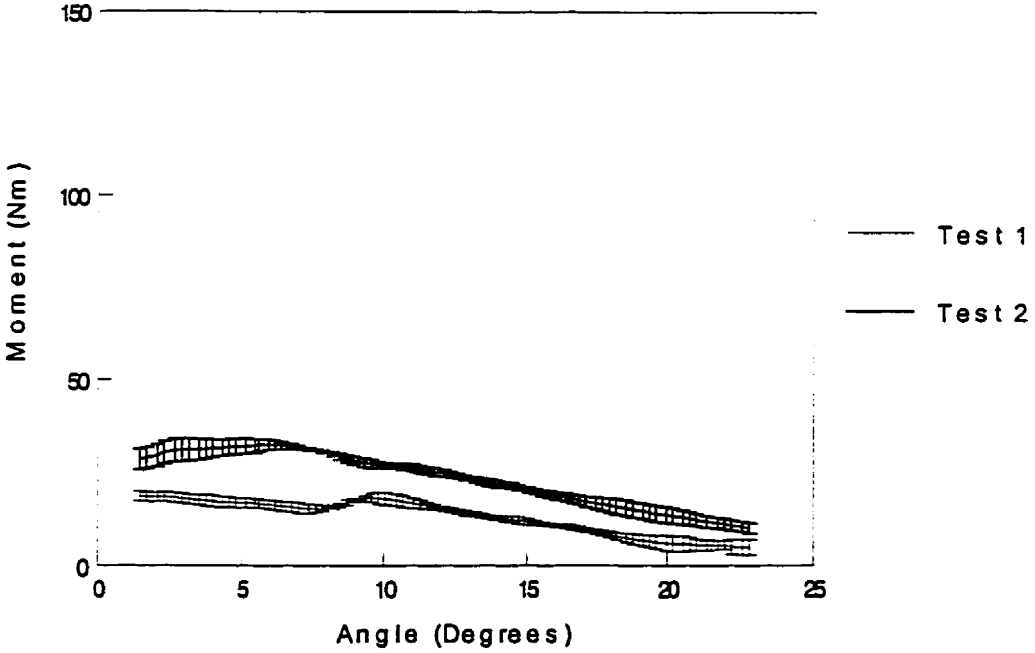
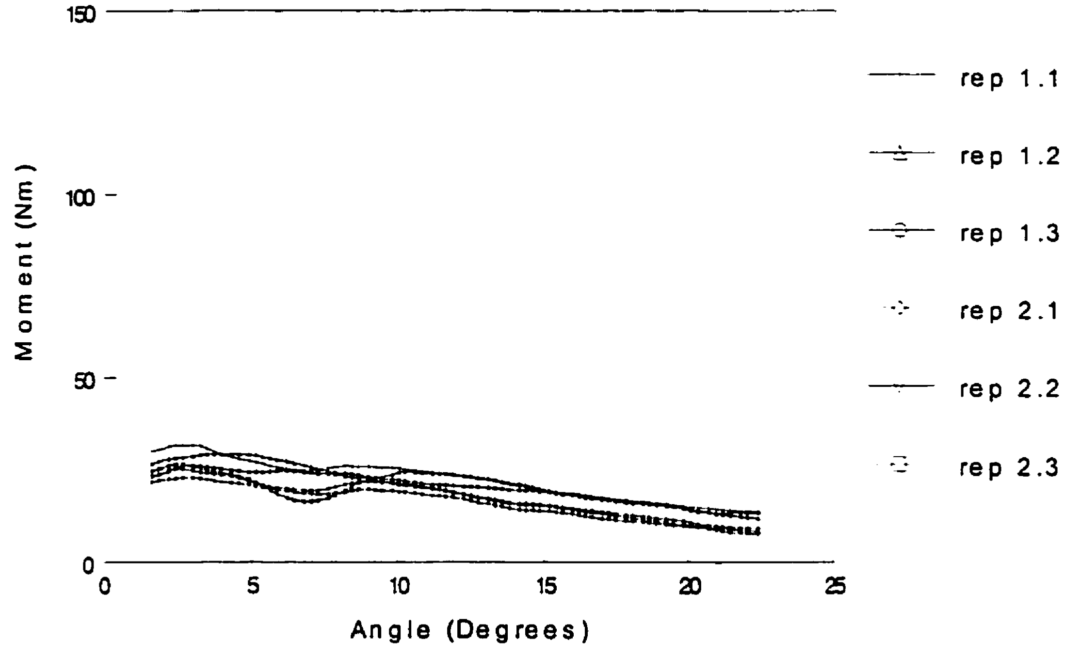

MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q2 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants



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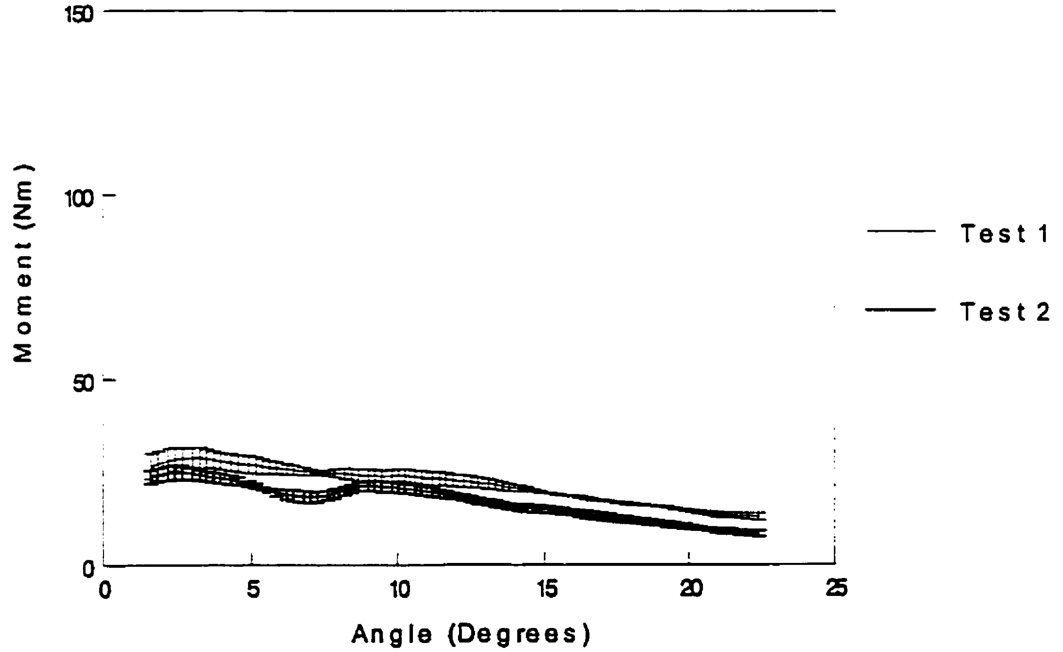
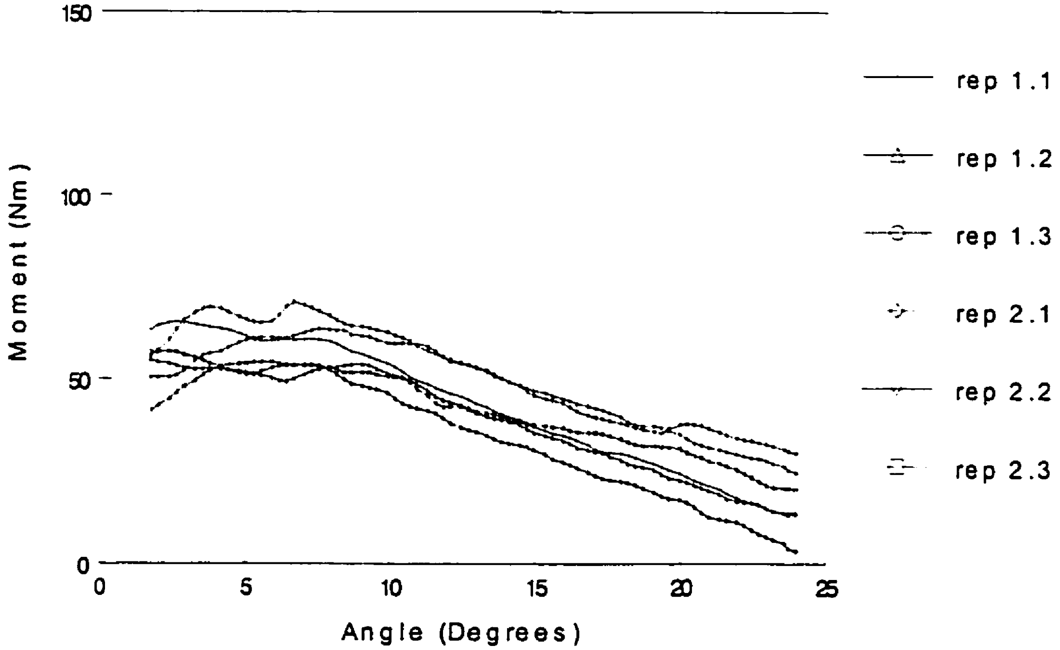
QUADRIPLEGIC PARTICIPANT Q3 – RAW DATA



MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q3 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants



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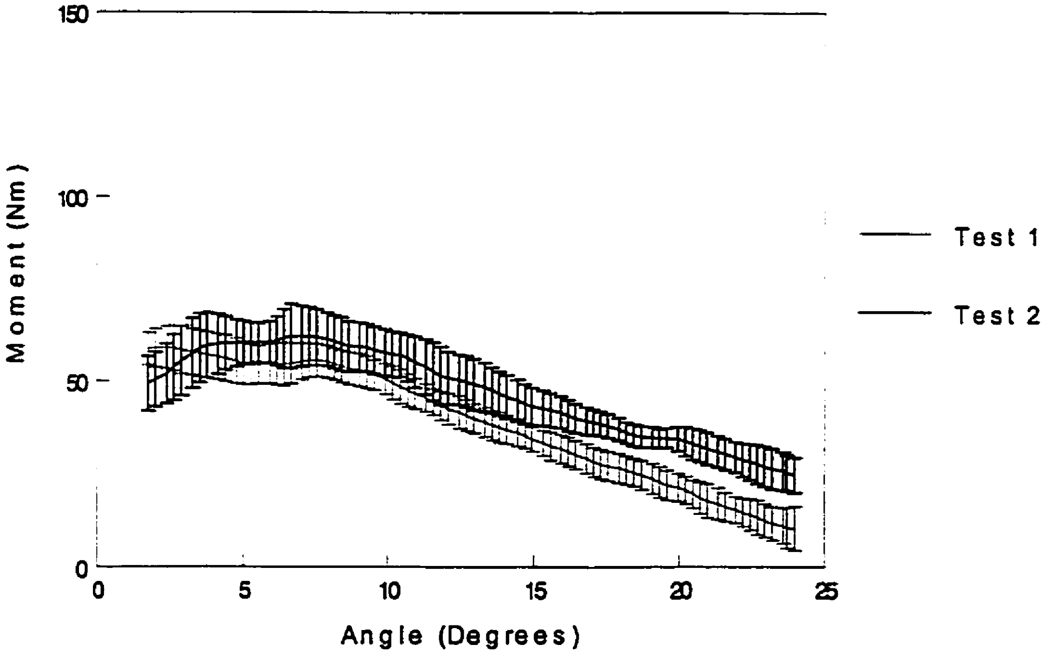
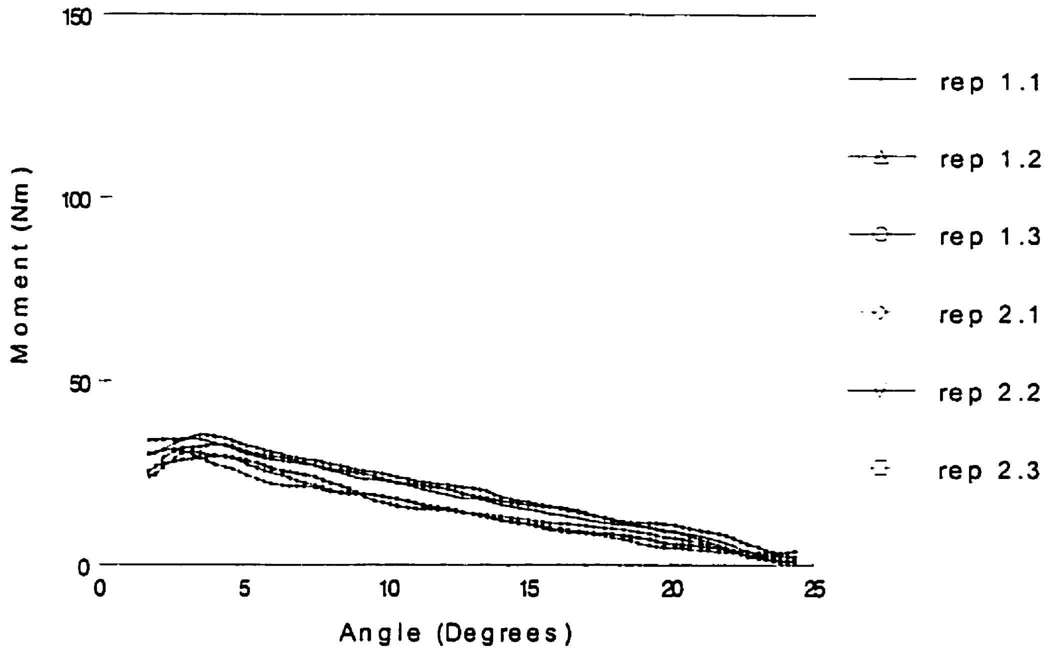
QUADRIPLEGIC PARTICIPANT Q4 – RAW DATA



MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q4 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants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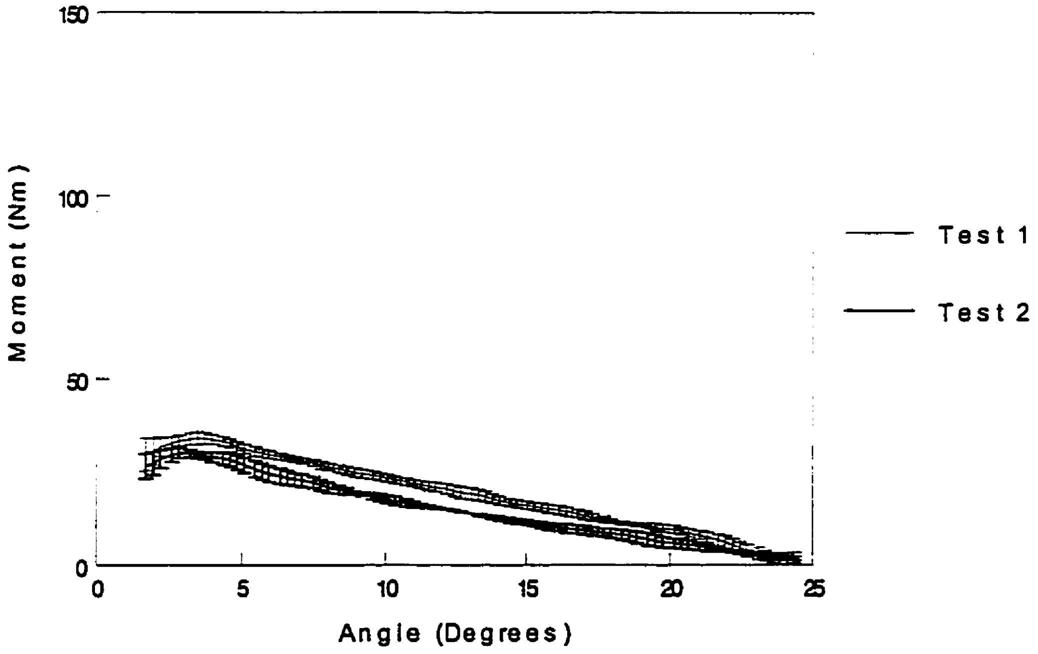
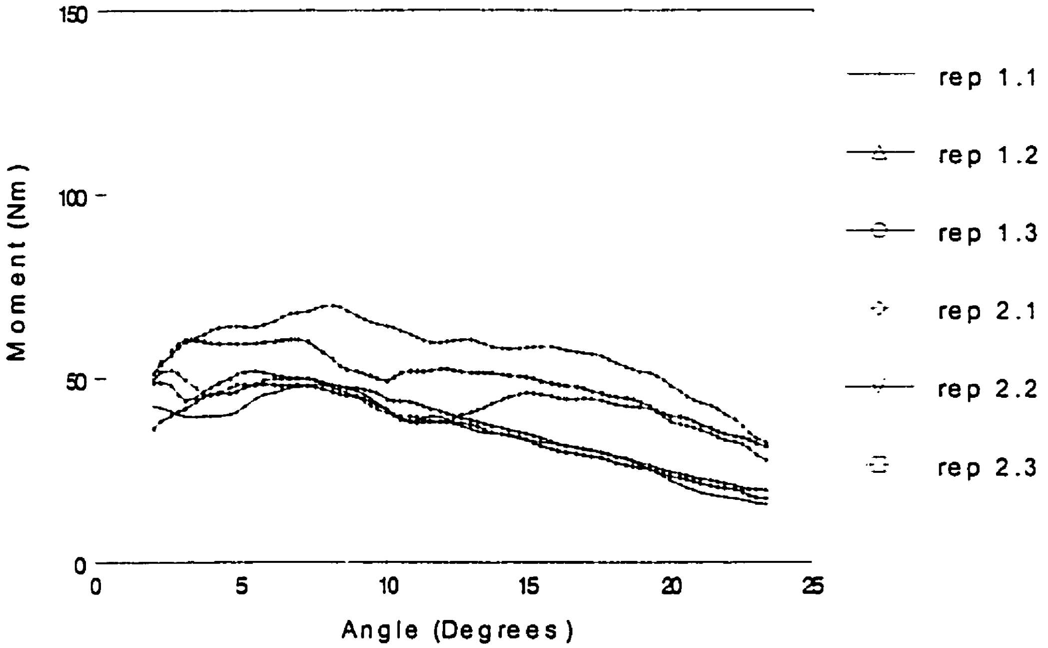
QUADRIPLEGIC PARTICIPANT Q5 – RAW DATA

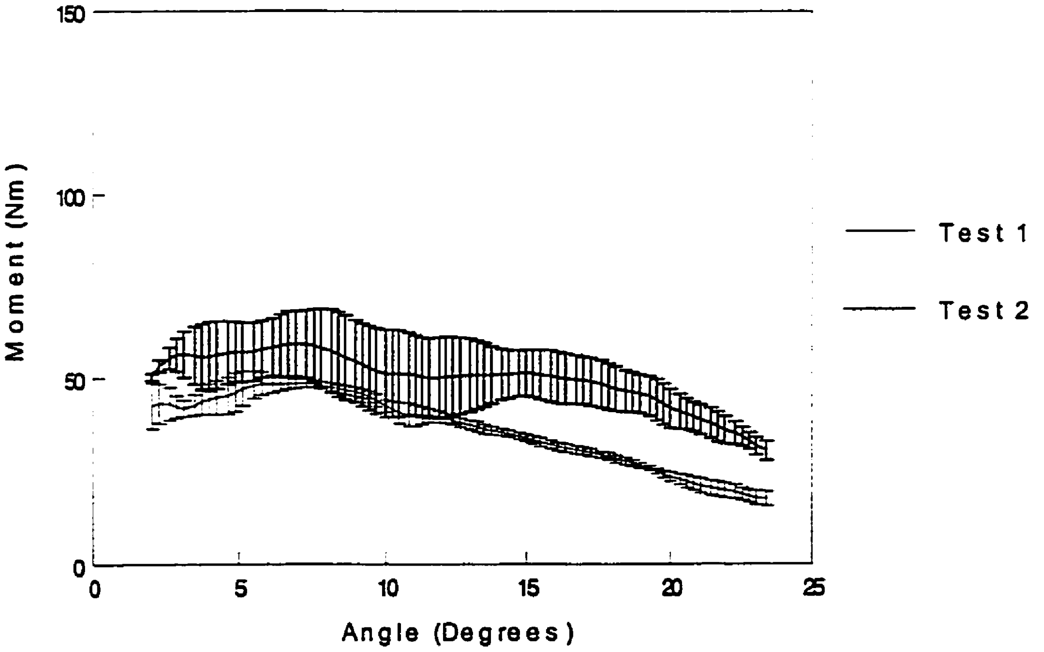

MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q5 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
QUADRIPLEGIC PARTICIPANT Q6 – RAW DATA

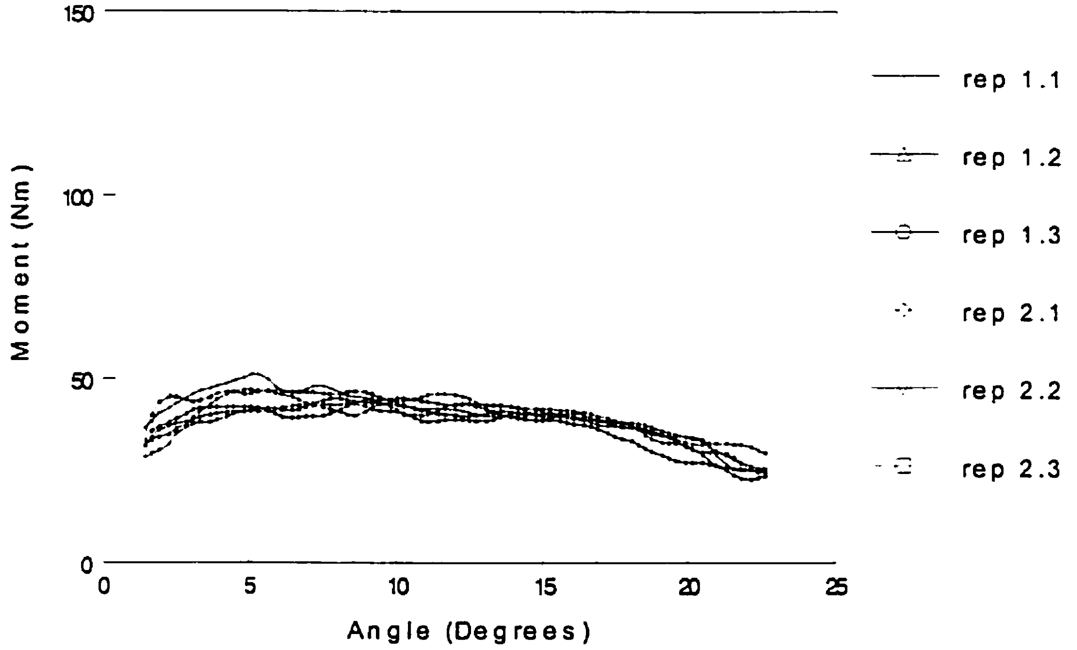


MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q6 – STANDARD DEVIATION

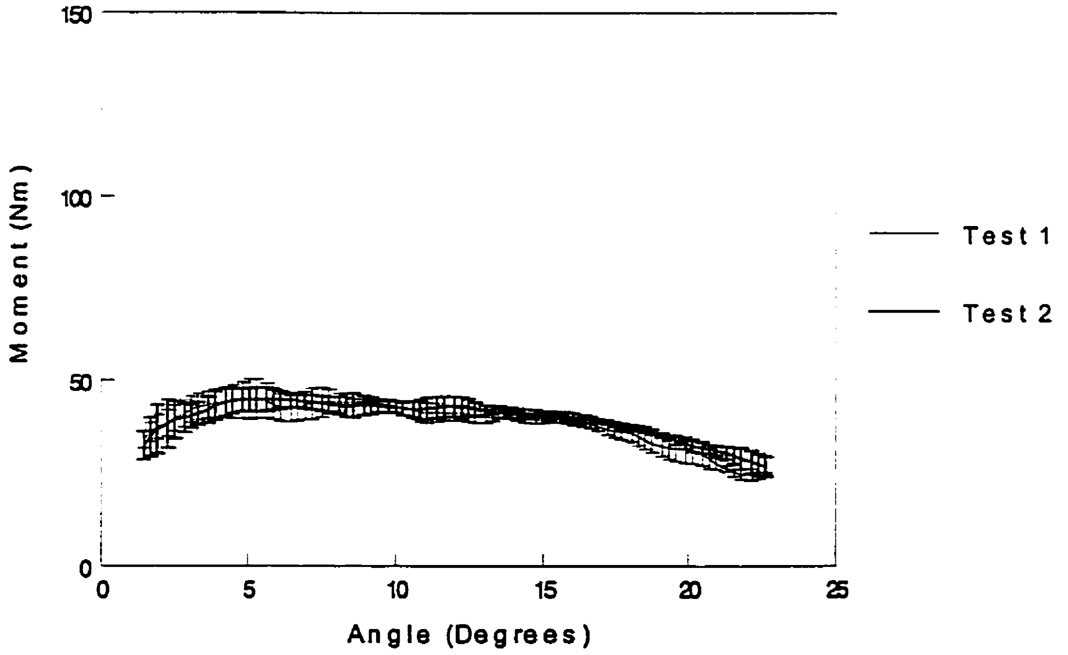


Appendix E – Moment / Angle Graphs for Individual Participants

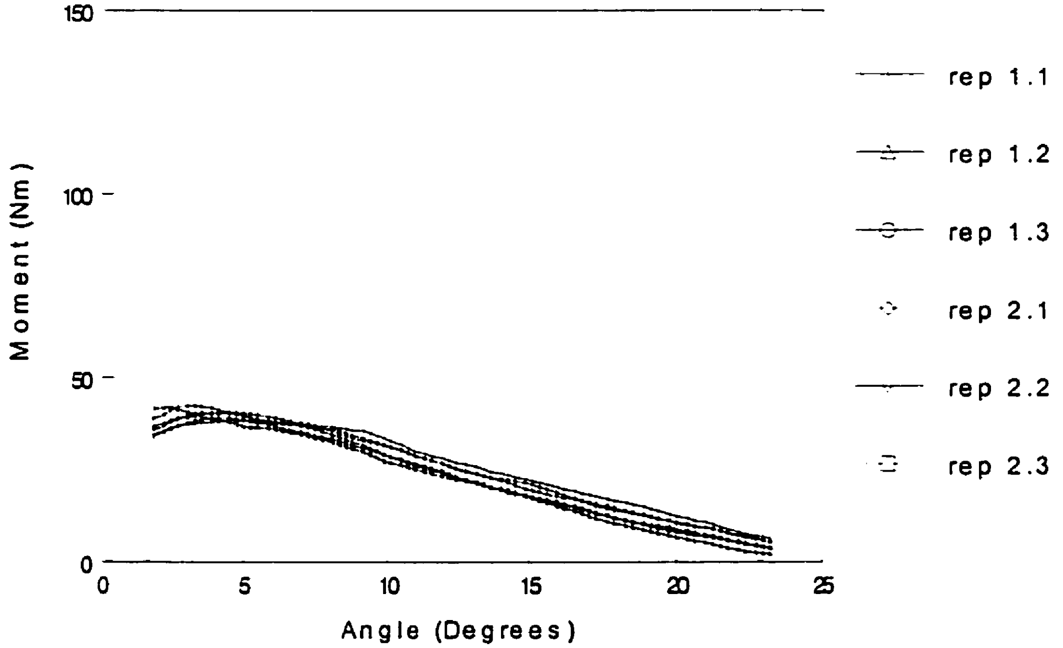
FORCE GENERATED DURING SCAPULAR ELEVATION
QUADRIPLEGIC PARTICIPANT Q7 – RAW DATA



MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q7 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants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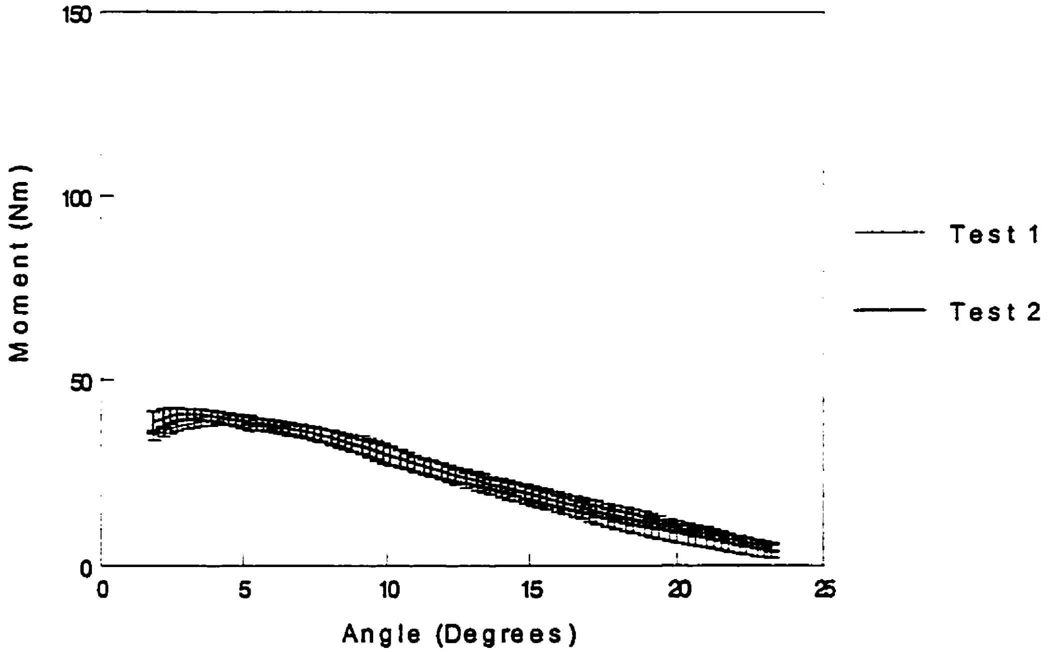
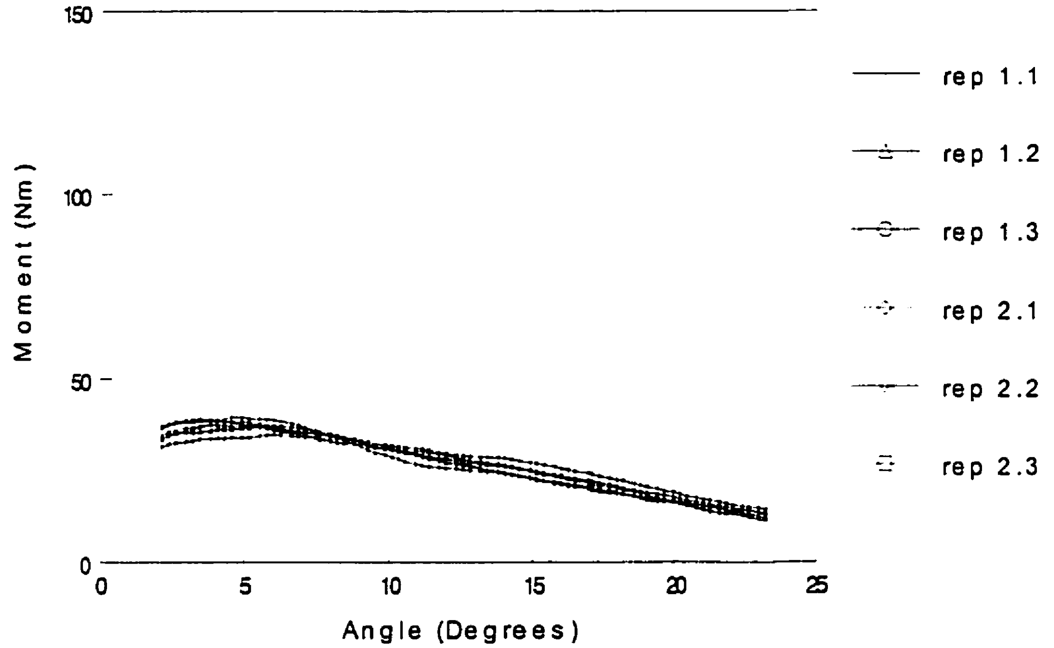
QUADRIPLEGIC PARTICIPANT Q8 – RAW DATA


MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q8 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
QUADRIPLEGIC PARTICIPANT Q9 – RAW DATA



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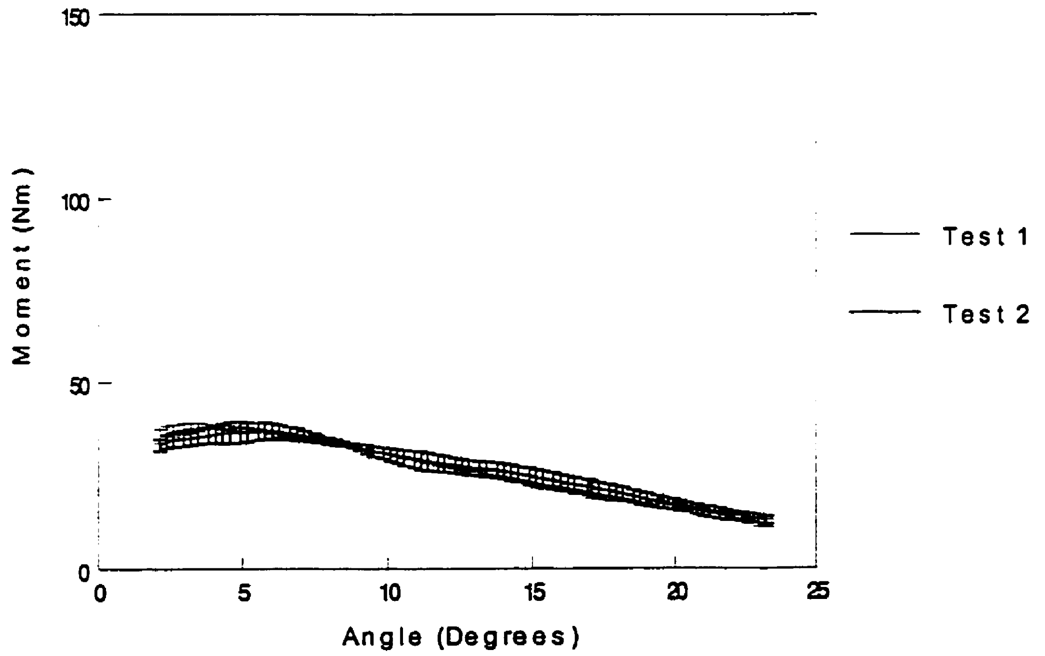
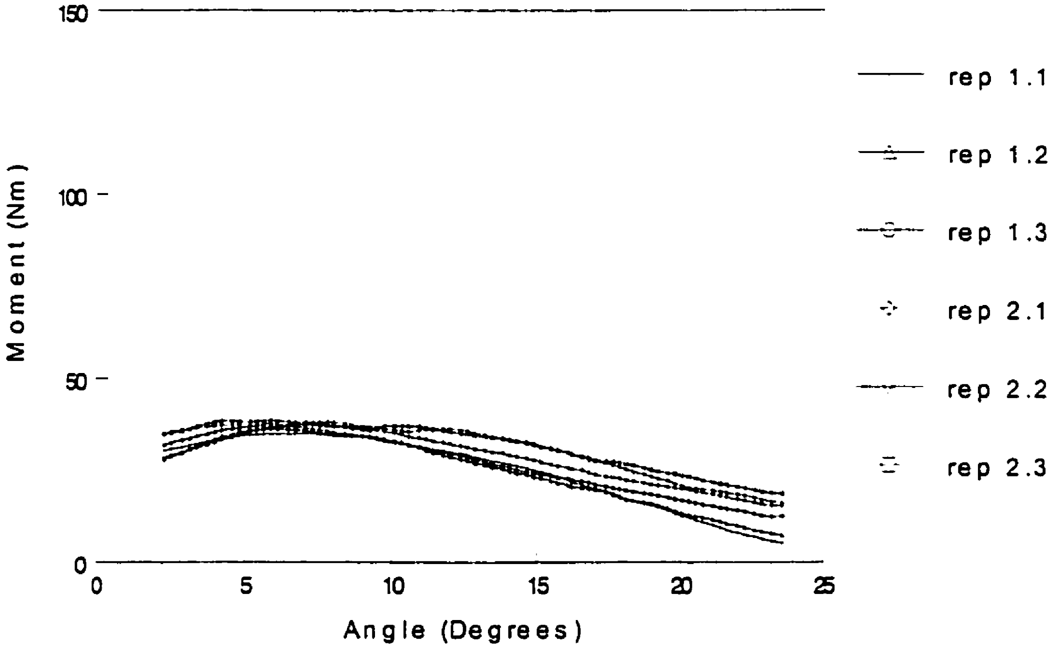
Q9 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants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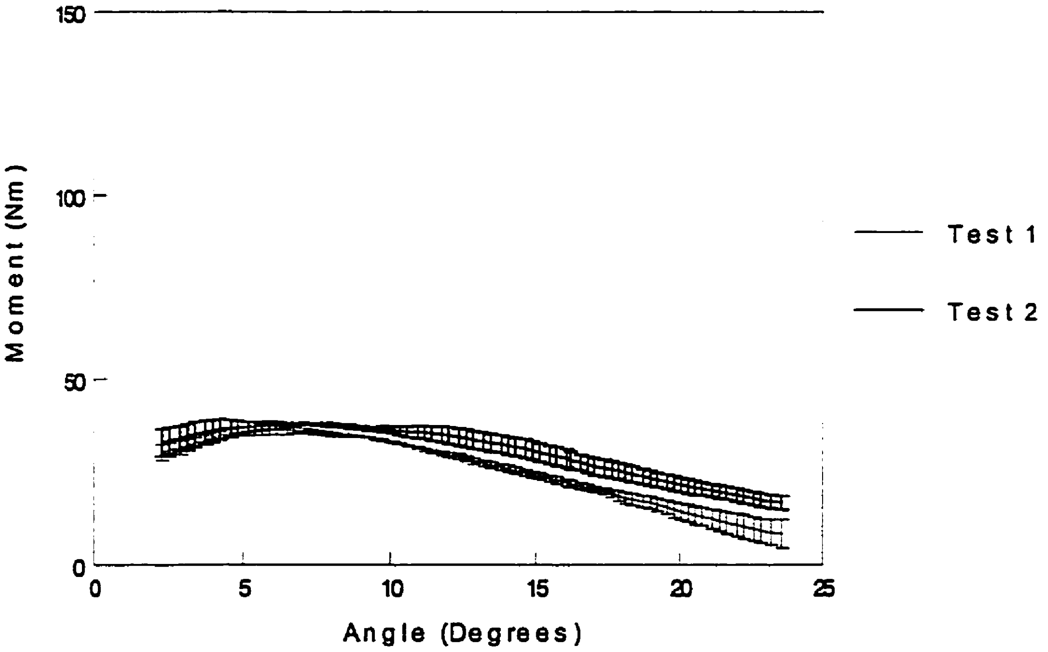
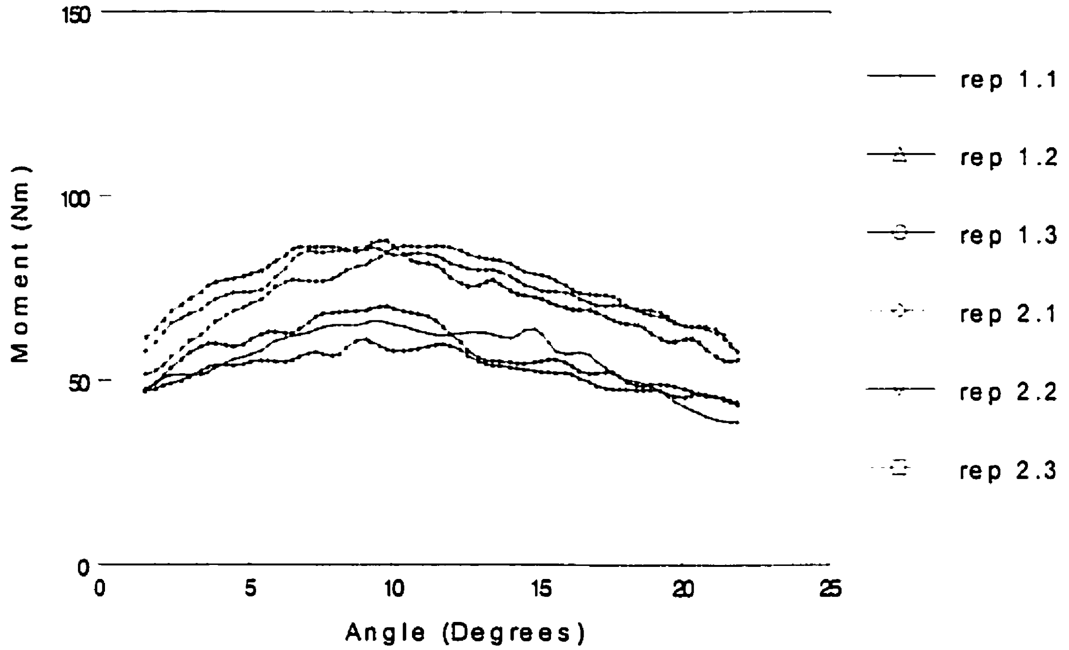
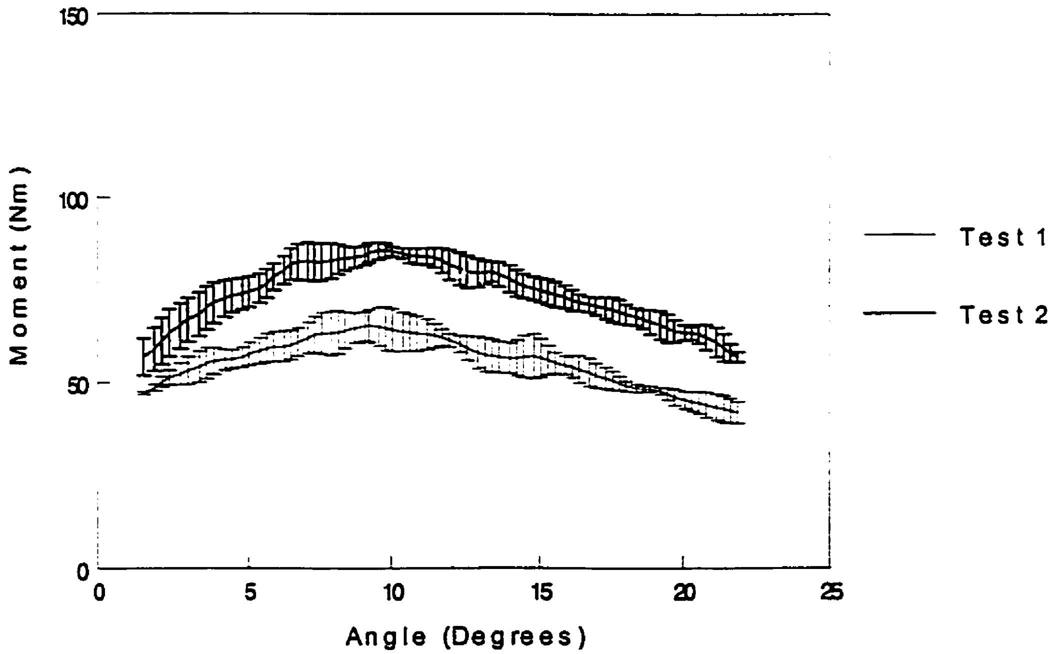
QUADRIPLEGIC PARTICIPANT Q10 – RAW DATA


MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
Q10 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants



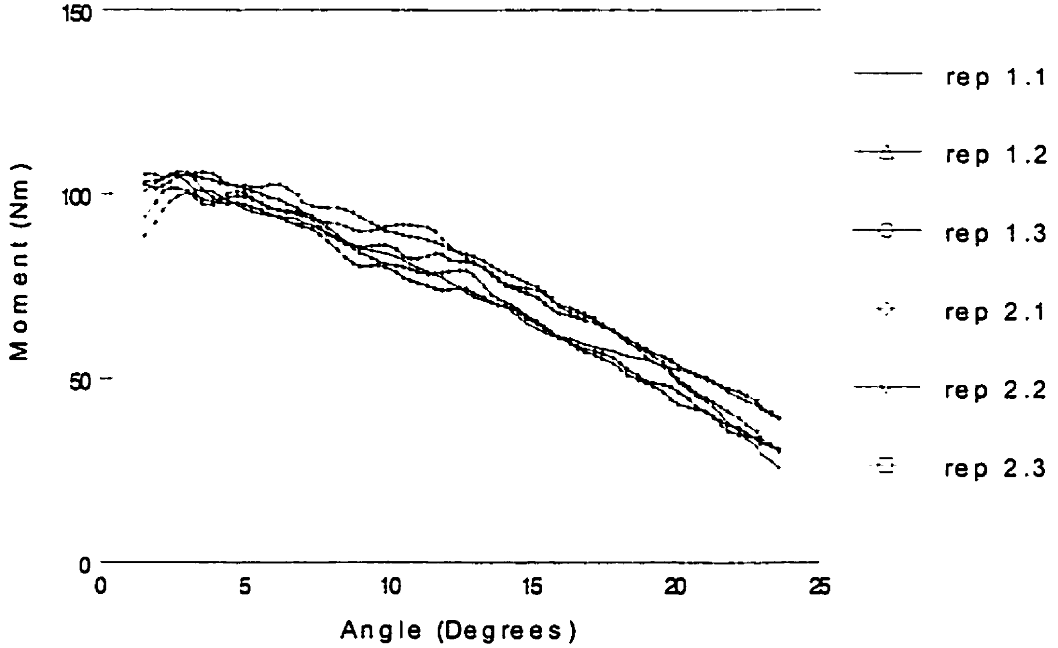
FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P1 – RAW DATA



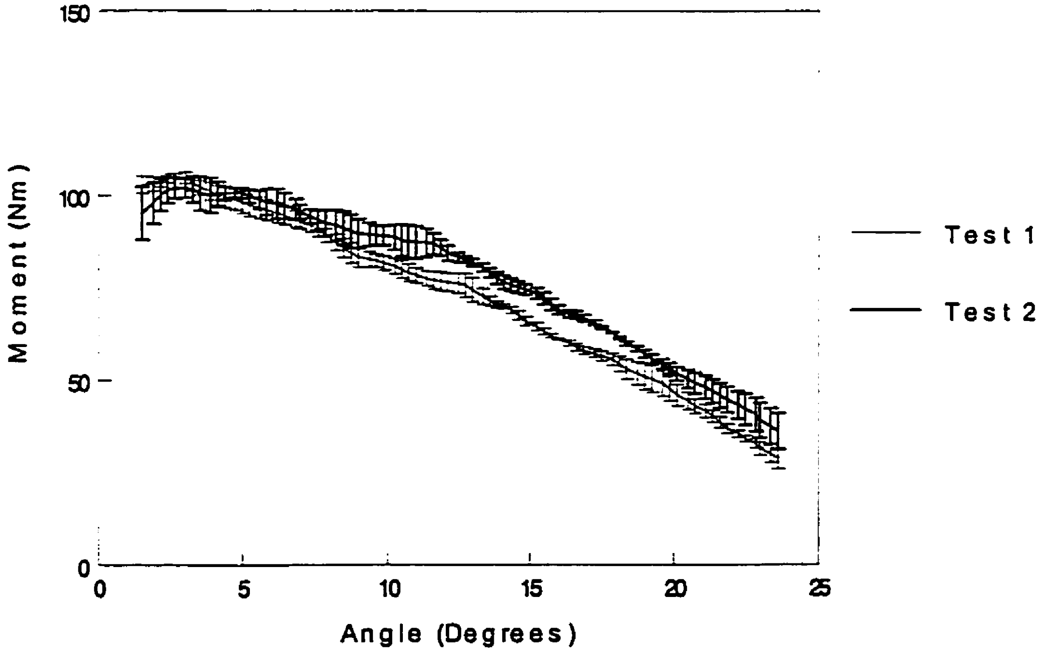
MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
P1 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P2 – RAW DATA

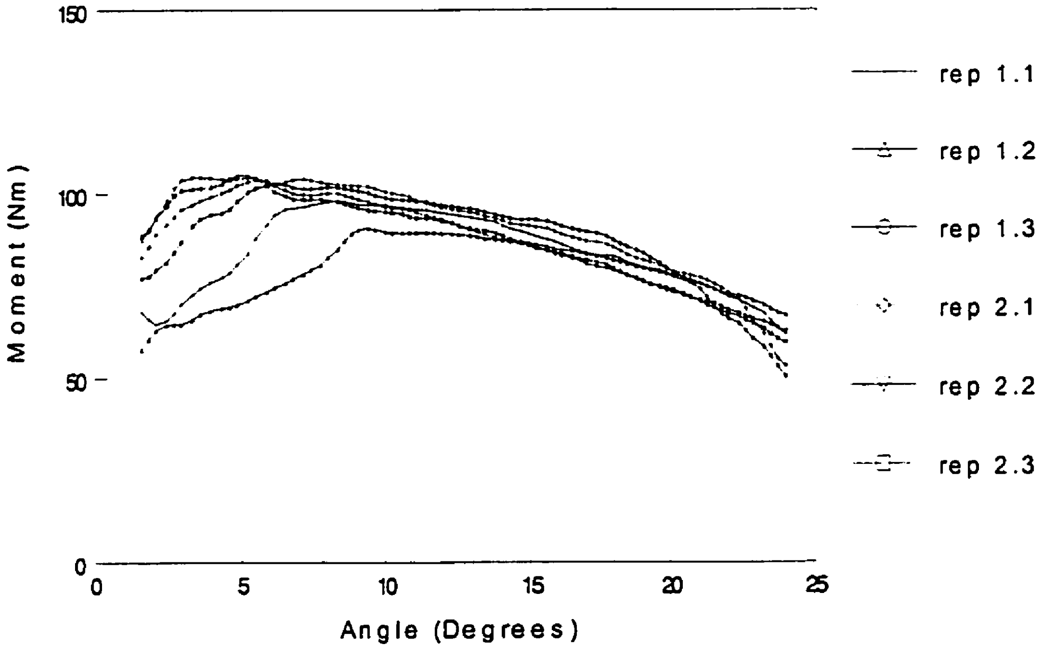


MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
P2 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P3 – RAW DATA



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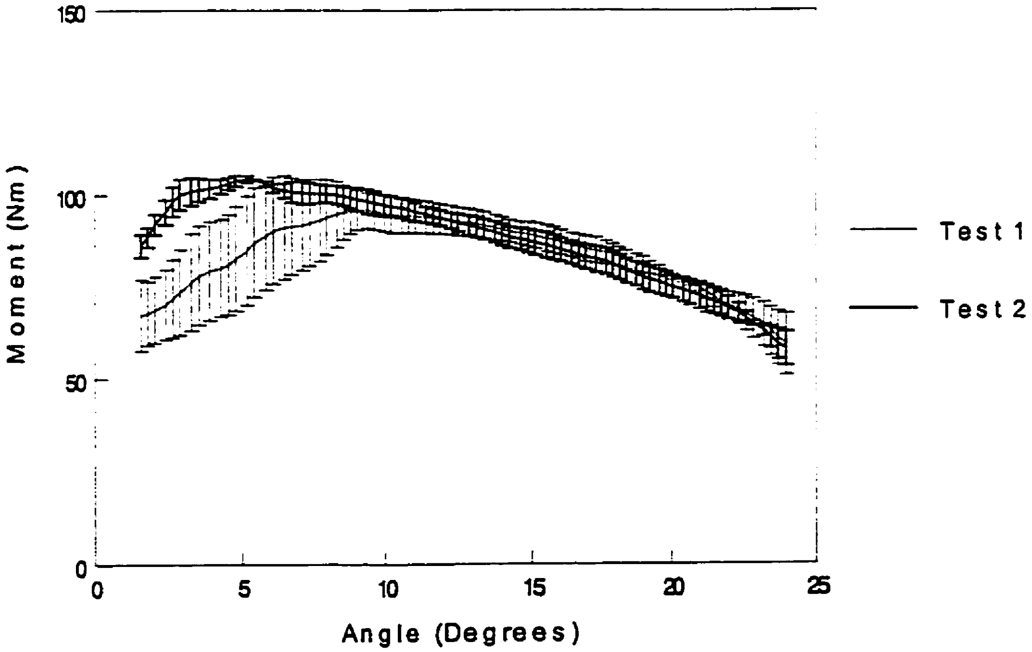
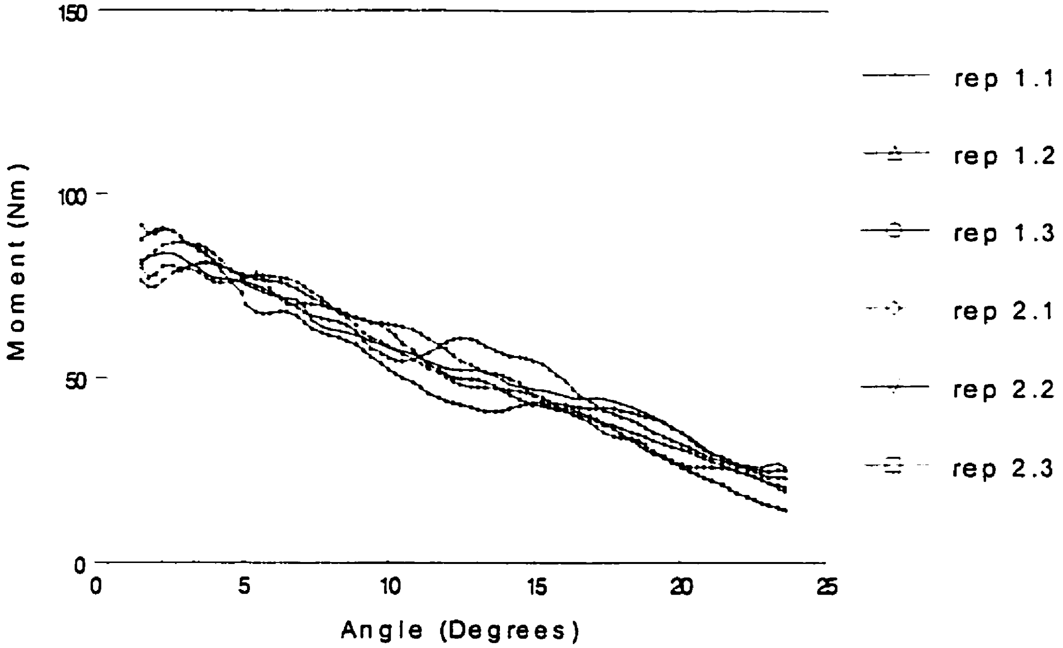
P3 – STANDARD DEVIATION

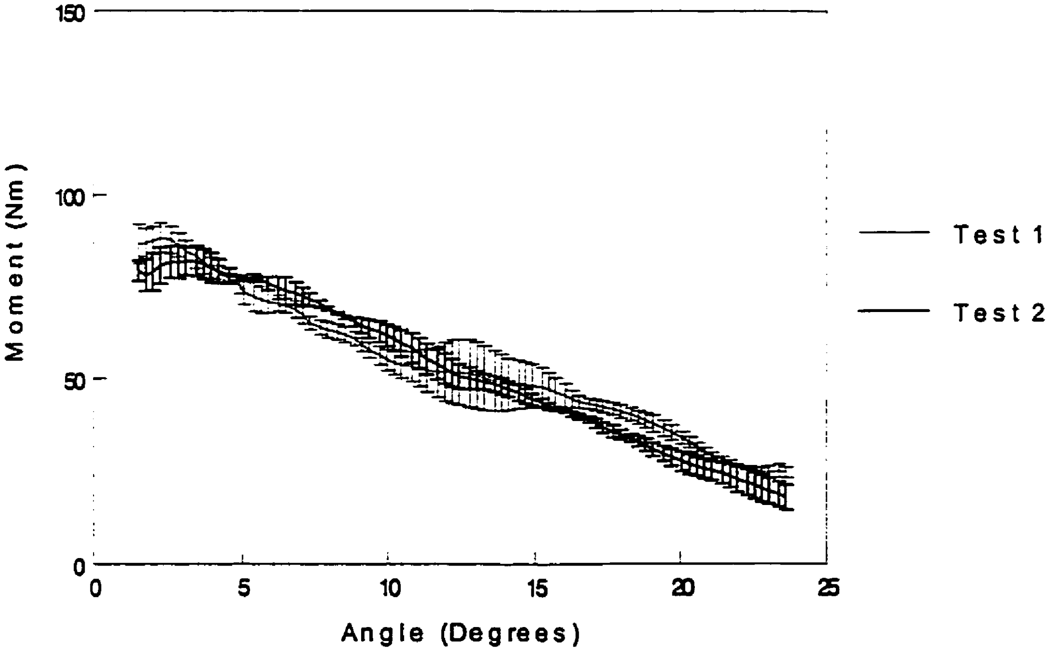


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P4 – RAW DATA

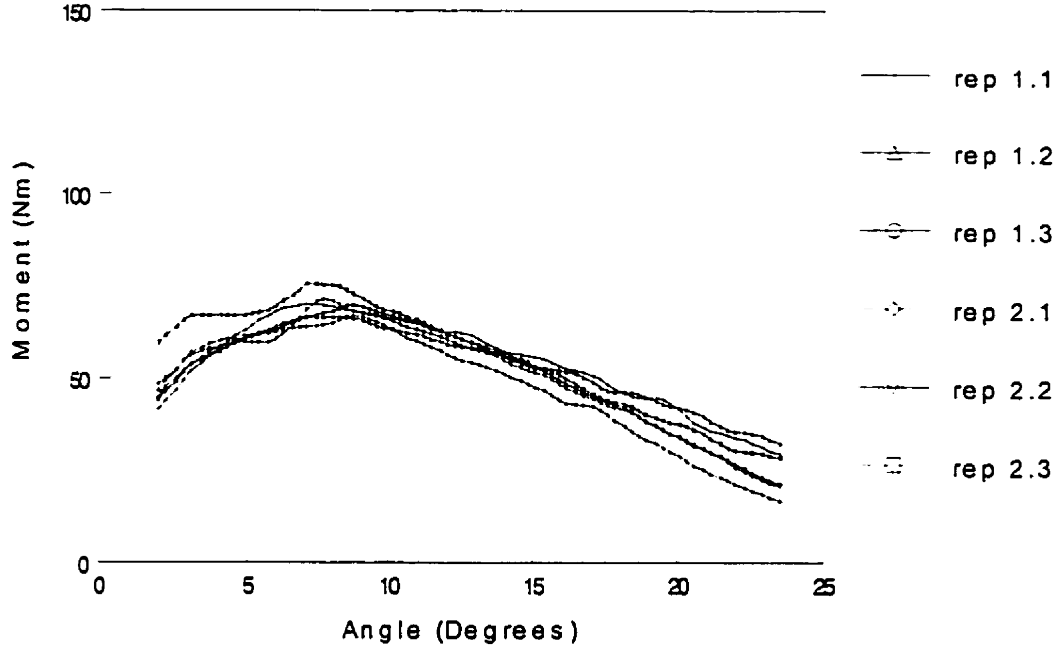


MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
P4 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P5 – RAW DATA



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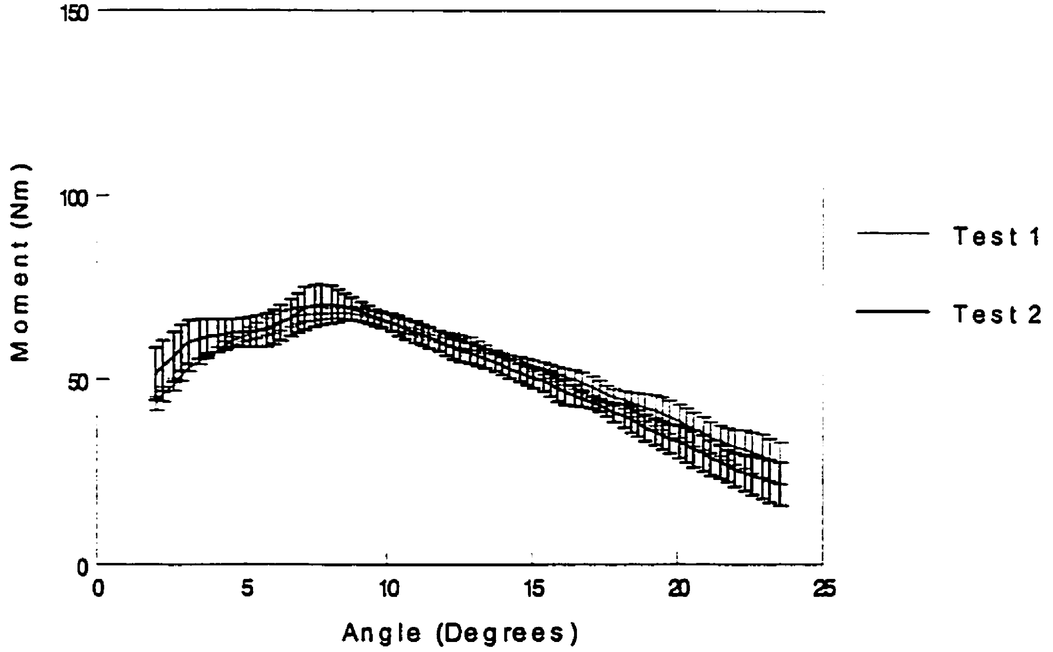
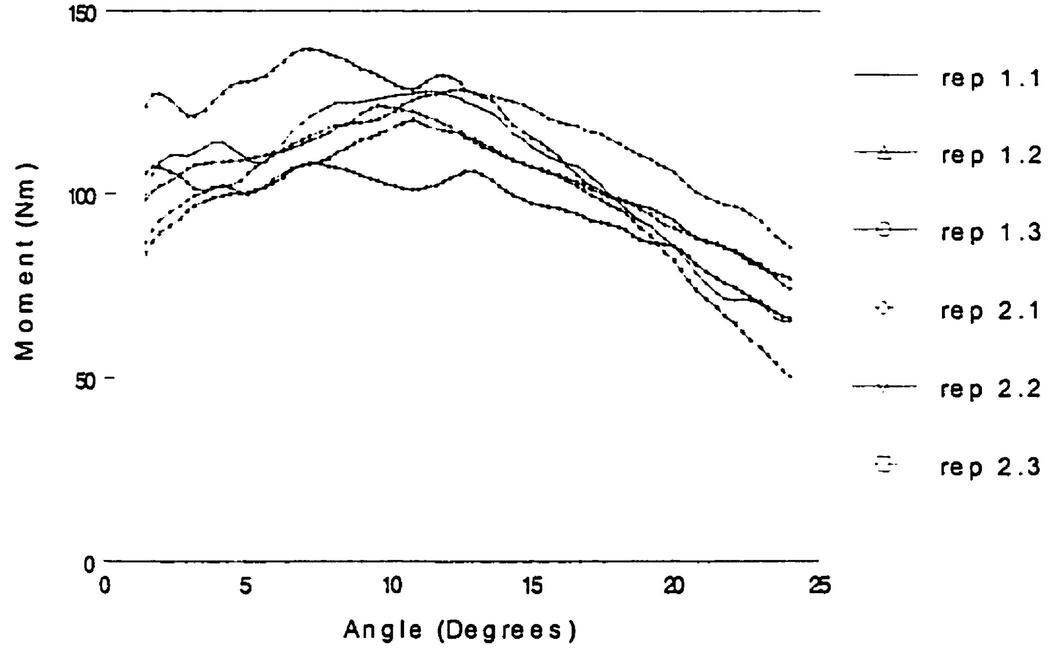
P5 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P6 – RAW DATA



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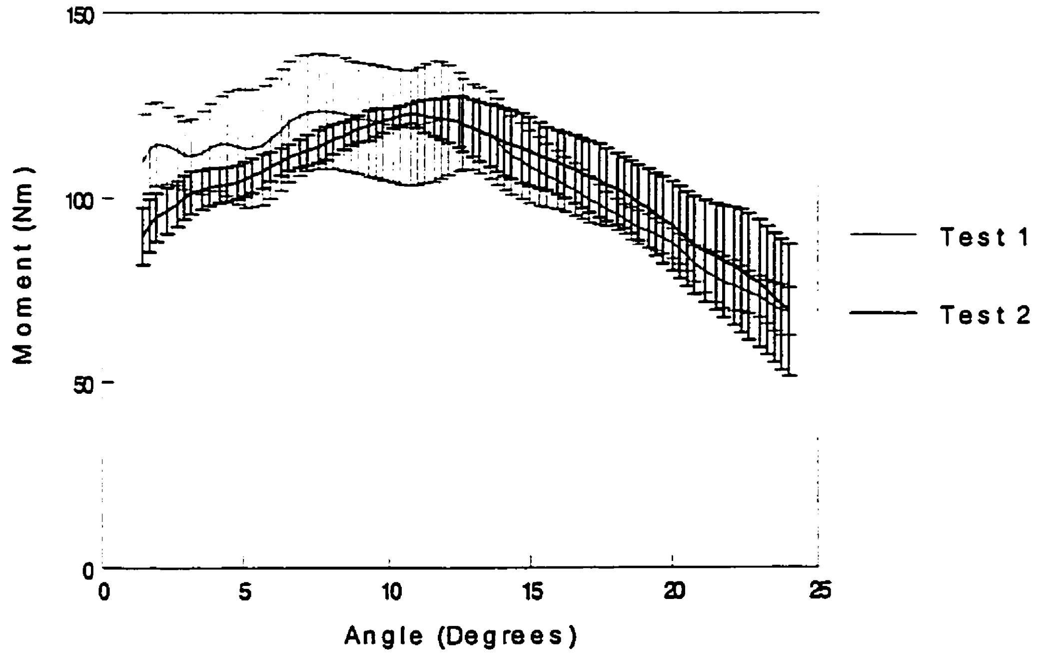
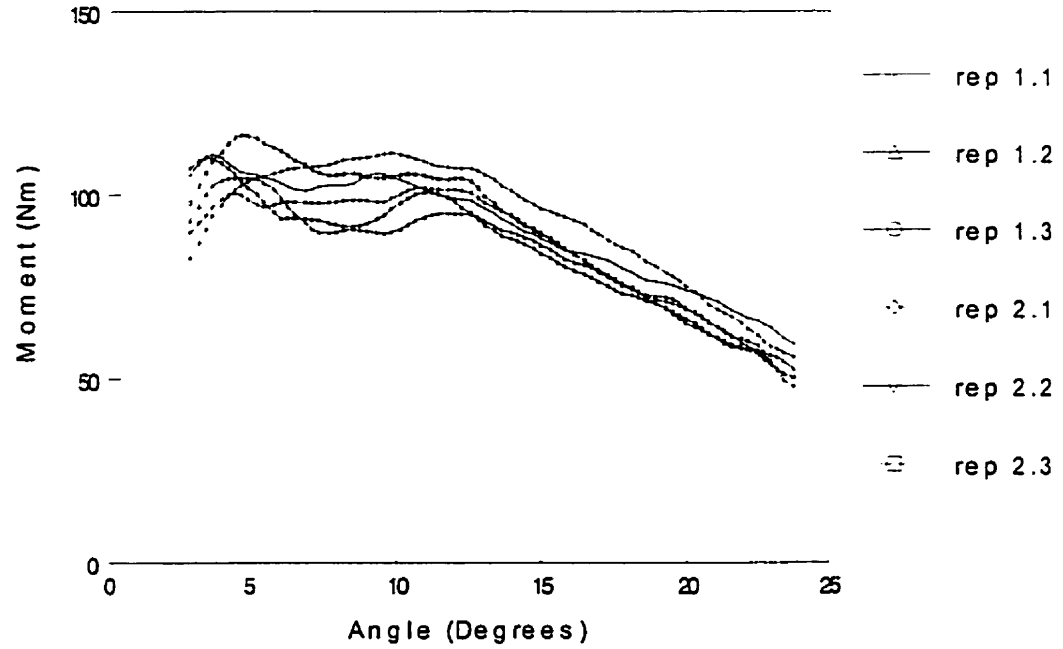
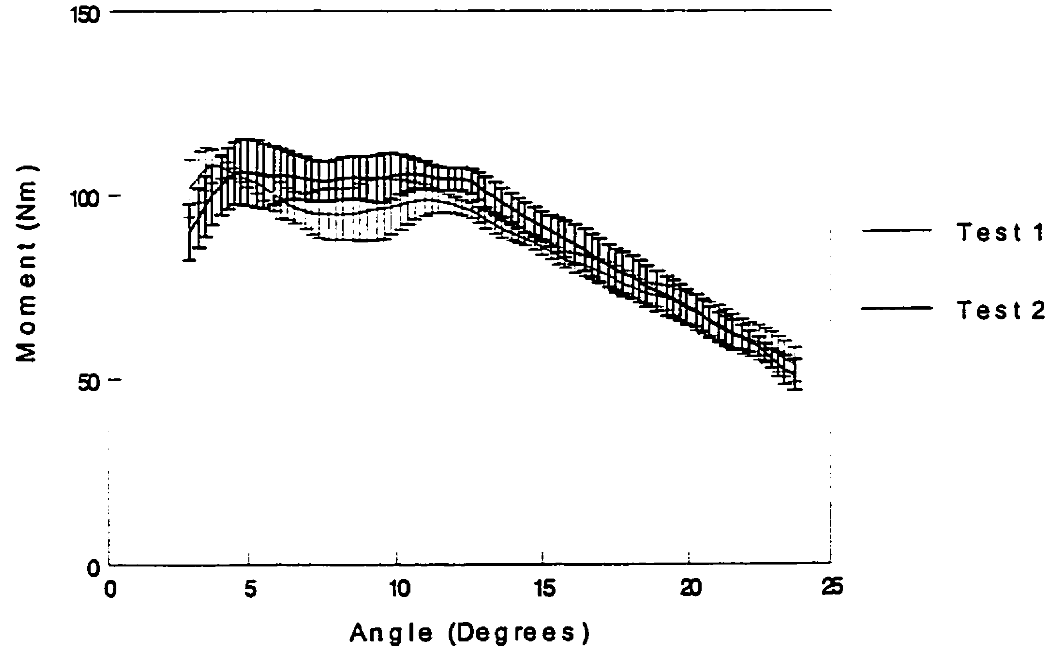
P6 – STANDARD DEVIATION



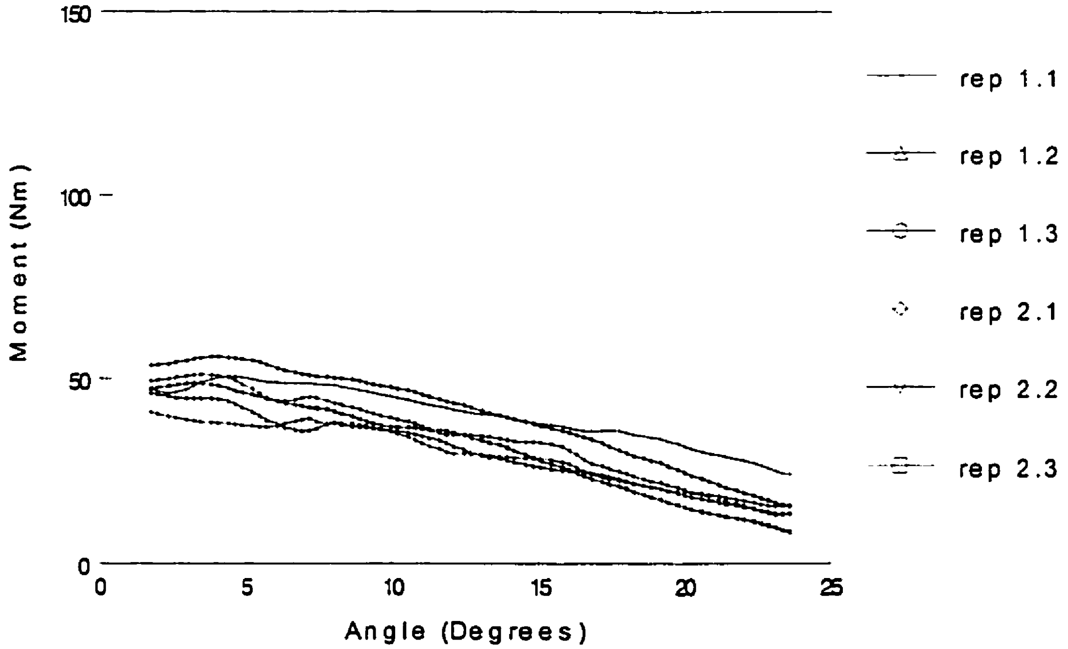
Appendix E – Moment / Angle Graphs for Individual Participants


FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P7 – RAW DATA

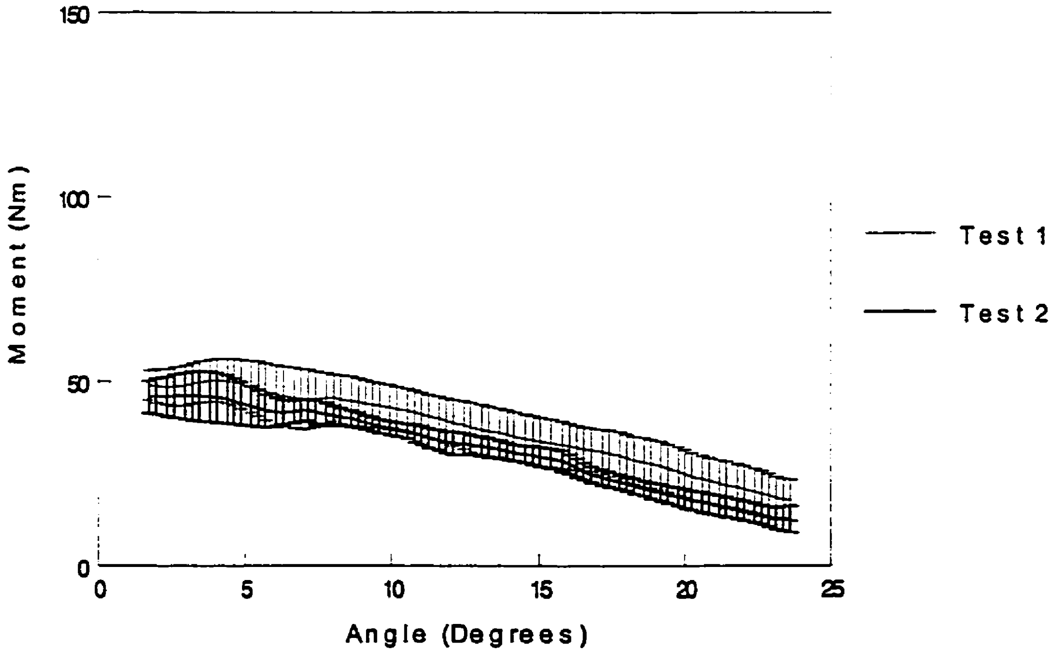

MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
P7 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P8 – RAW DATA

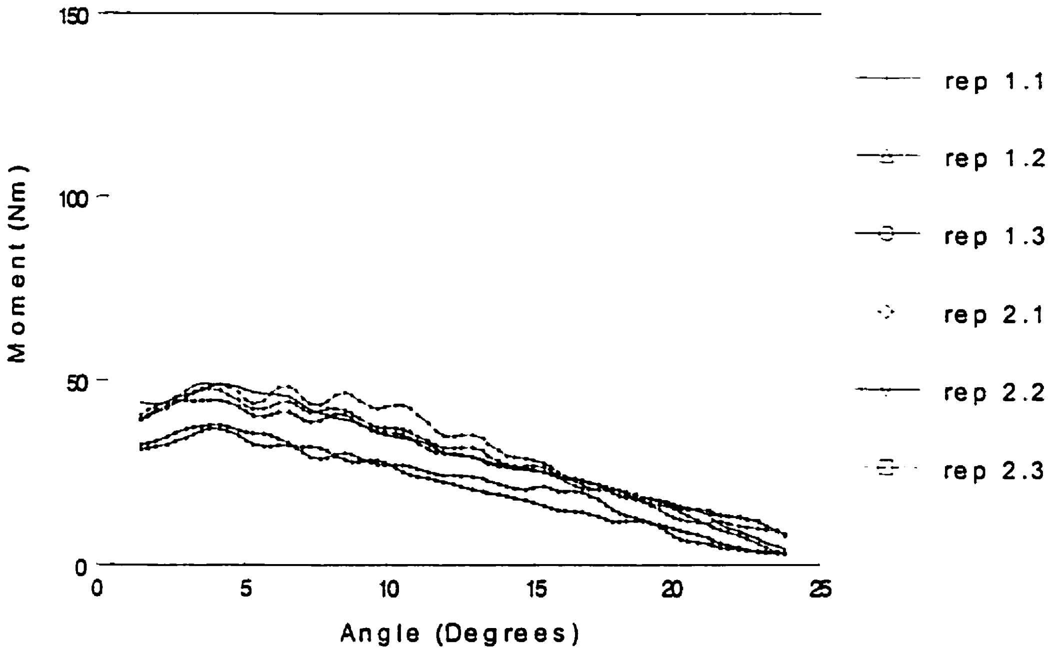


MEAN FORCE GENERATED DURING SCAPULAR ELEVATION
P8 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR ELEVATION
PARAPLEGIC PARTICIPANT P9 – RAW DATA



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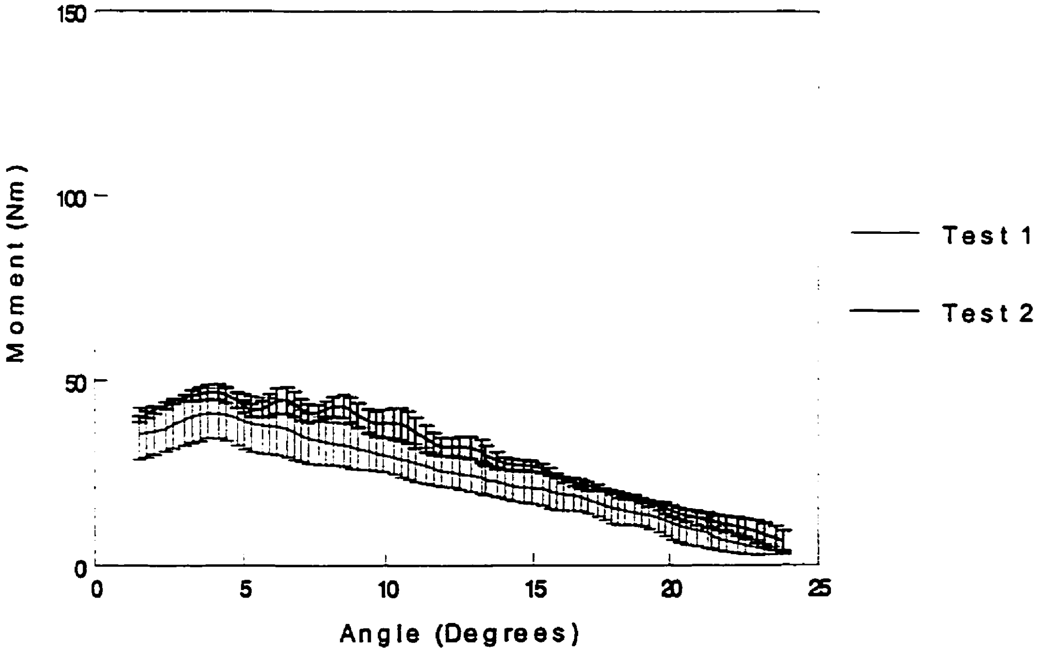
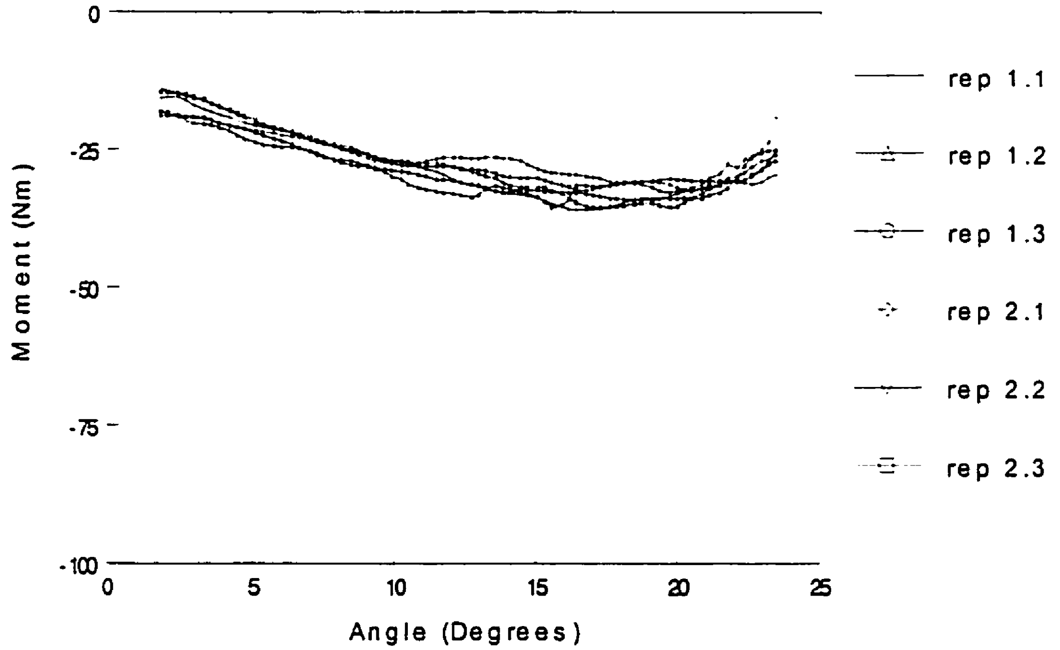
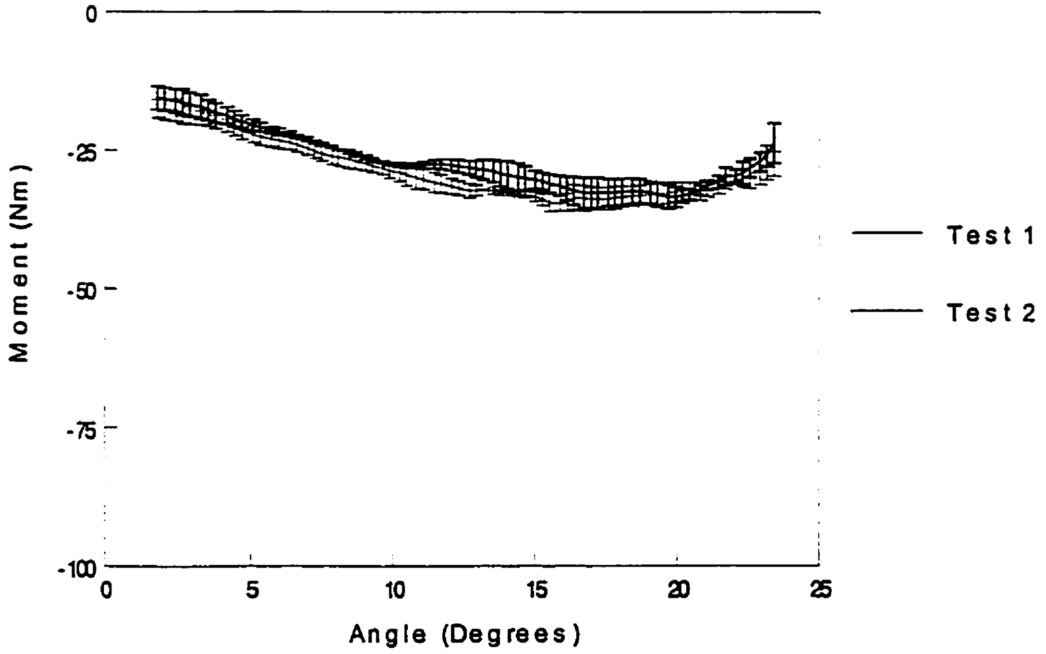
P9 – STANDARD DEVIATION

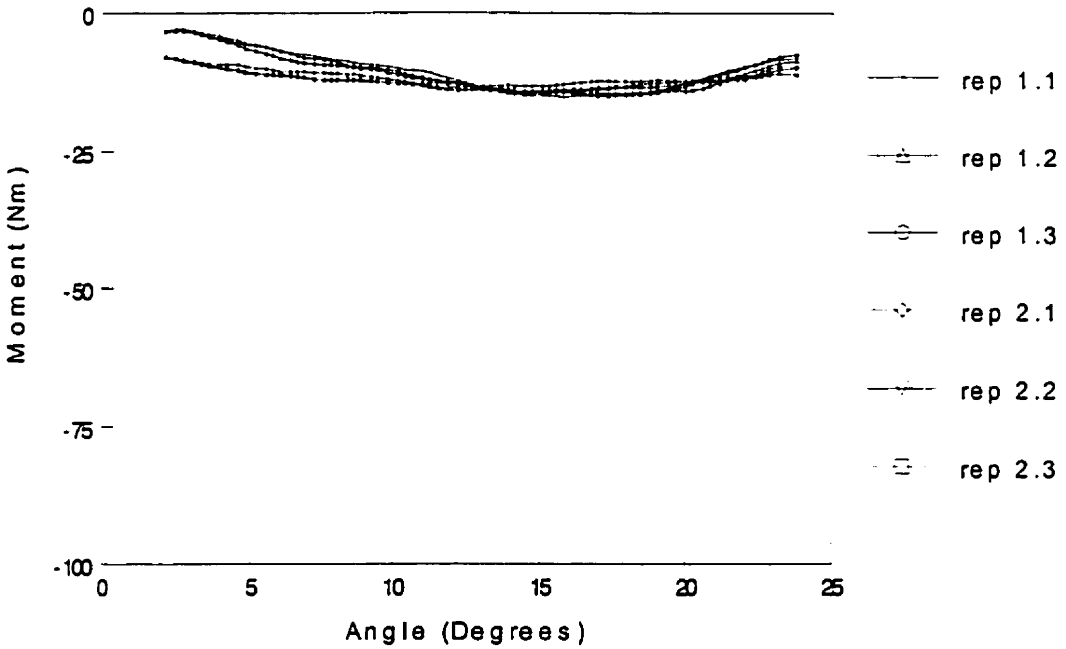


Appendix E – Moment / Angle Graphs for Individual Participants

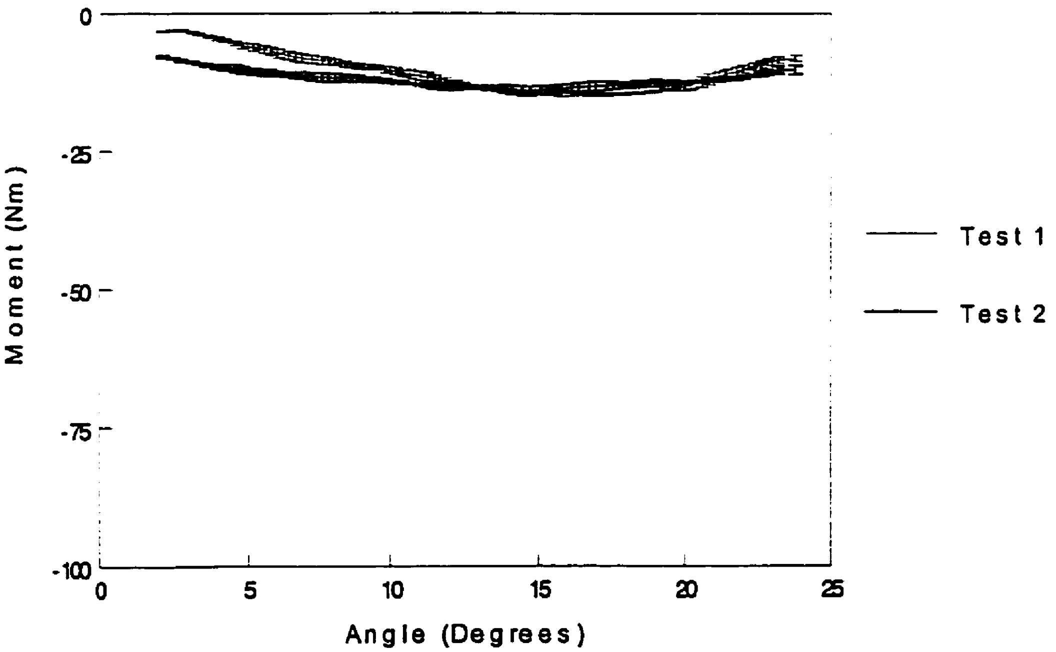

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q1 – RAW DATA


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
Q1 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants



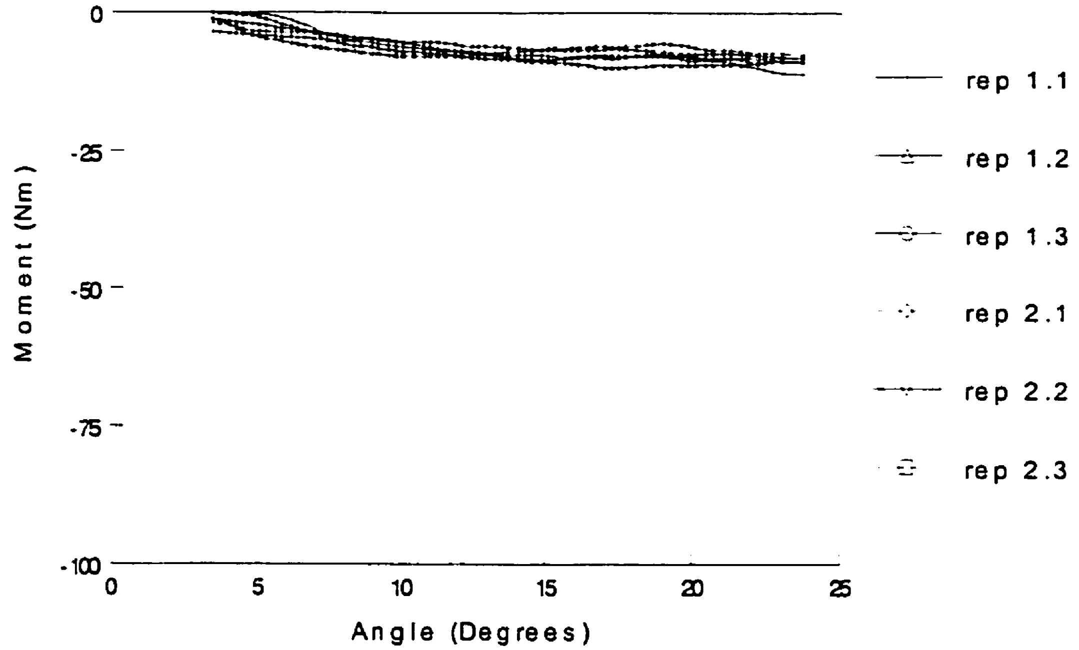
FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q2 – RAW DATA



MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
Q2 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q3 – RAW DATA



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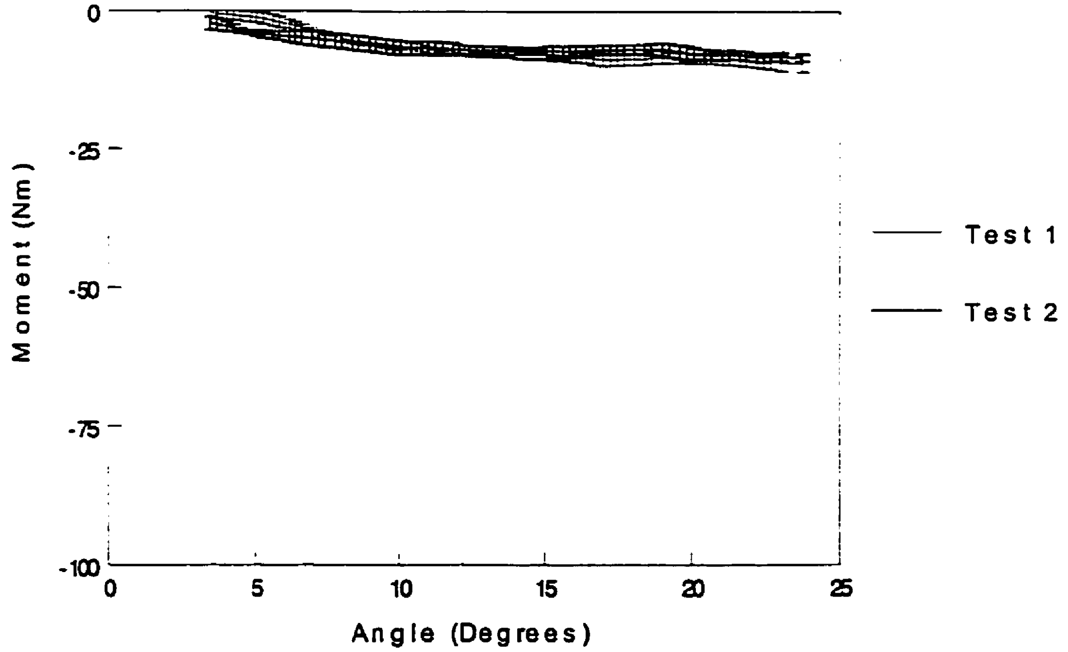
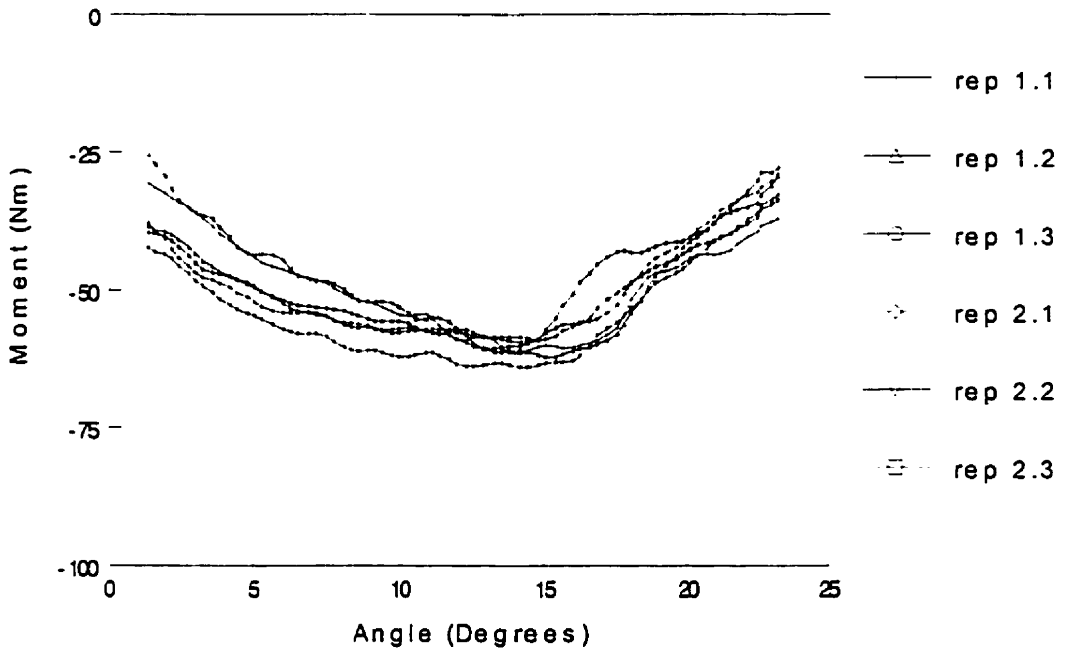
Q3 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q4 – RAW DATA



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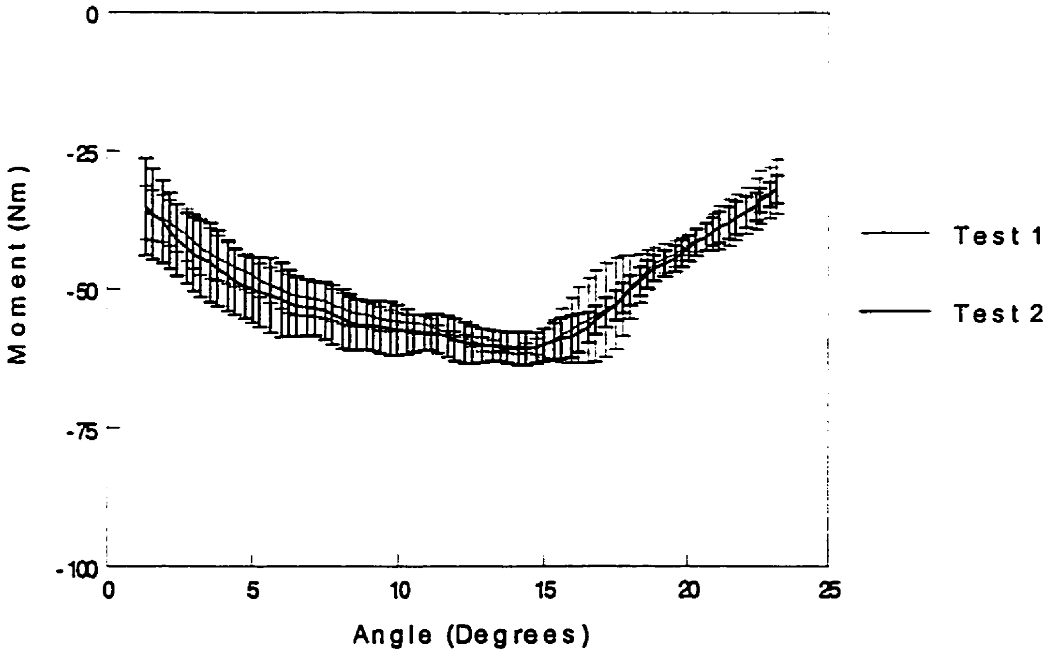
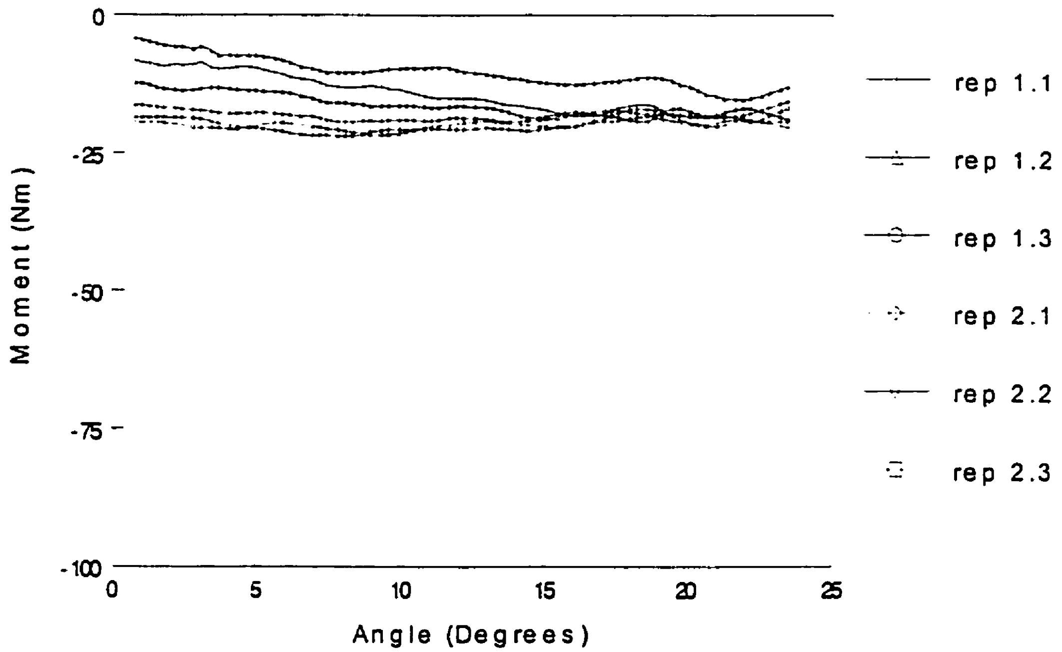
Q4 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q5 – RAW DATA



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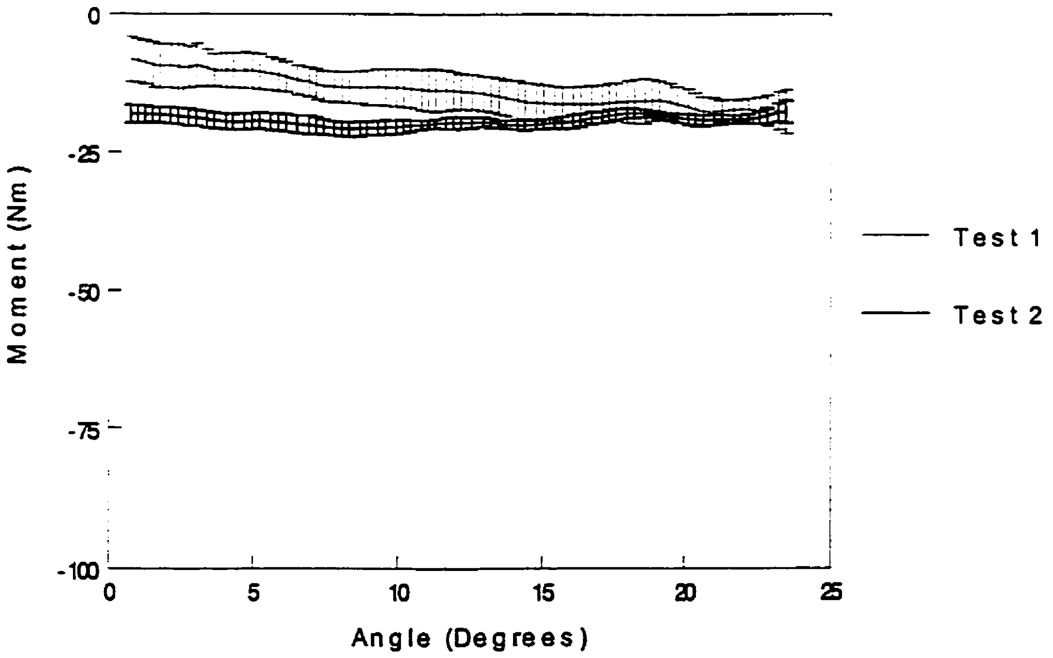
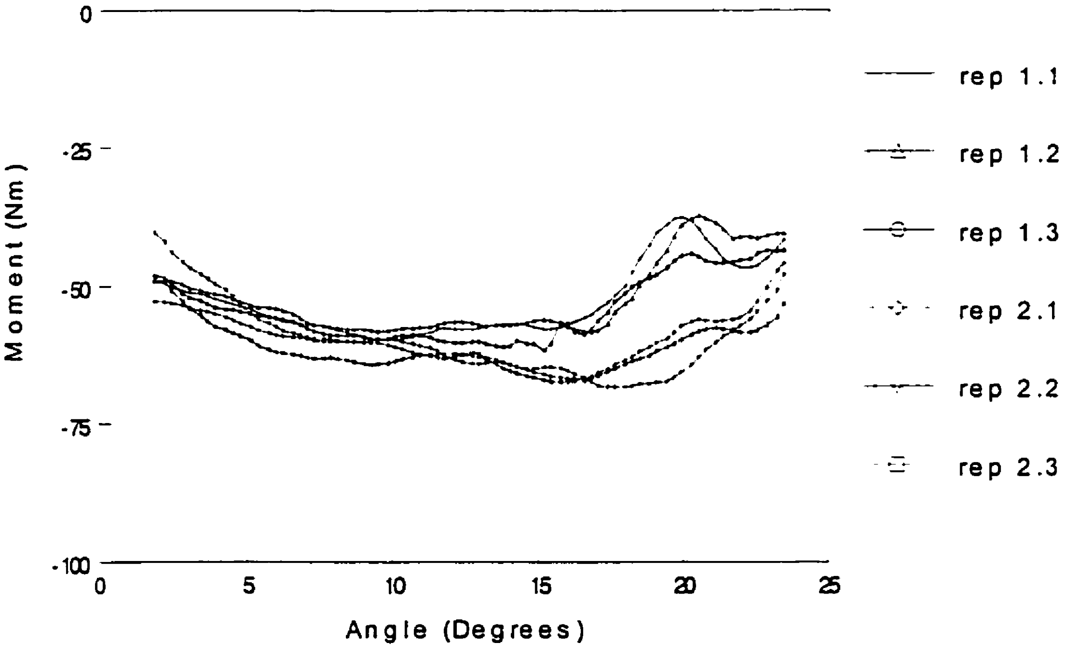
Q5 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants



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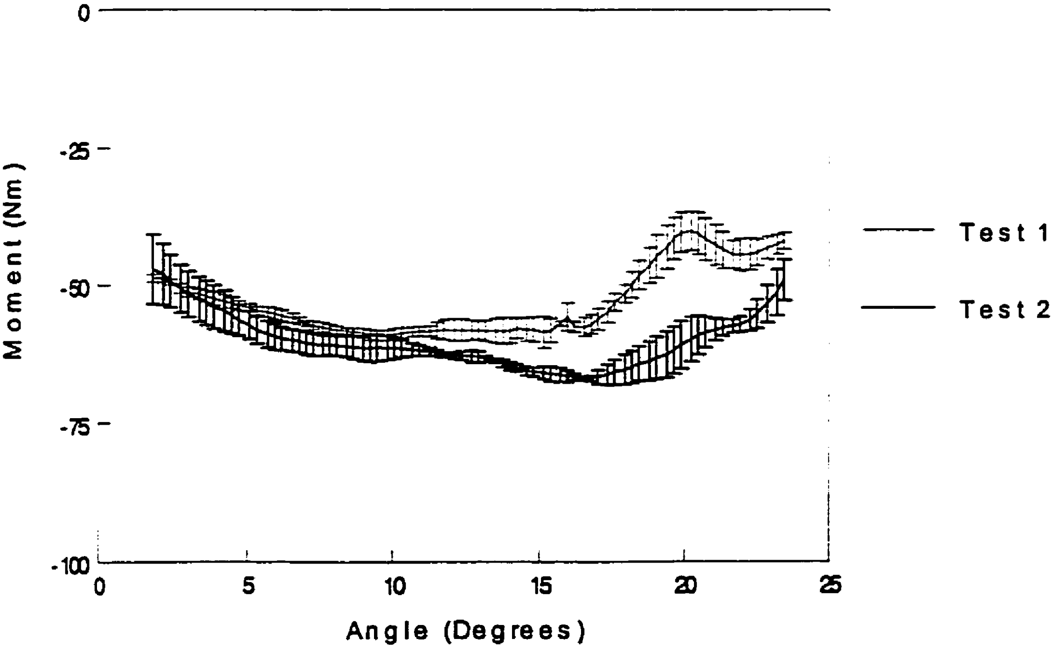
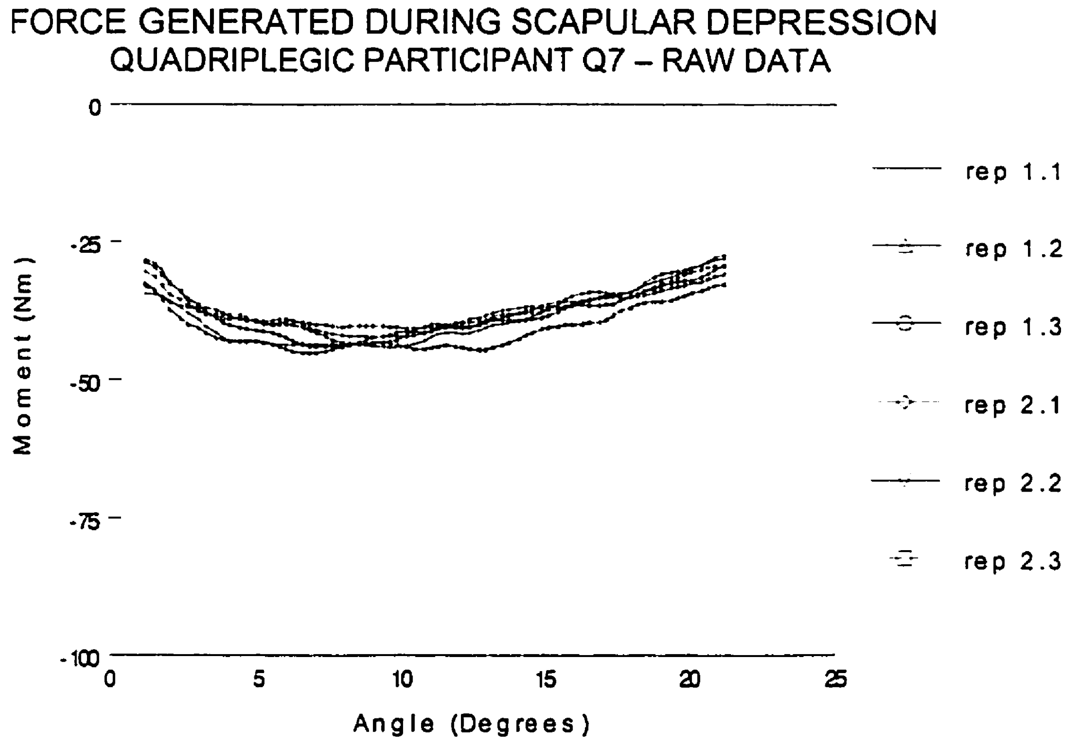
QUADRIPLEGIC PARTICIPANT Q6 – RAW DATA

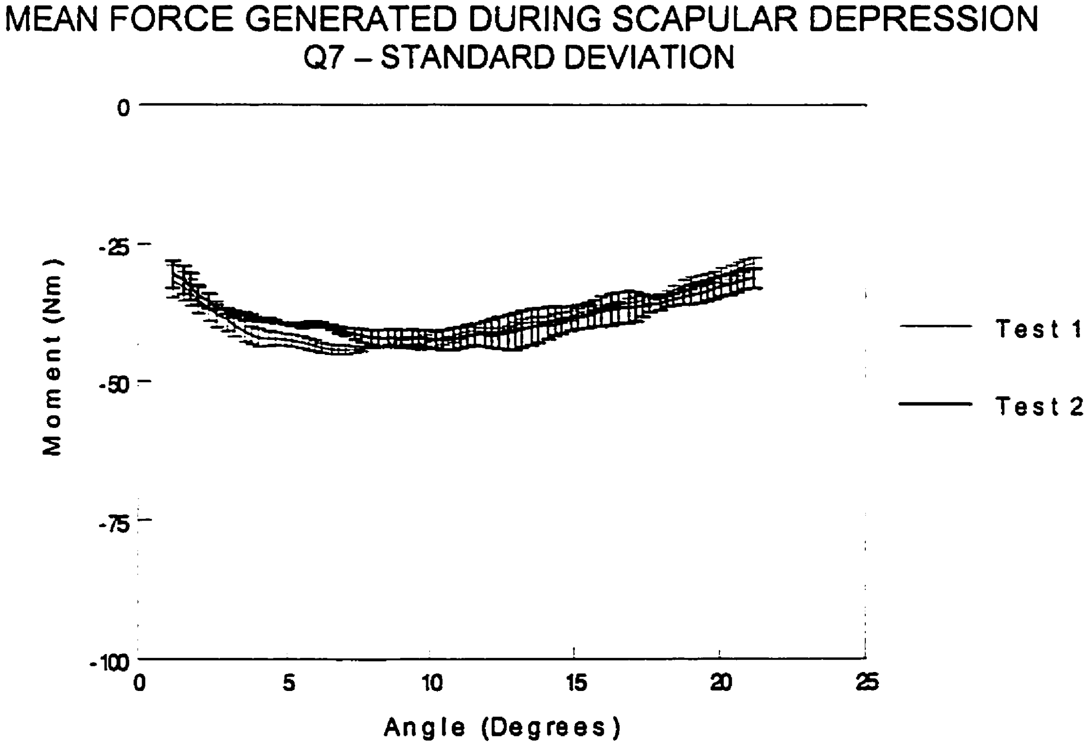


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
Q6 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants



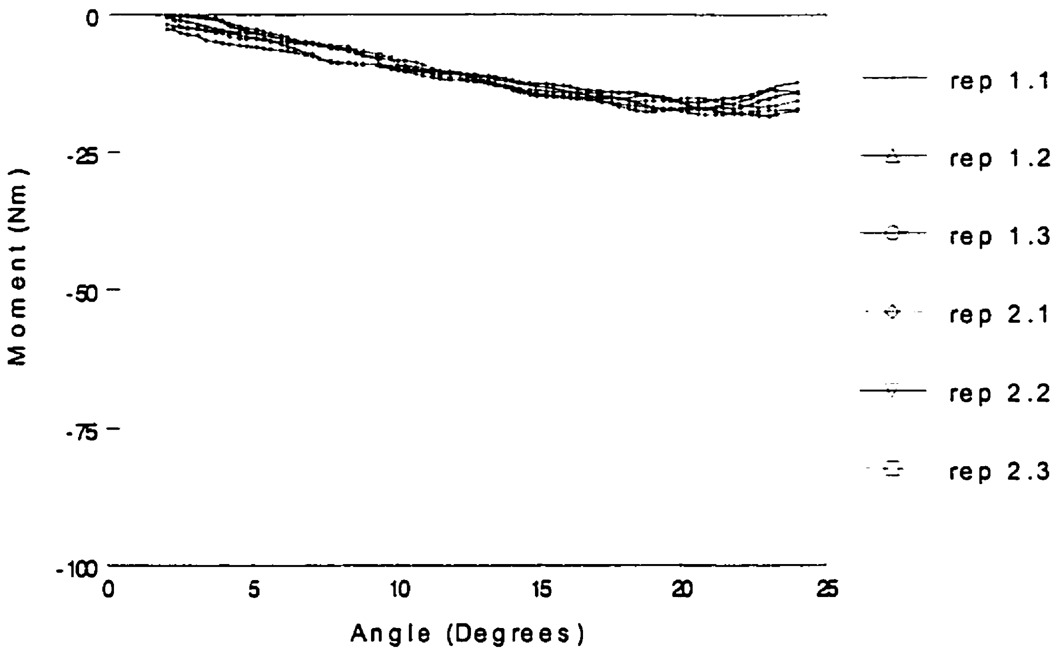
FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q7 – RAW DATA



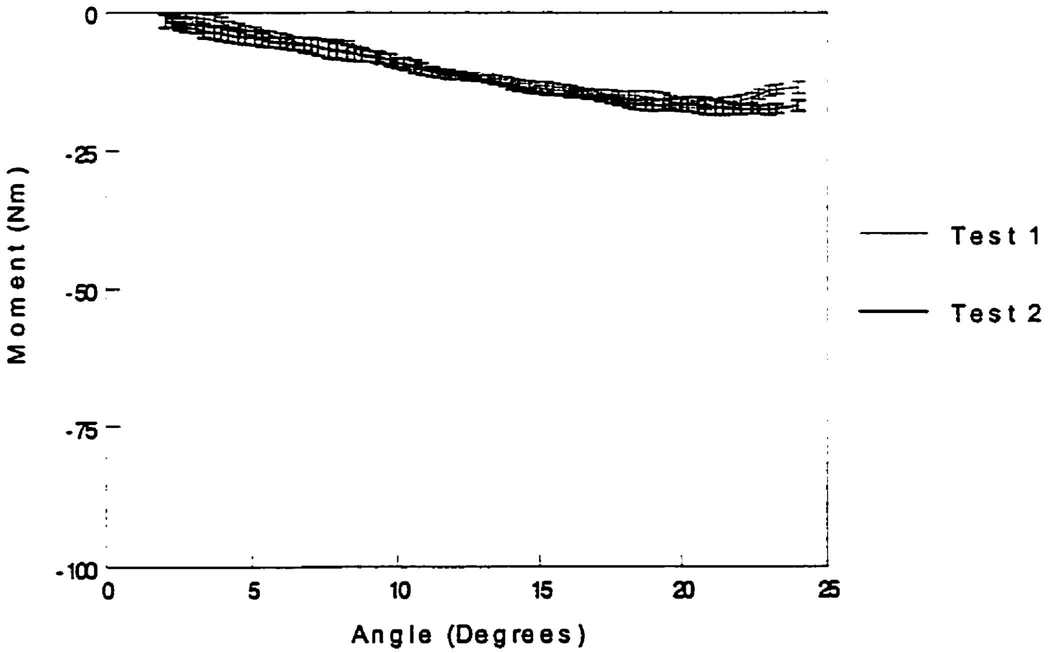
MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
Q7 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q8 – RAW DATA

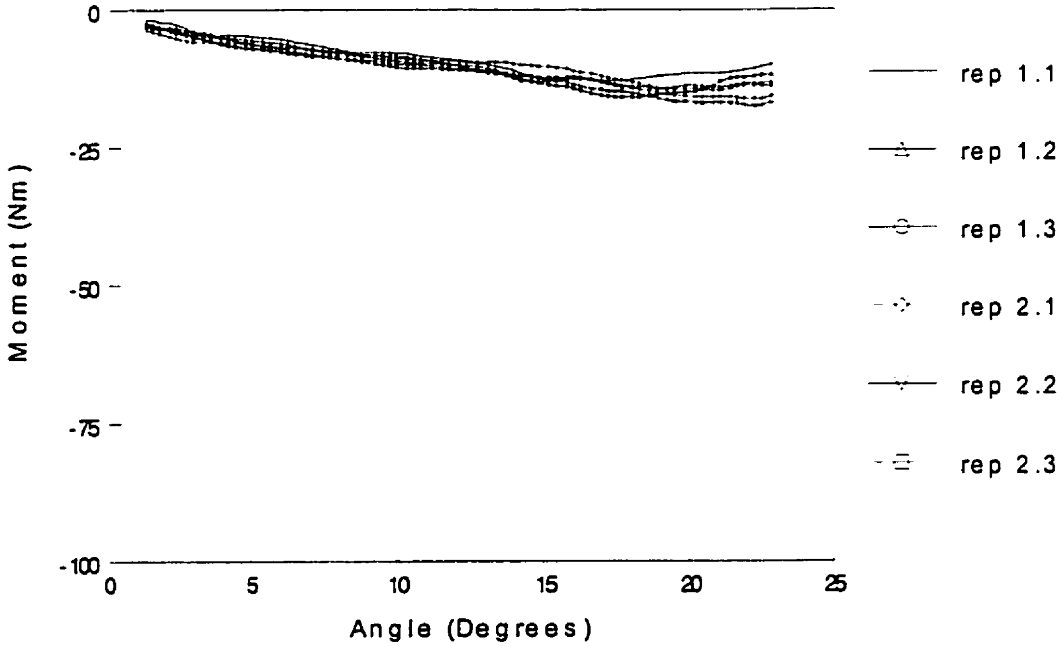


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
Q8 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q9 – RAW DATA



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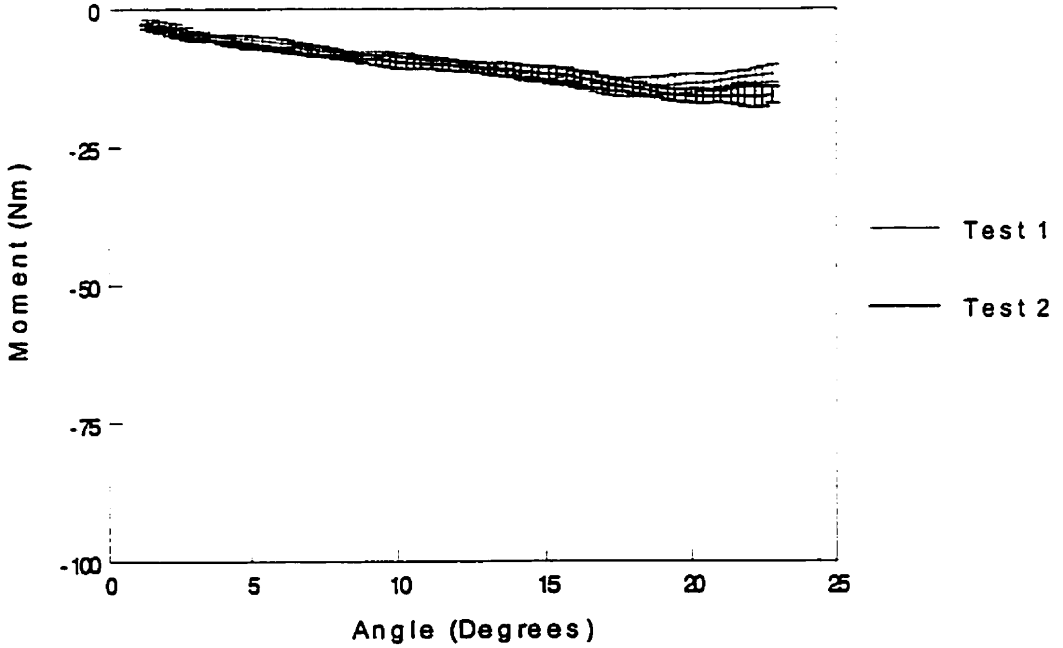
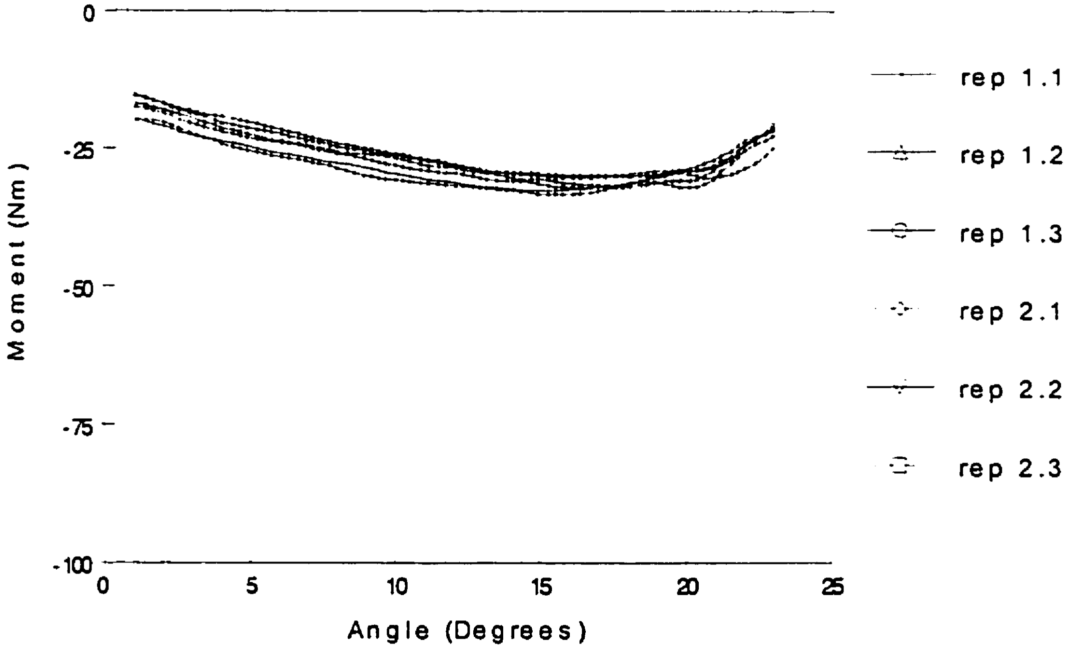
Q9 – STANDARD DEVIATION

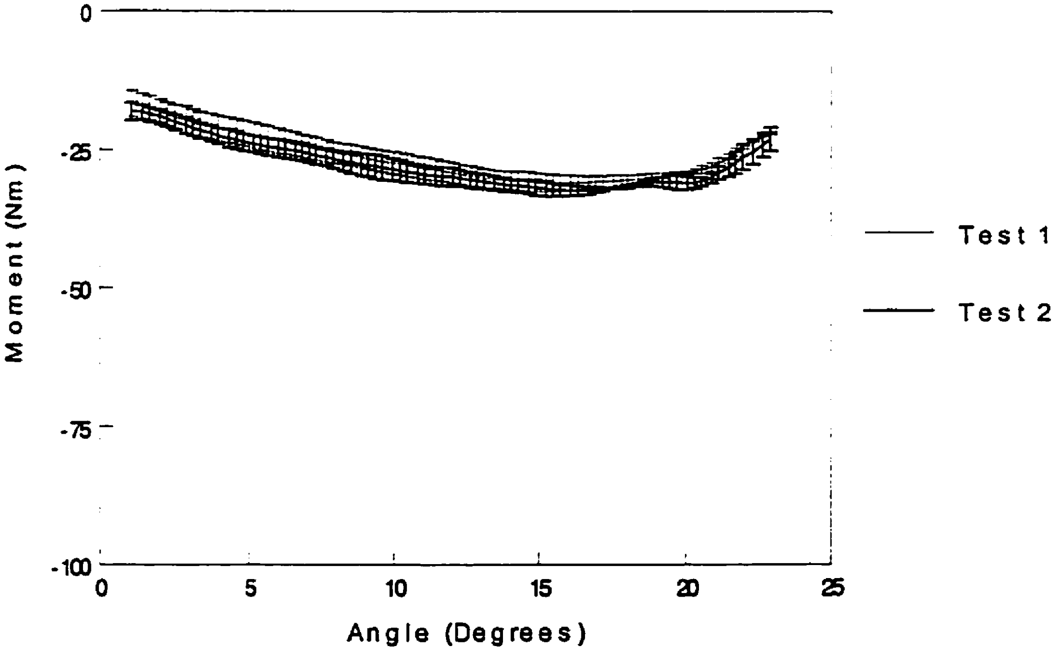


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
QUADRIPLEGIC PARTICIPANT Q10 – RAW DATA

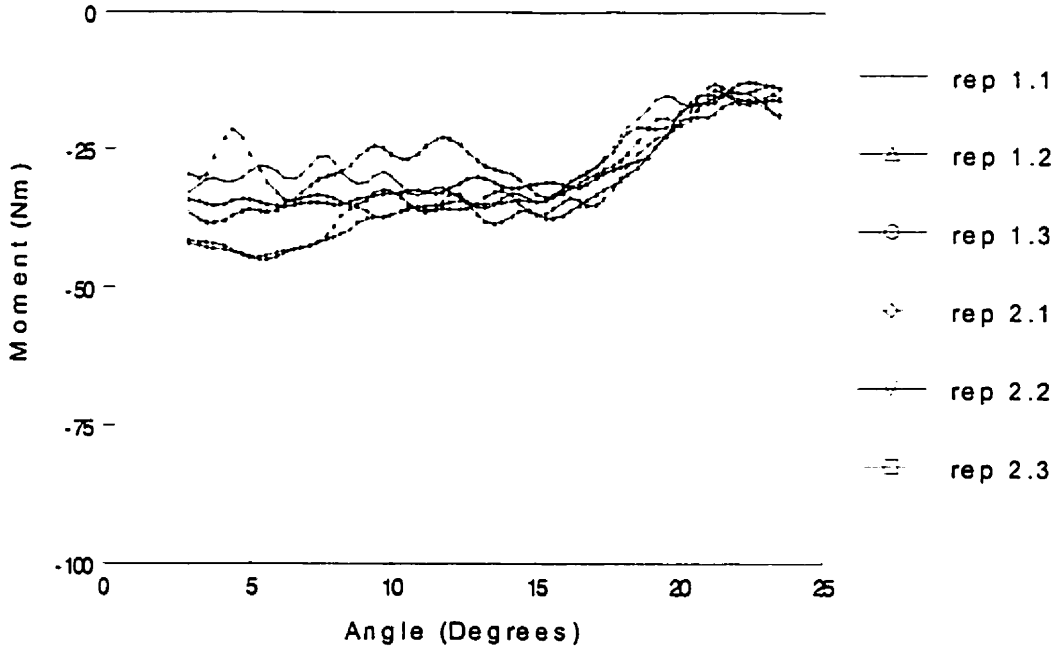


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
Q10 – STANDARD DEVIATION

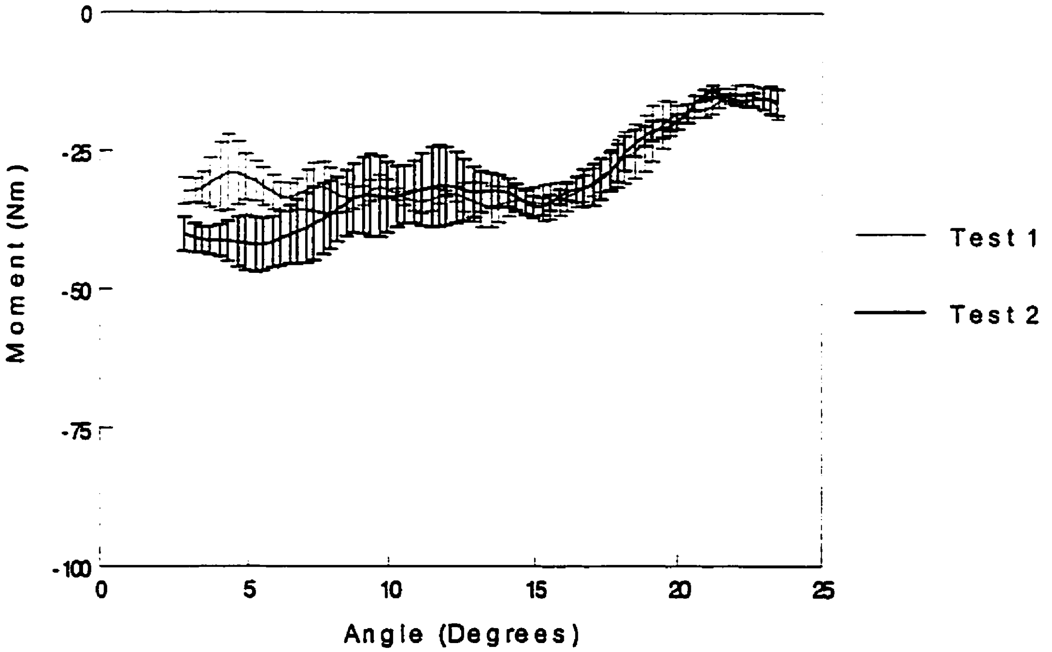


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P1 – RAW DATA

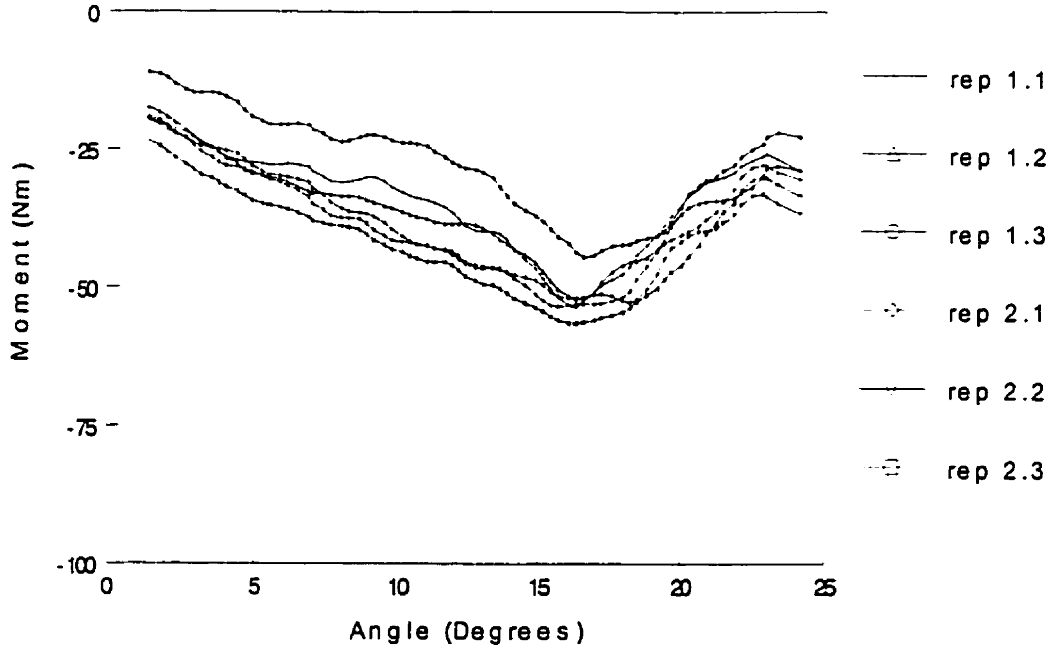


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
P1 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P2 – RAW DATA



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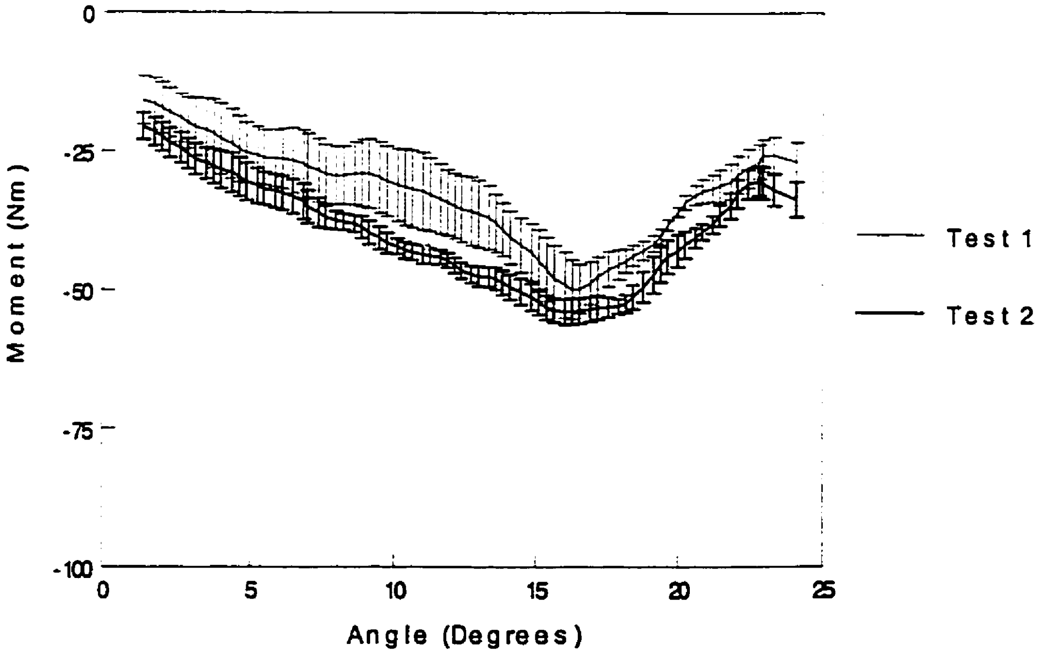
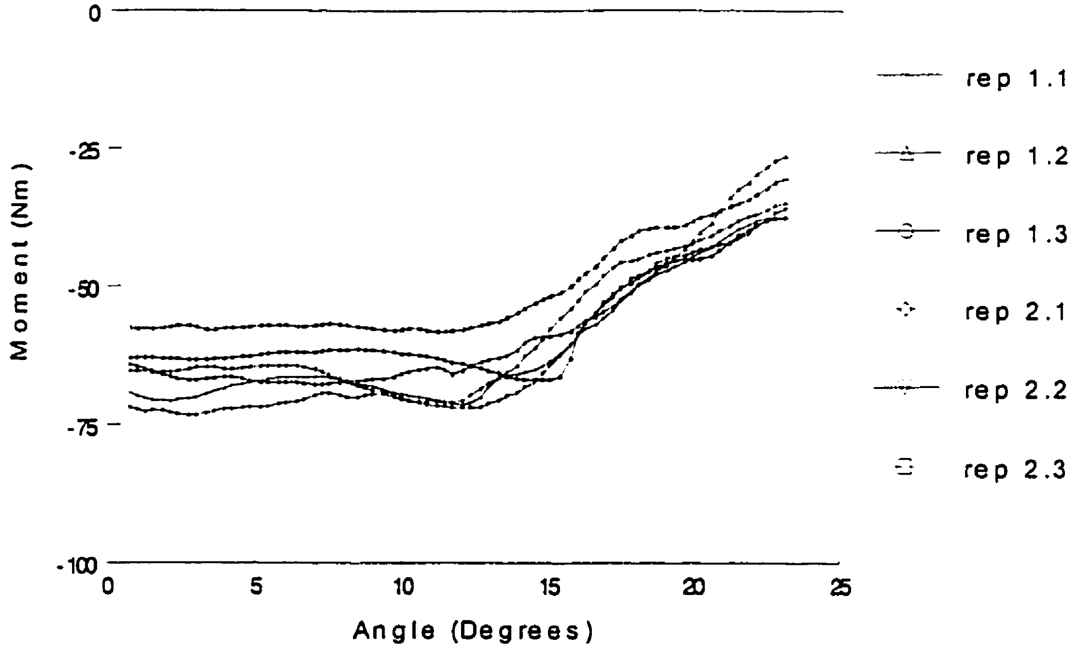
P2 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants



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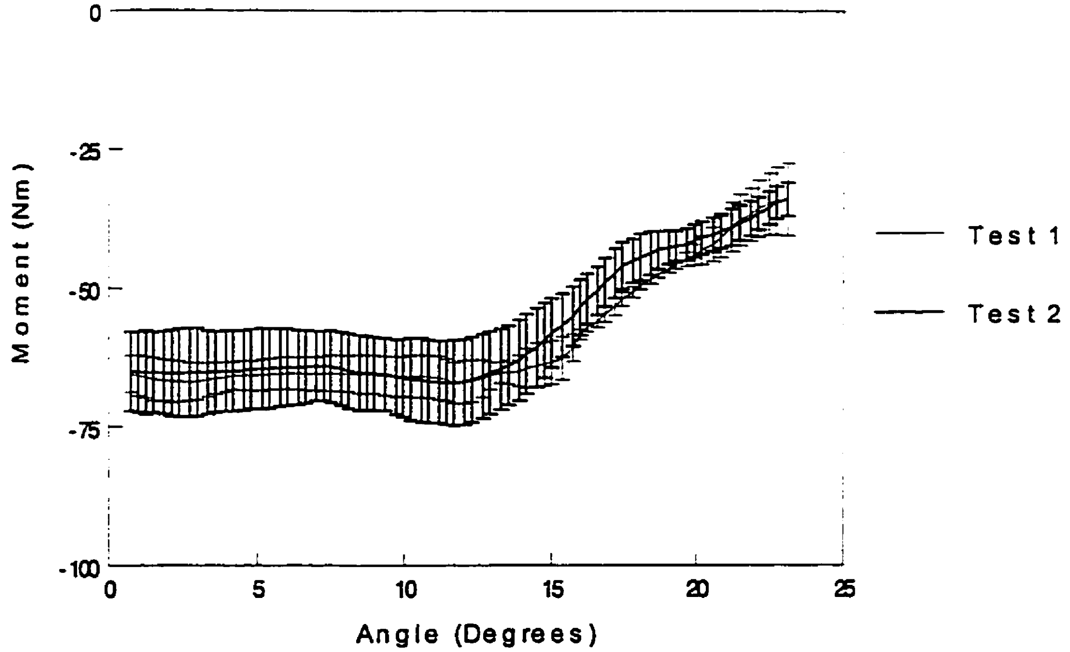
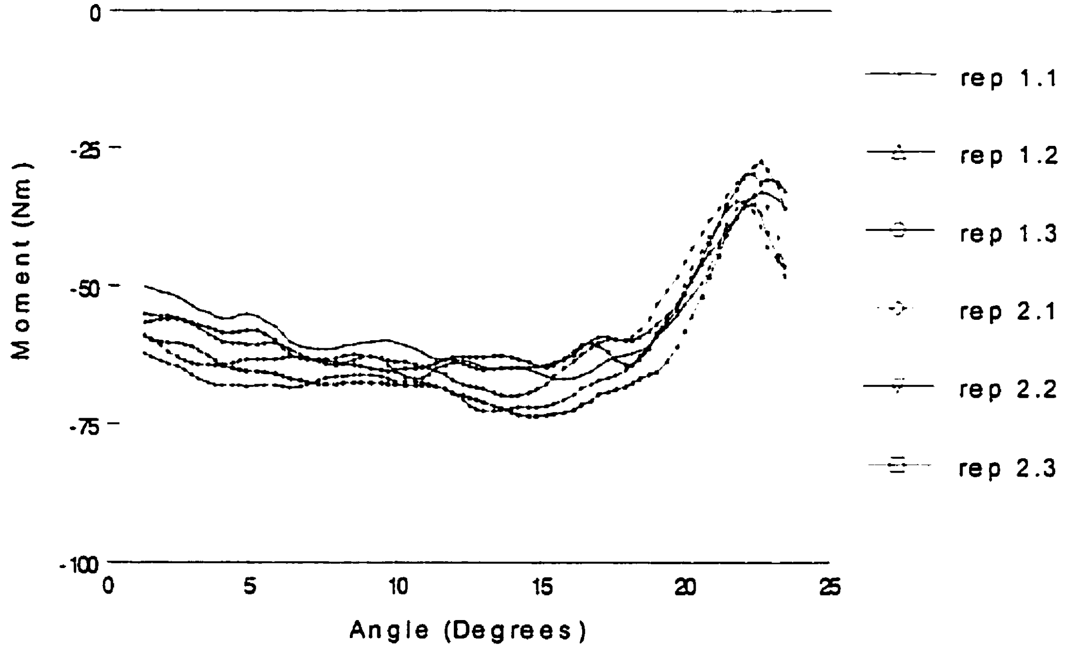
PARAPLEGIC PARTICIPANT P3 – RAW DATA



MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
P3 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P4 – RAW DATA



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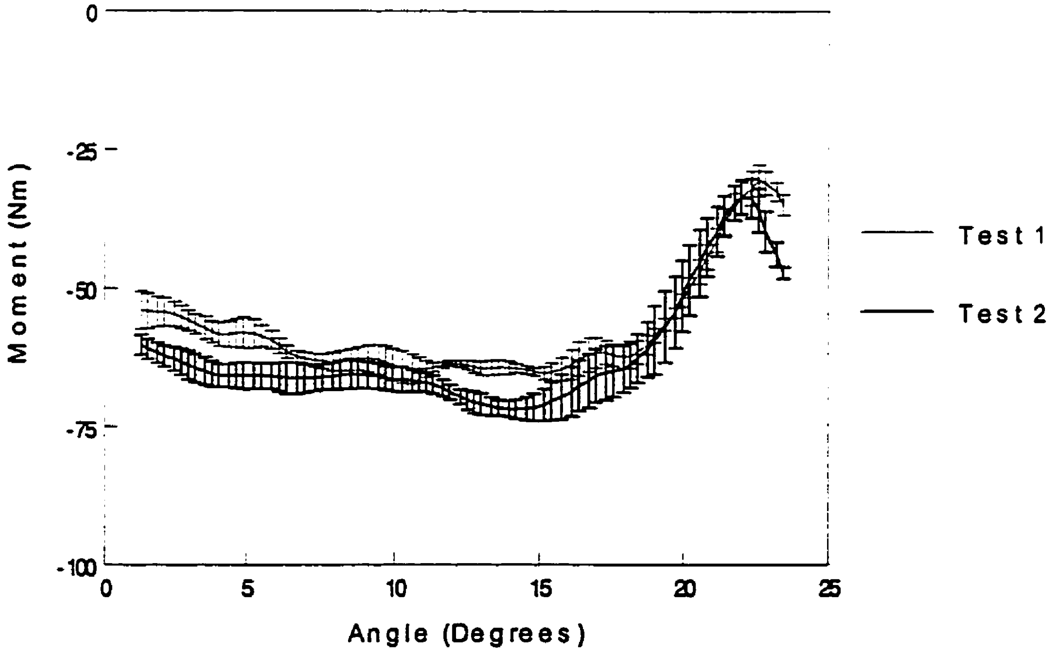
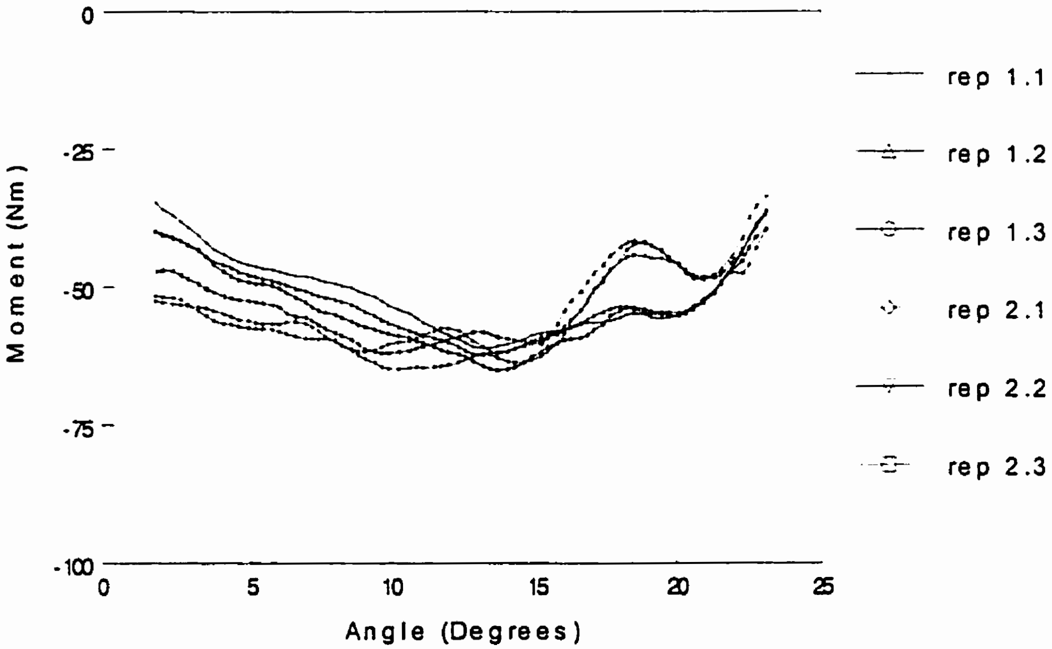
P4 – STANDARD DEVIATION

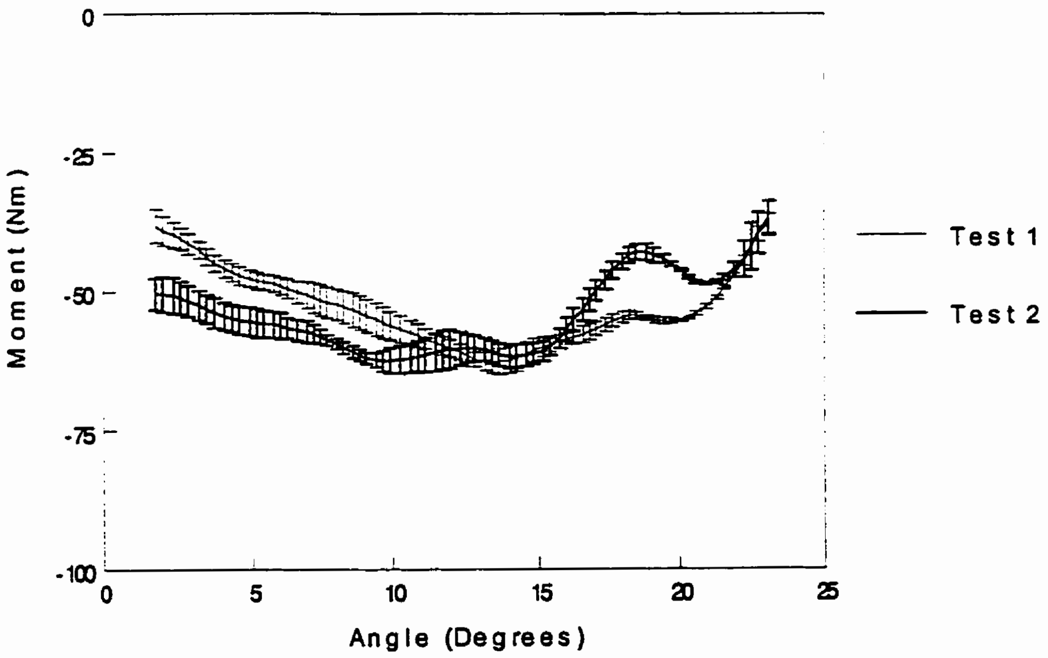


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P5 – RAW DATA

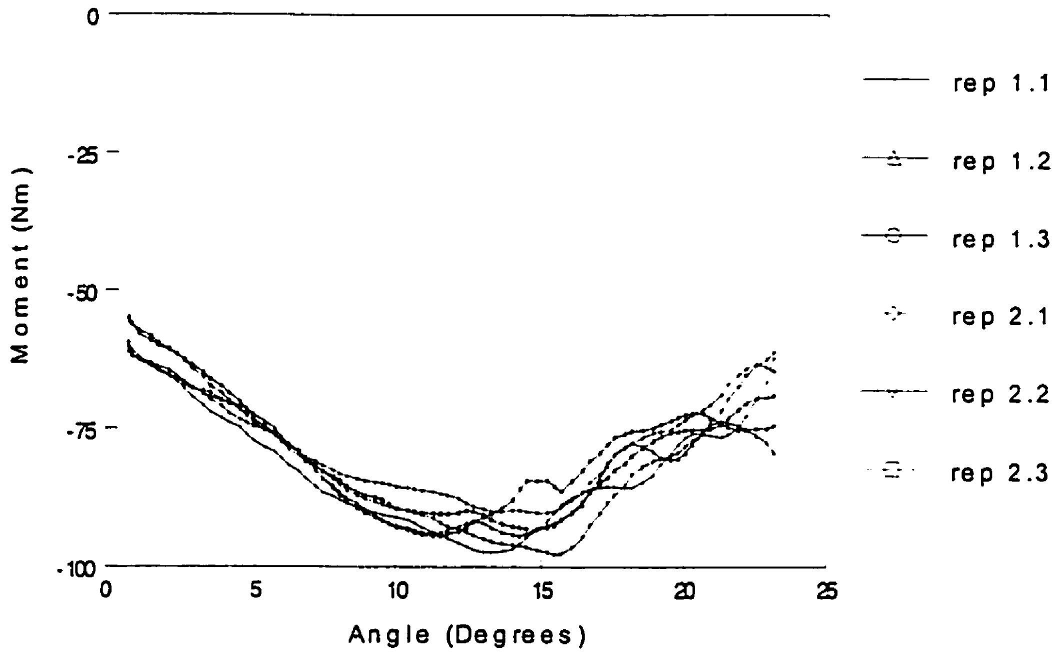


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
P5 – STANDARD DEVIATION

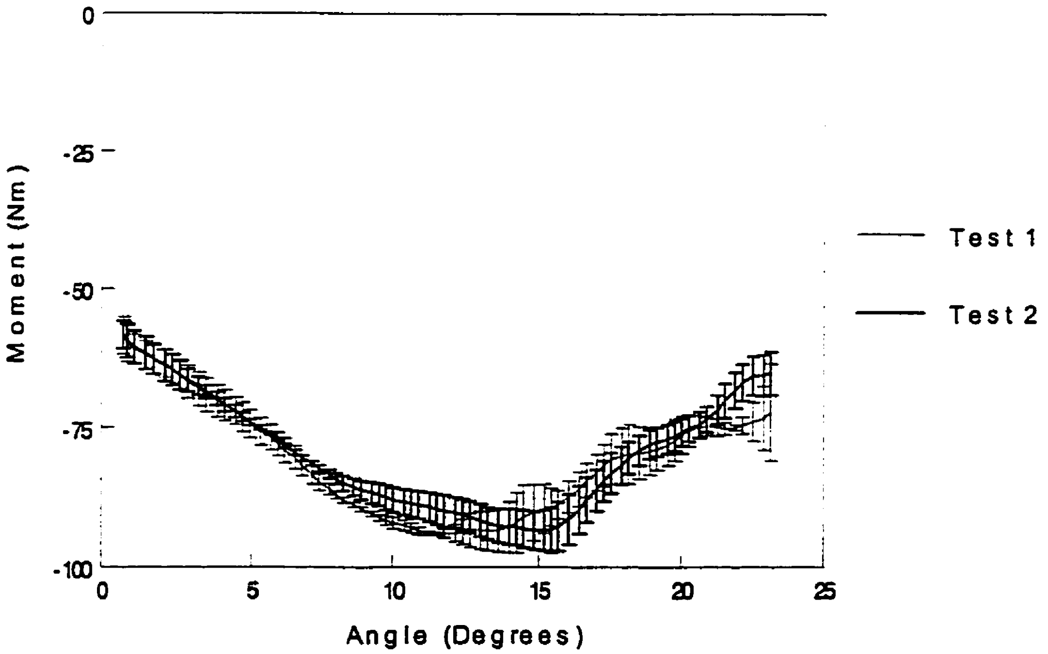


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P6 – RAW DATA

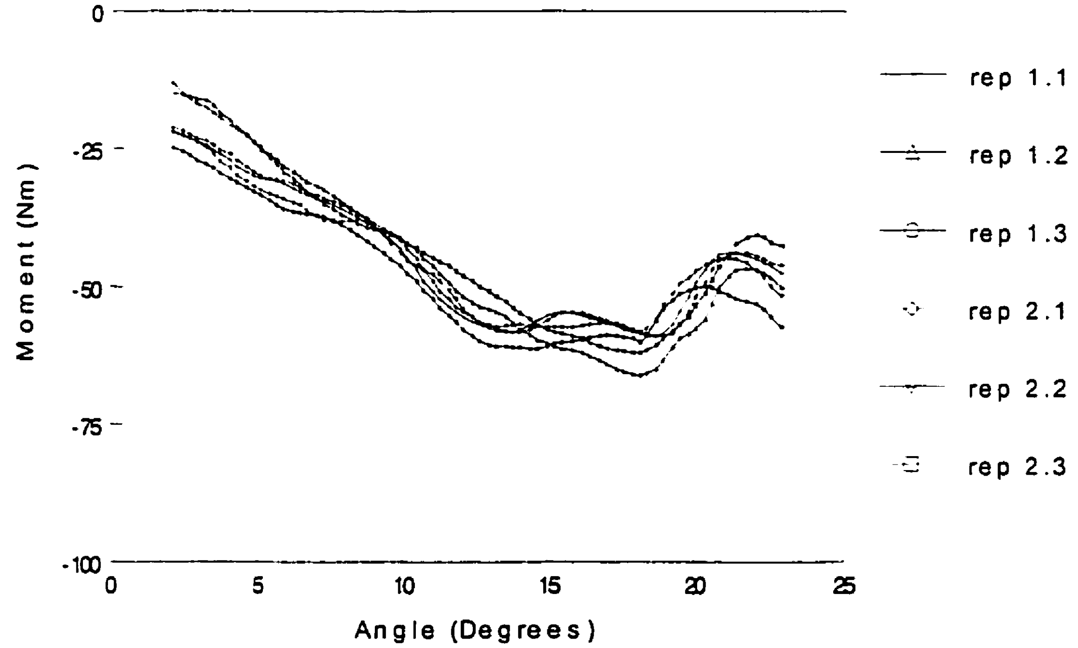


MEAN FORCE GENERATED DURING SCAPULAR DEPRESSION
P6 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P7 – RAW DATA



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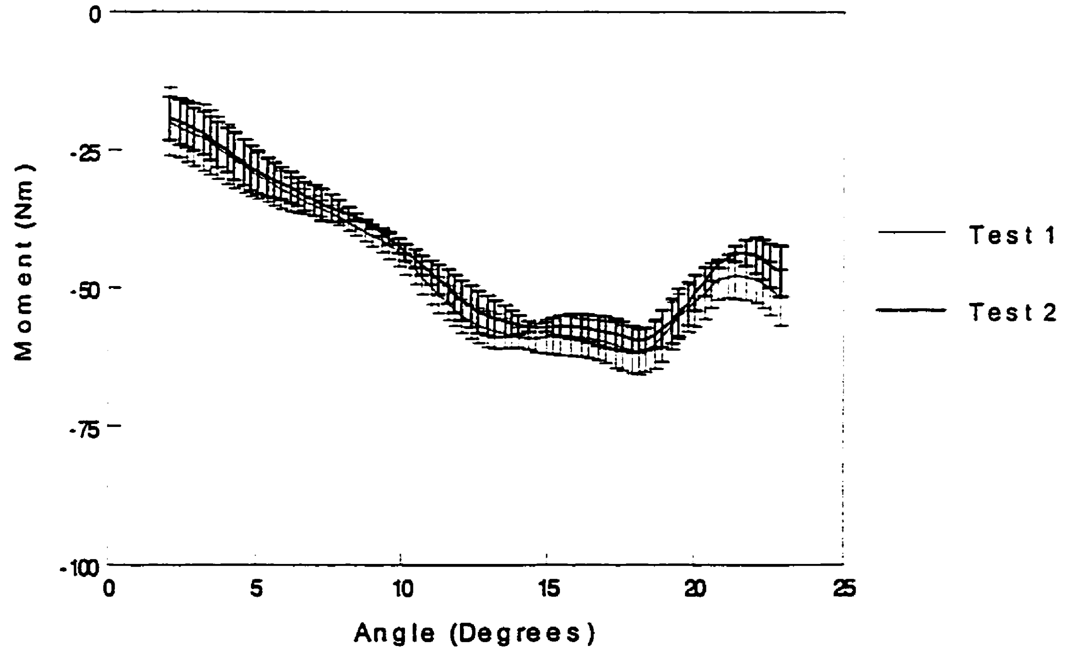
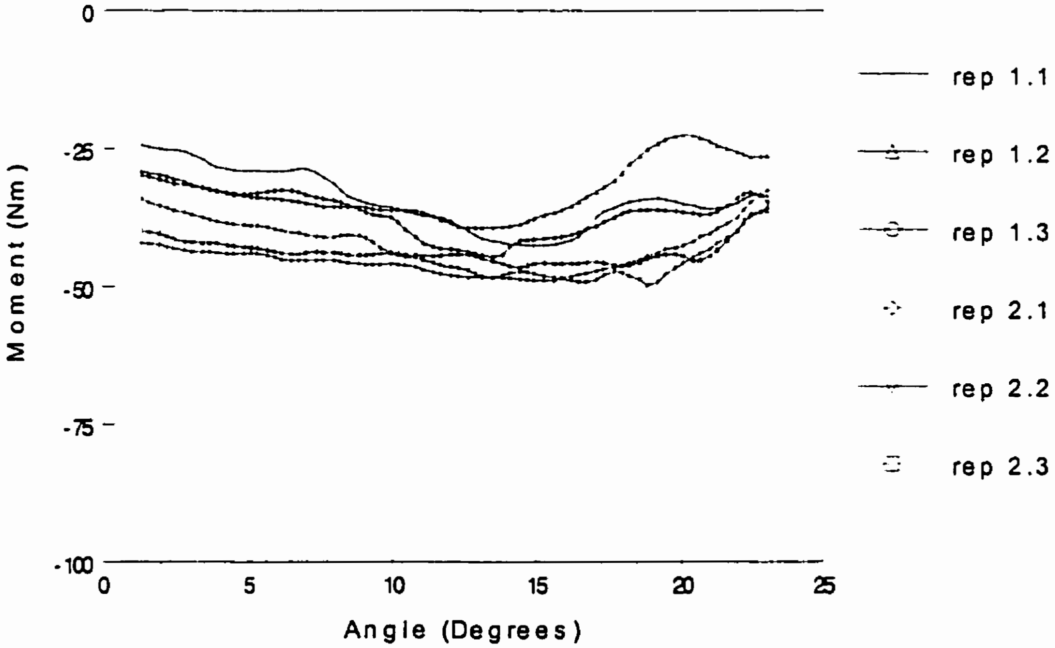
P7 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P8 – RAW DATA



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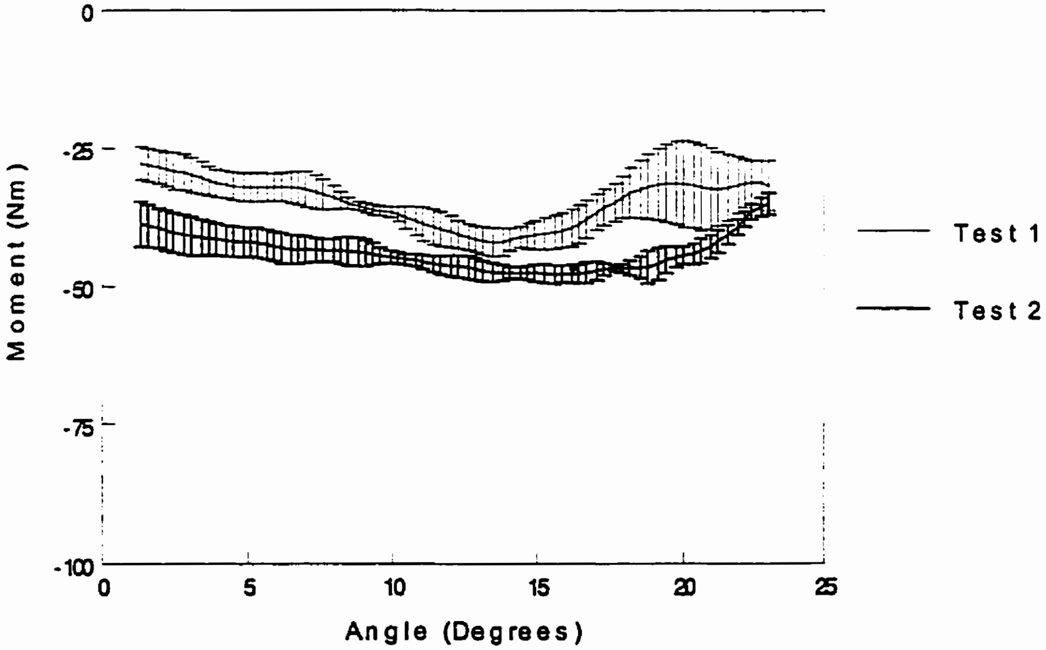
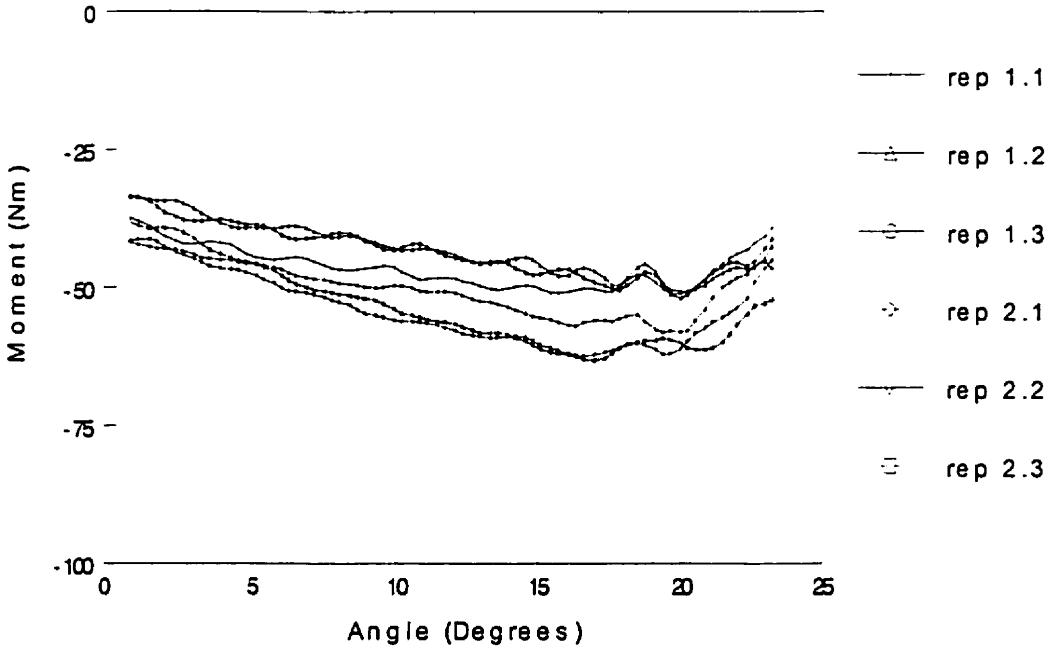
P8 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING SCAPULAR DEPRESSION
PARAPLEGIC PARTICIPANT P9 – RAW DATA



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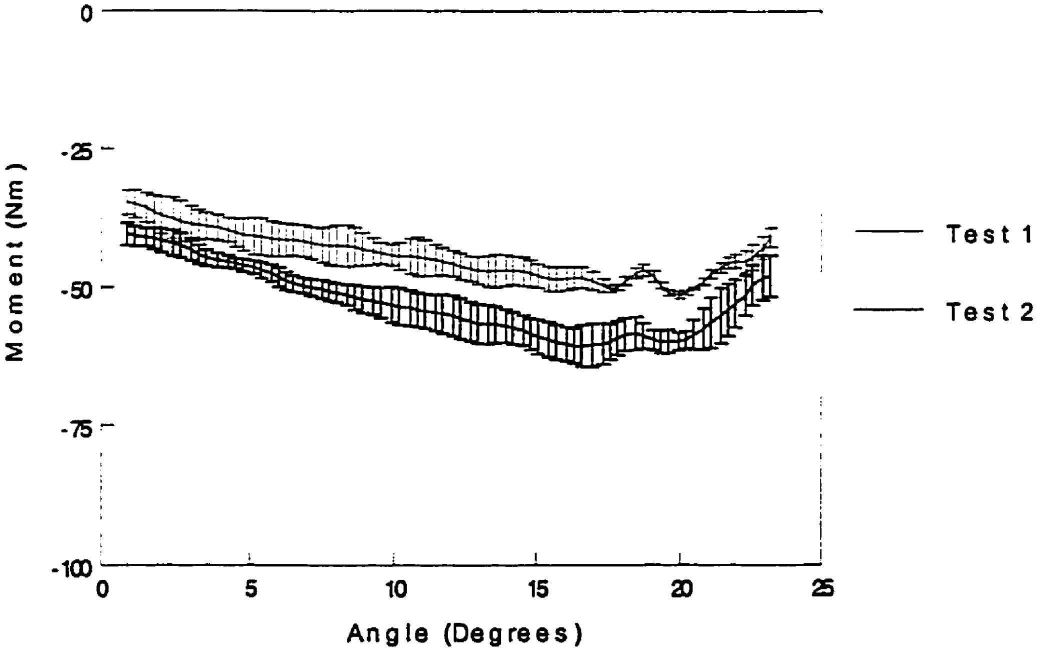
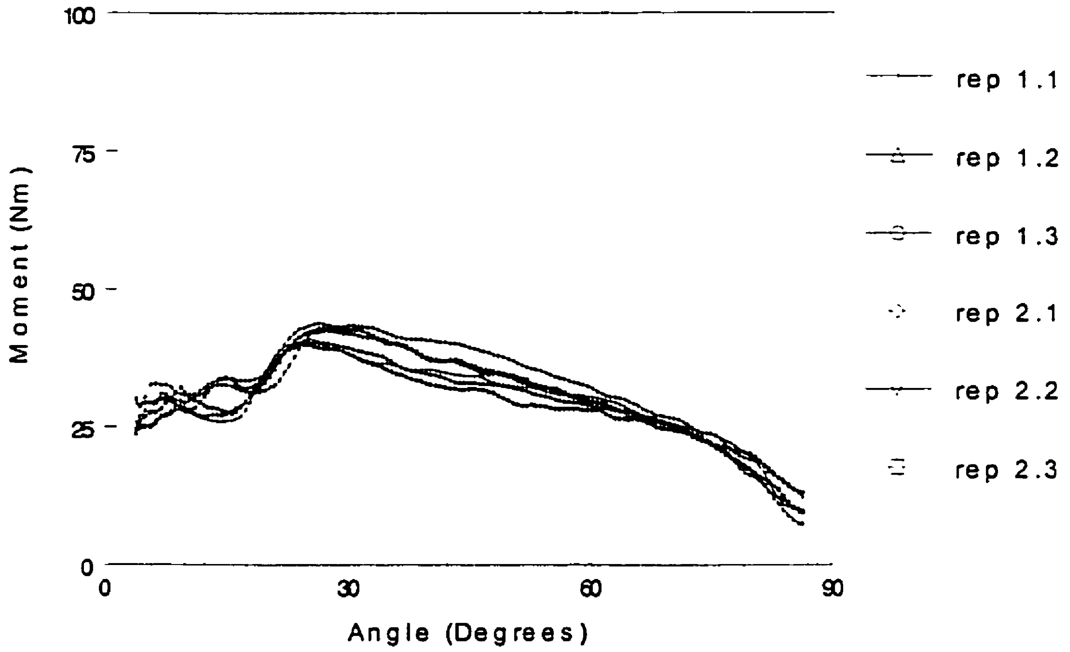
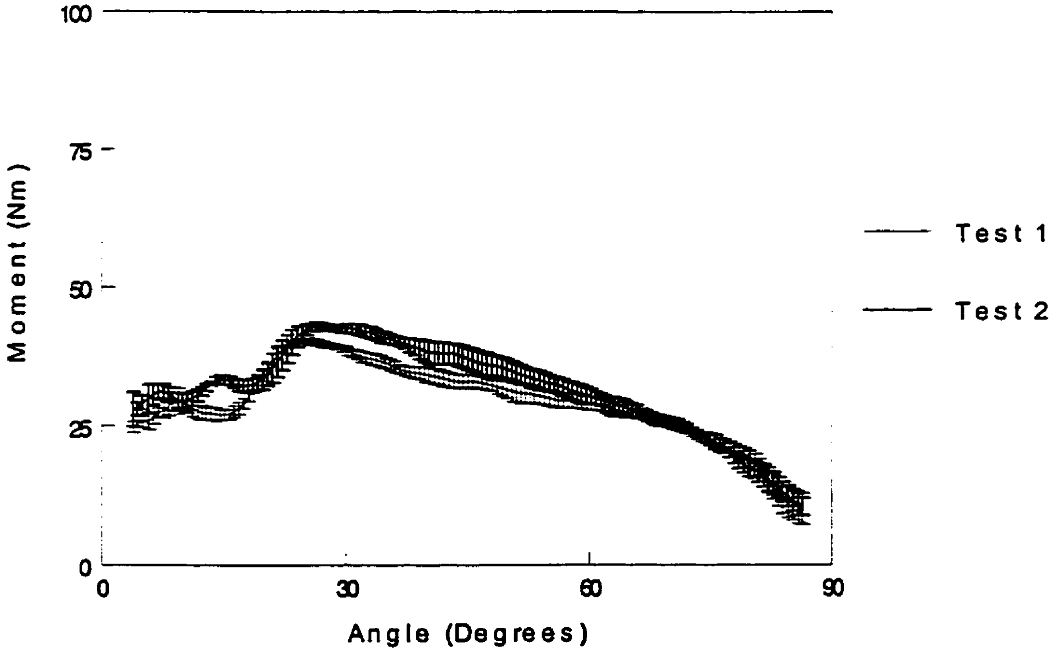
P9 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants



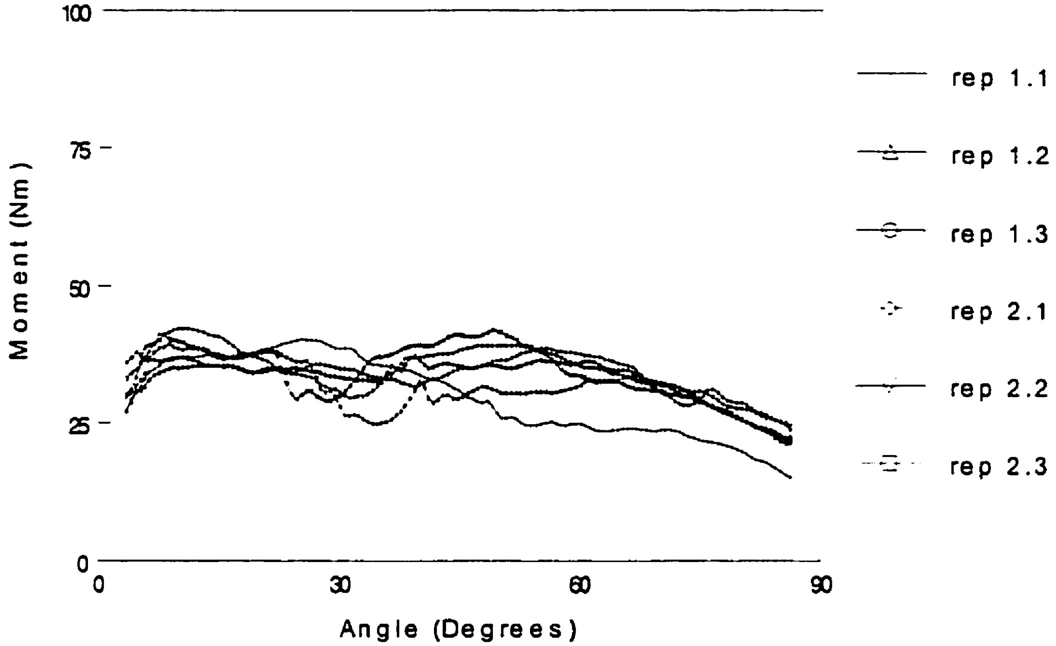
FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q1 – RAW DATA



MEAN FORCE GENERATED DURING GLENOHUMERAL FLEXION
Q1 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q2 – RAW DATA



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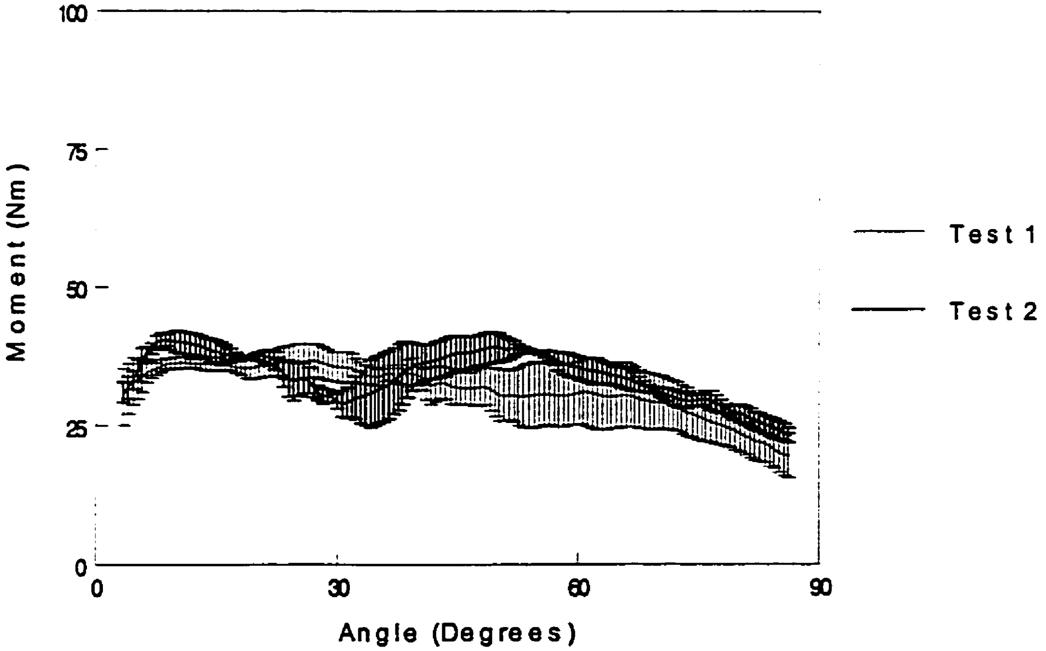
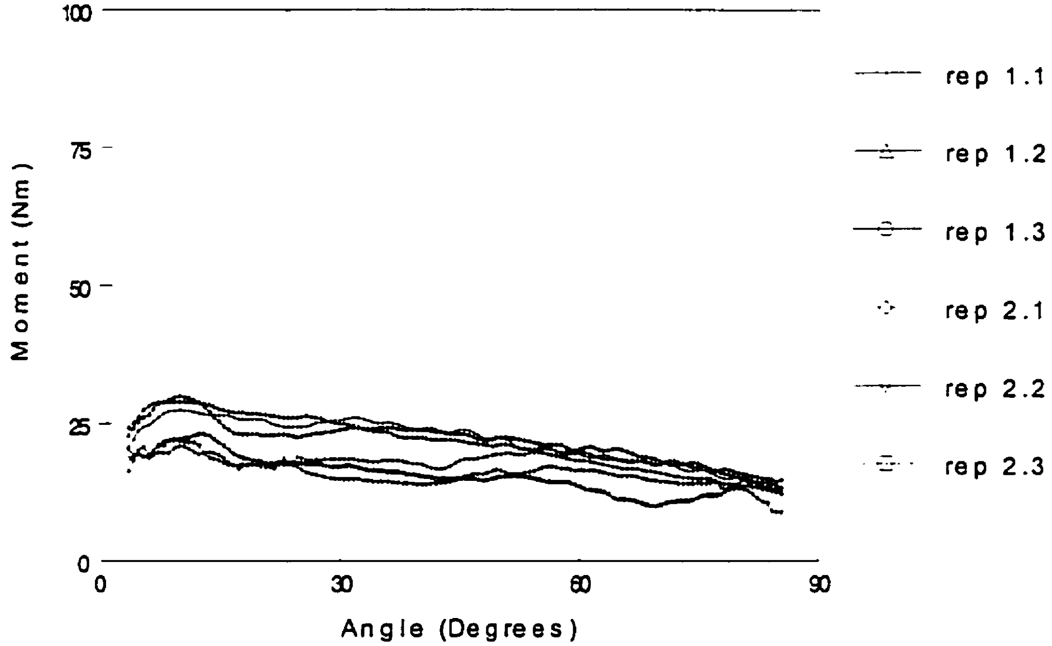
Q2 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q3 – RAW DATA



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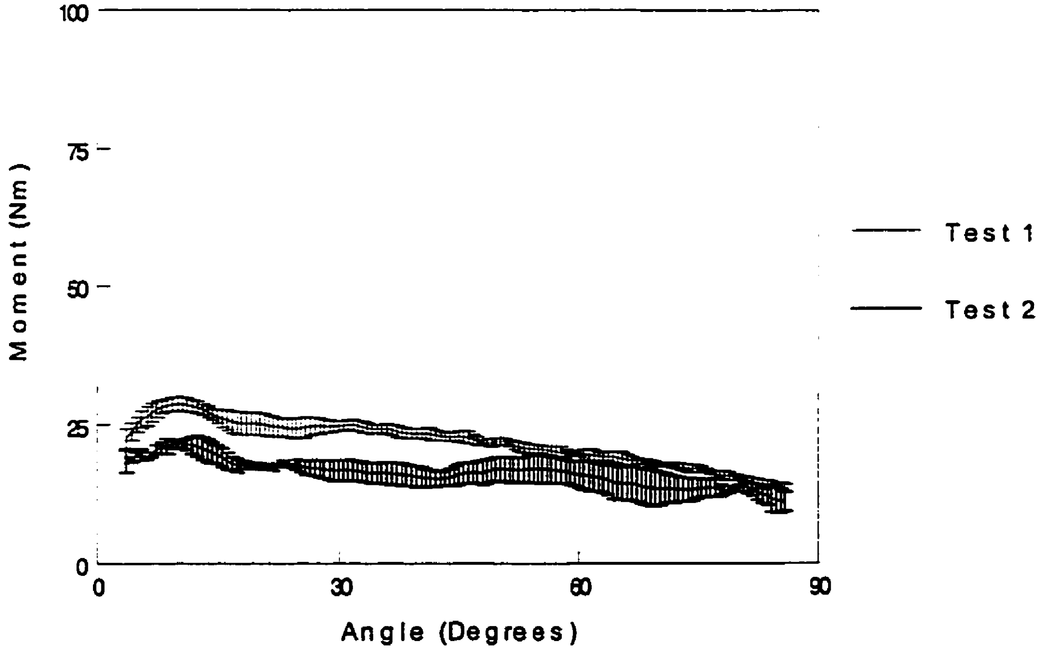
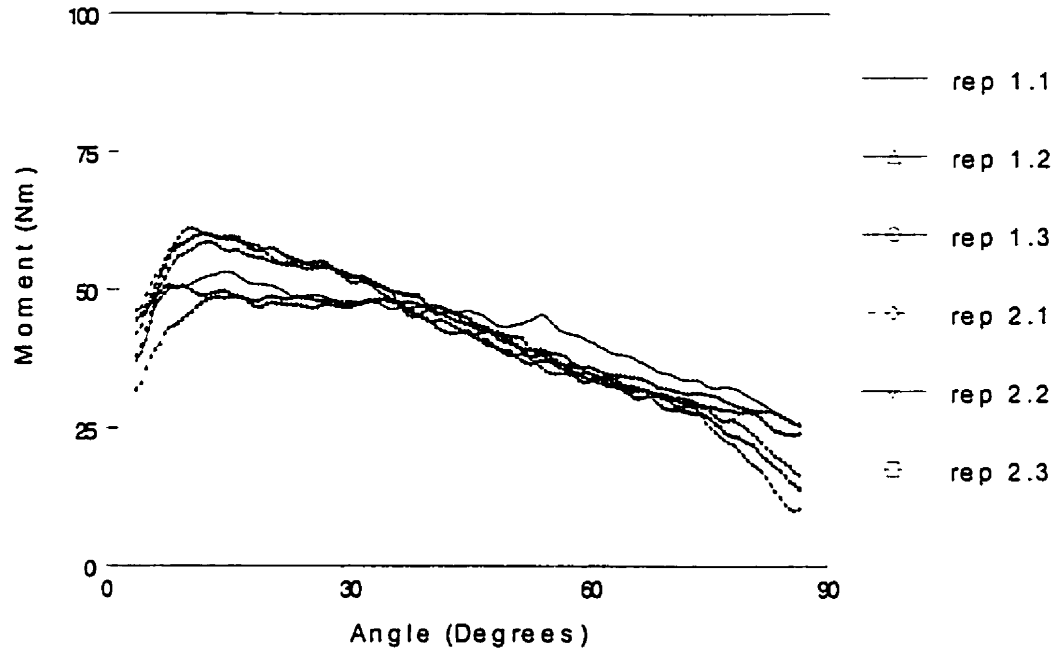
Q3 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q4 – RAW DATA



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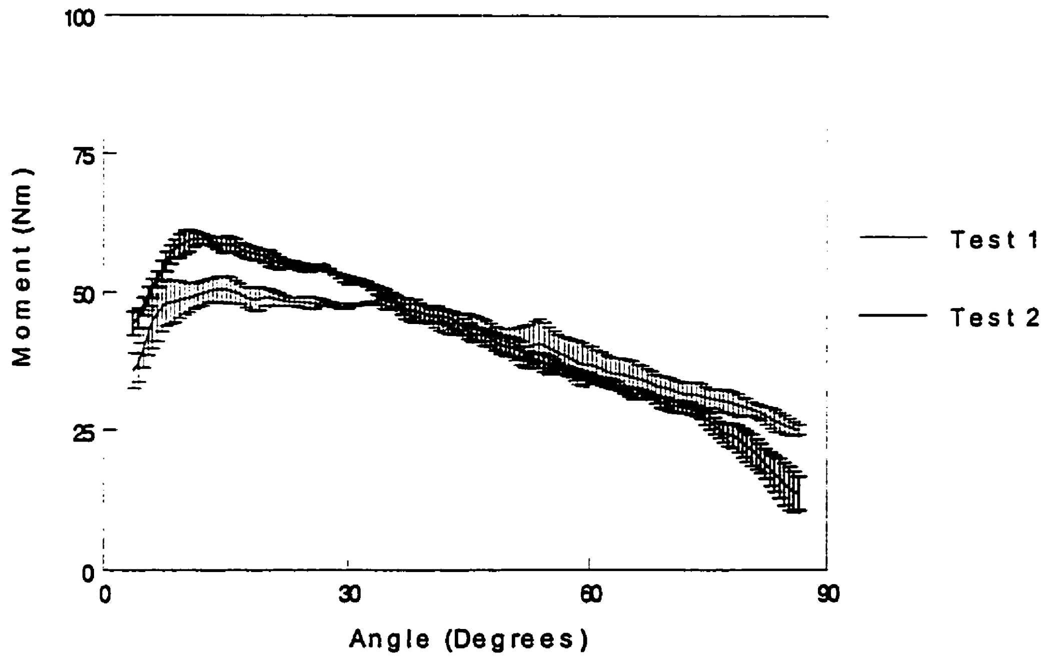
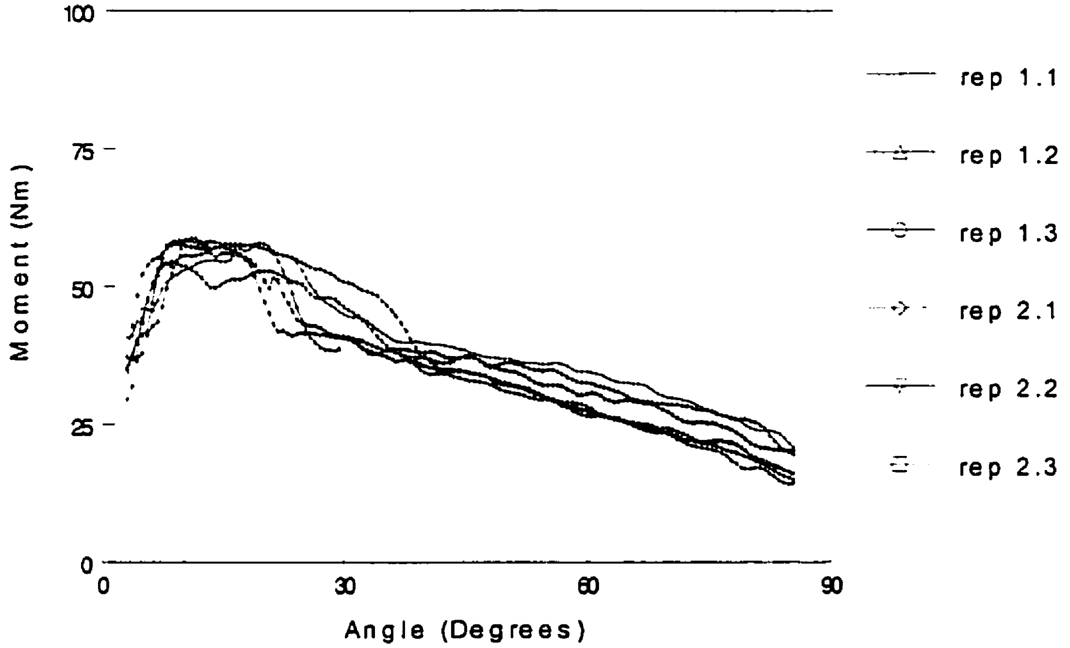
Q4 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q5 – RAW DATA



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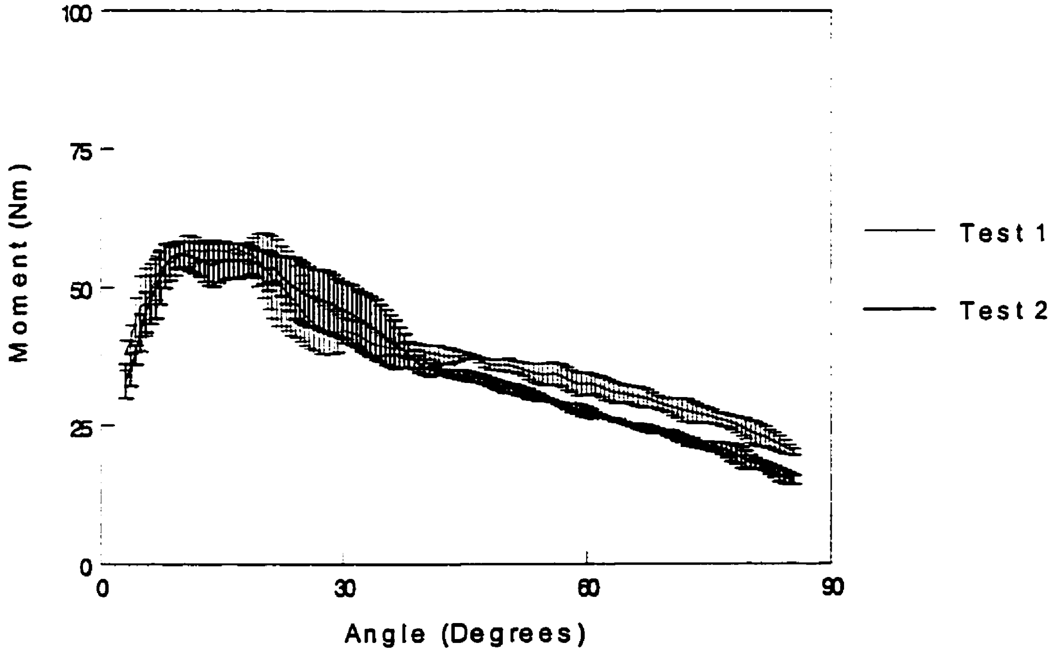
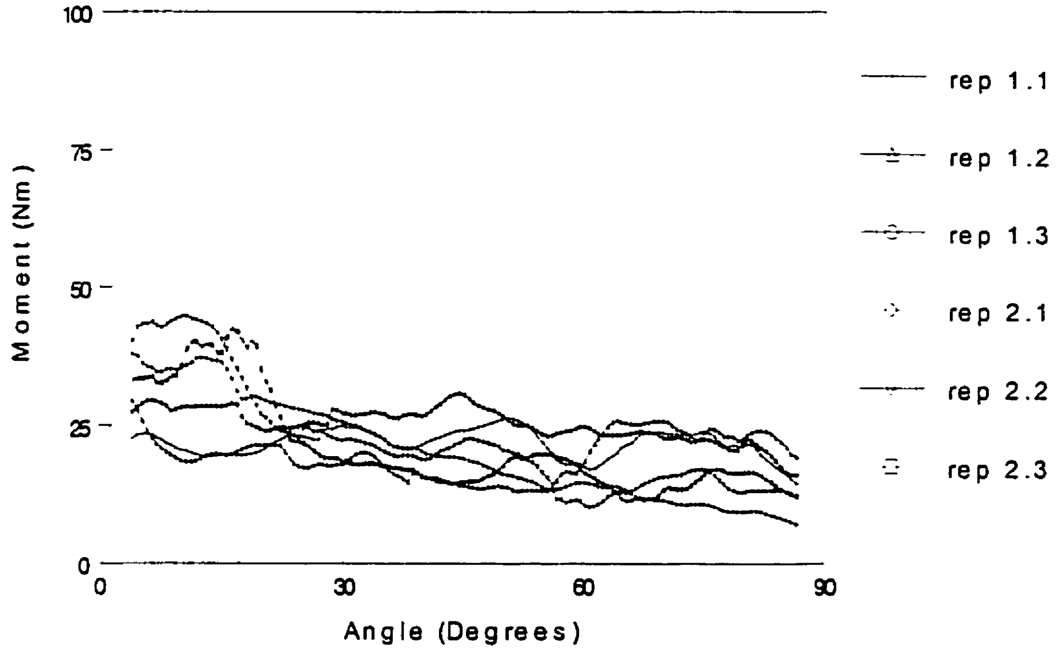
Q5 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q6 – RAW DATA



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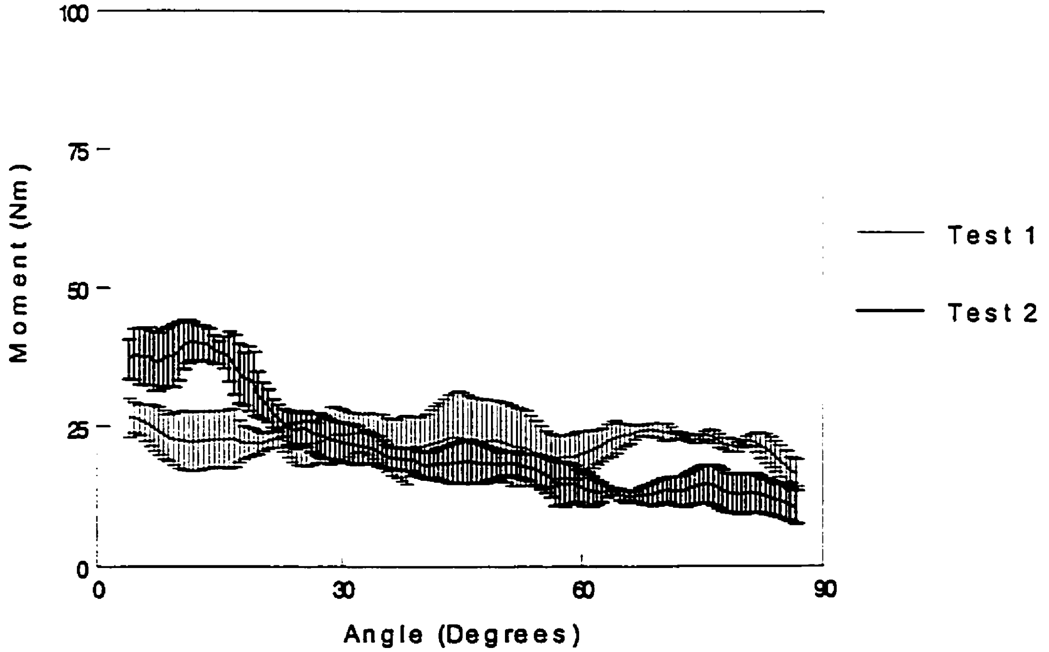
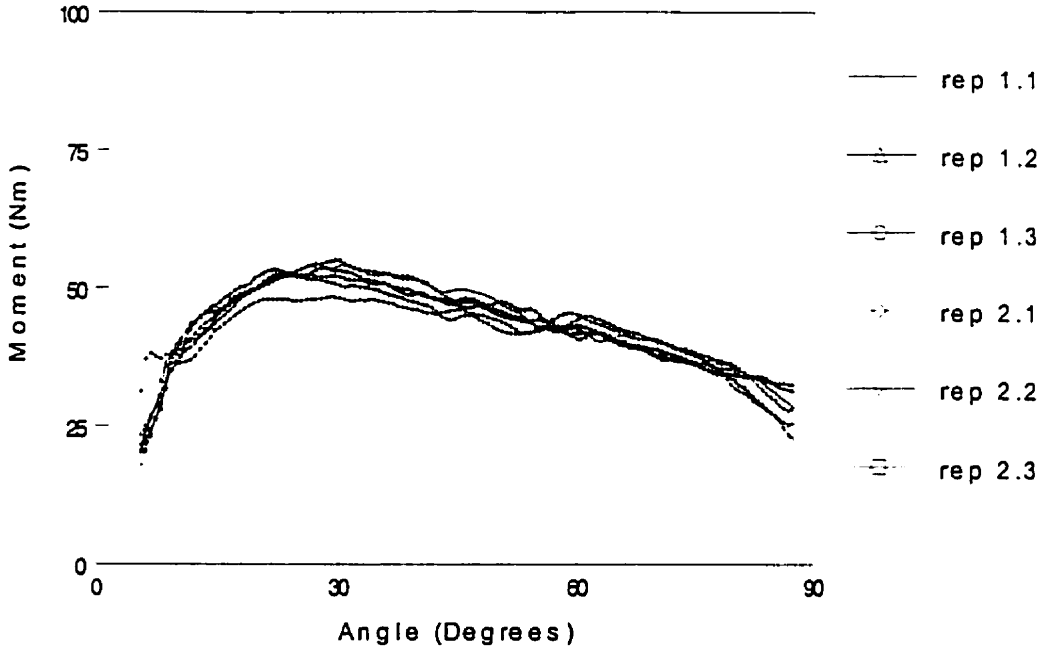
Q6 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q7 – RAW DATA



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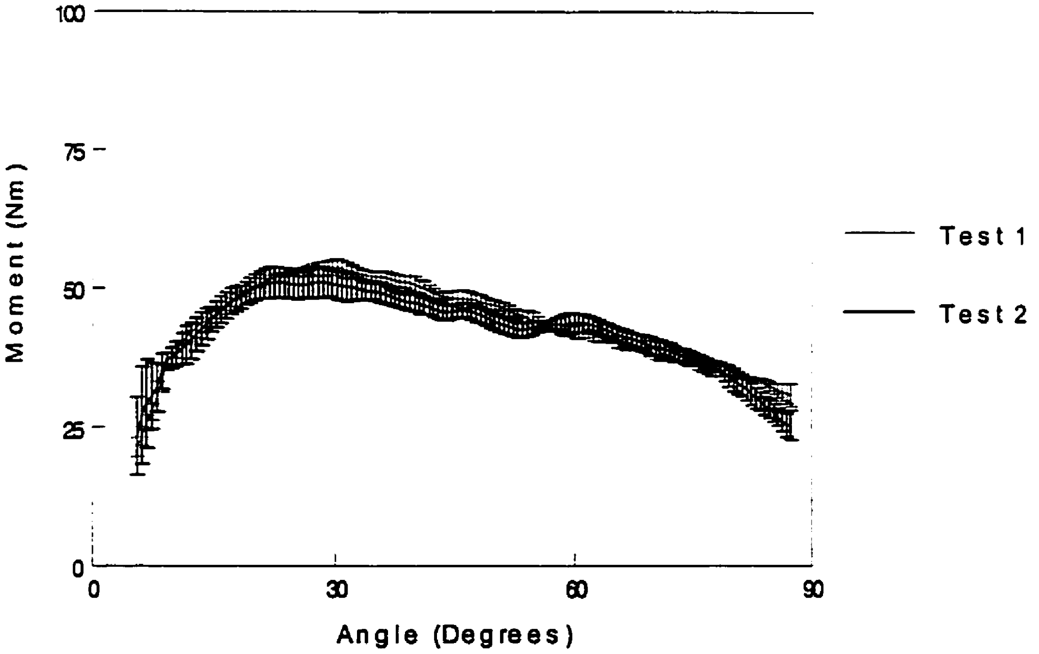
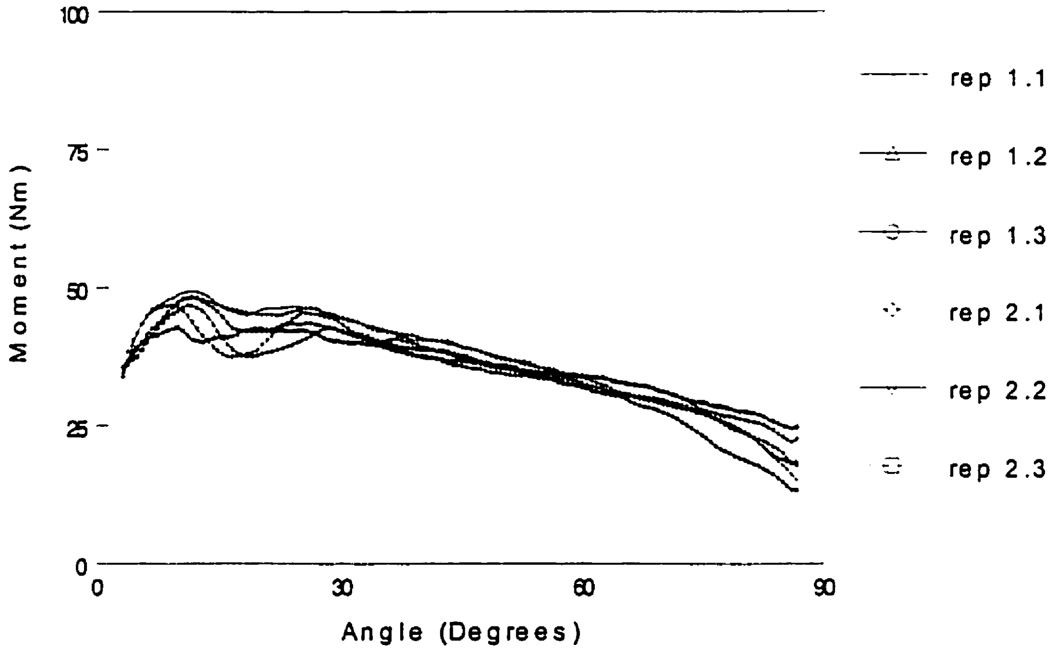
Q7 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q8 – RAW DATA



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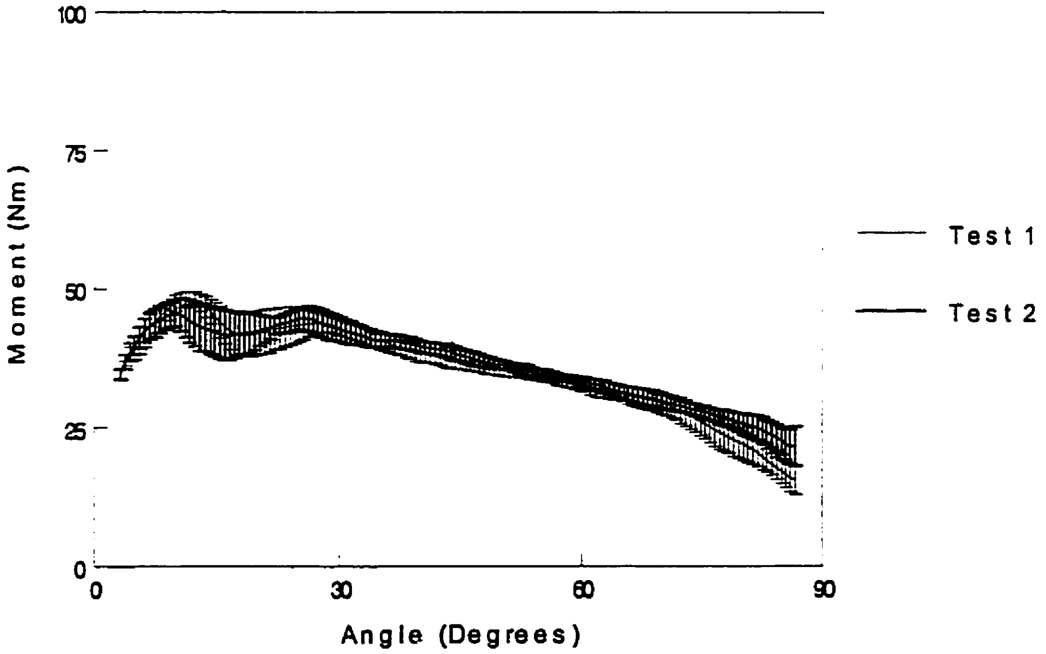
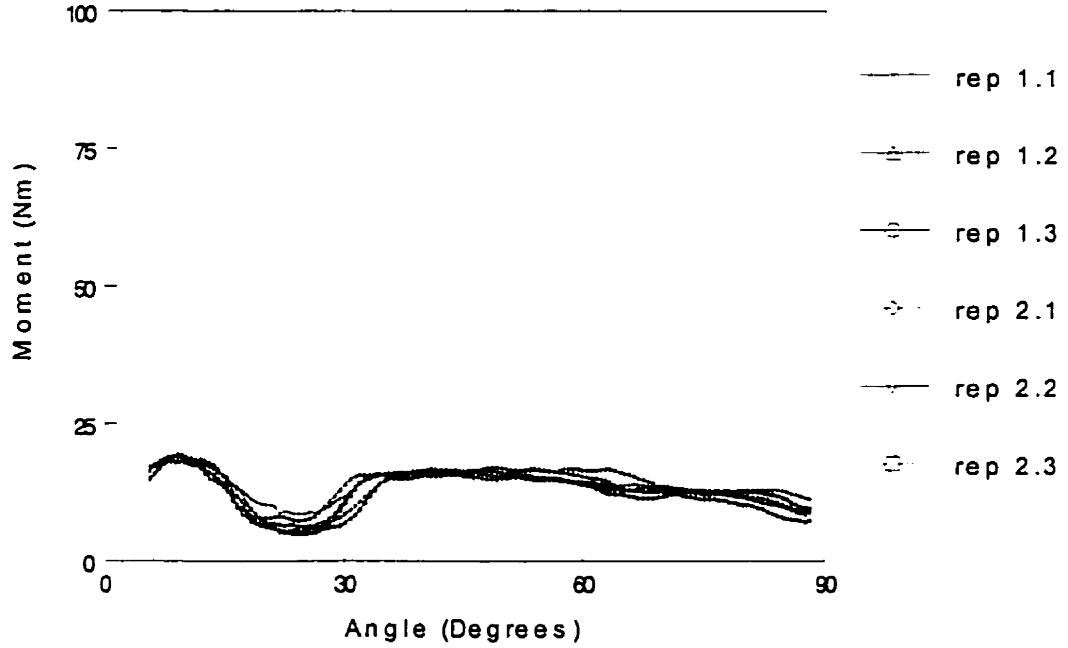
Q8 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants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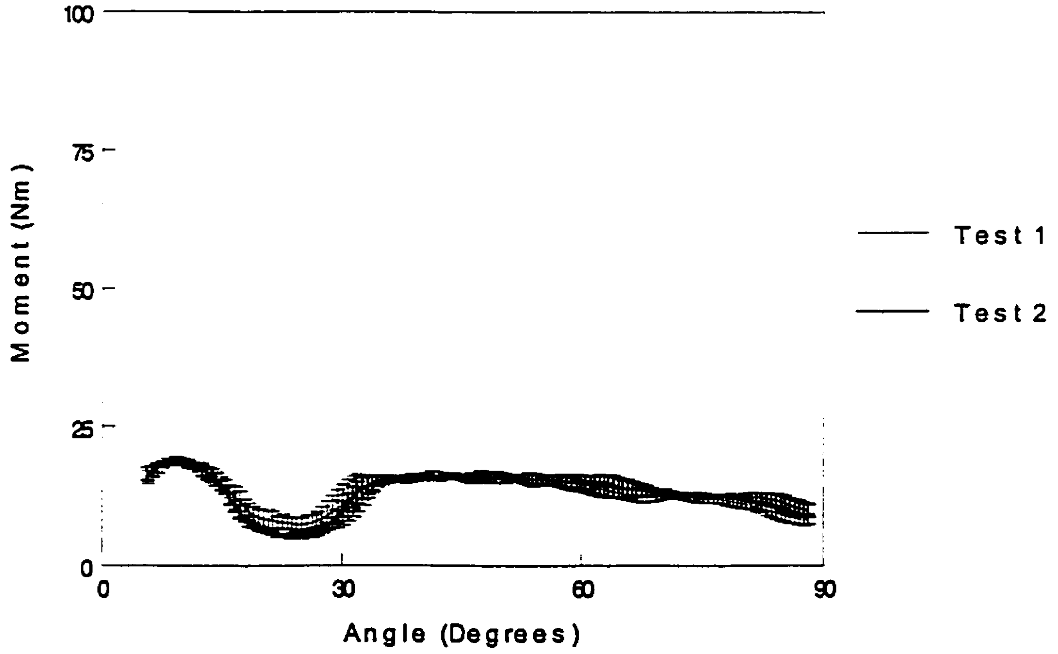
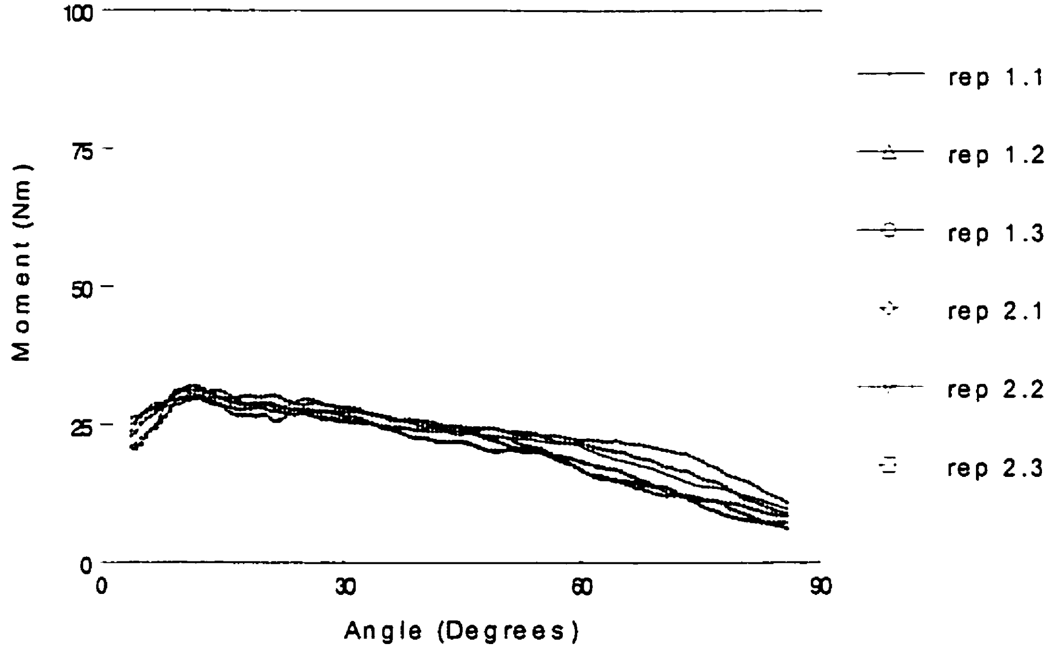
QUADRIPLEGIC PARTICIPANT Q9 – RAW DATA


MEAN FORCE GENERATED DURING GLENOHUMERAL FLEXION
Q9 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
QUADRIPLEGIC PARTICIPANT Q10 – RAW DATA



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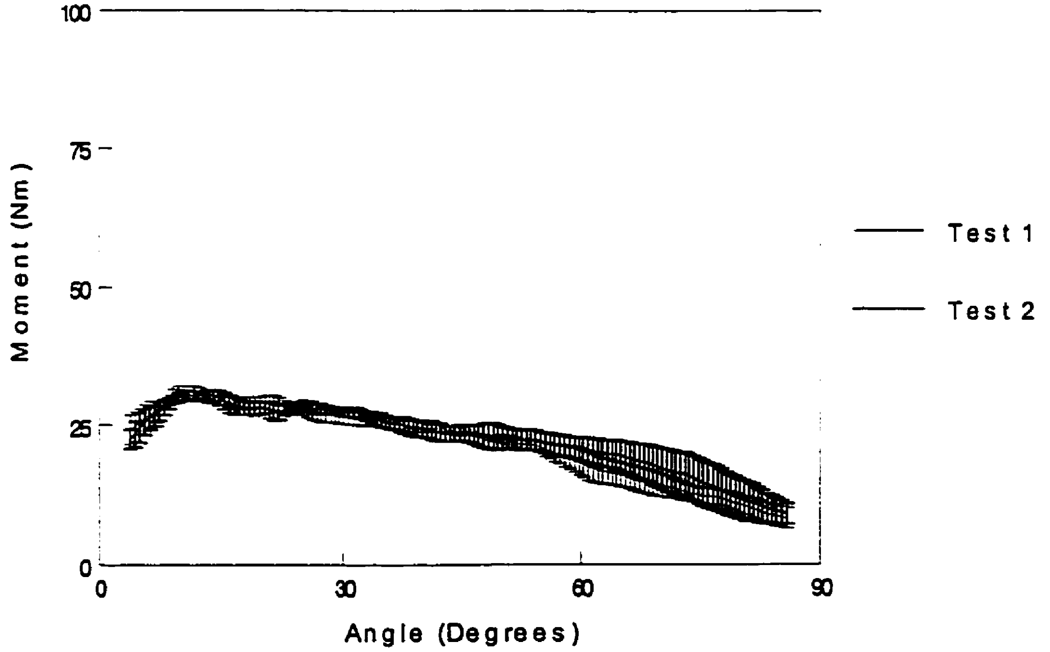
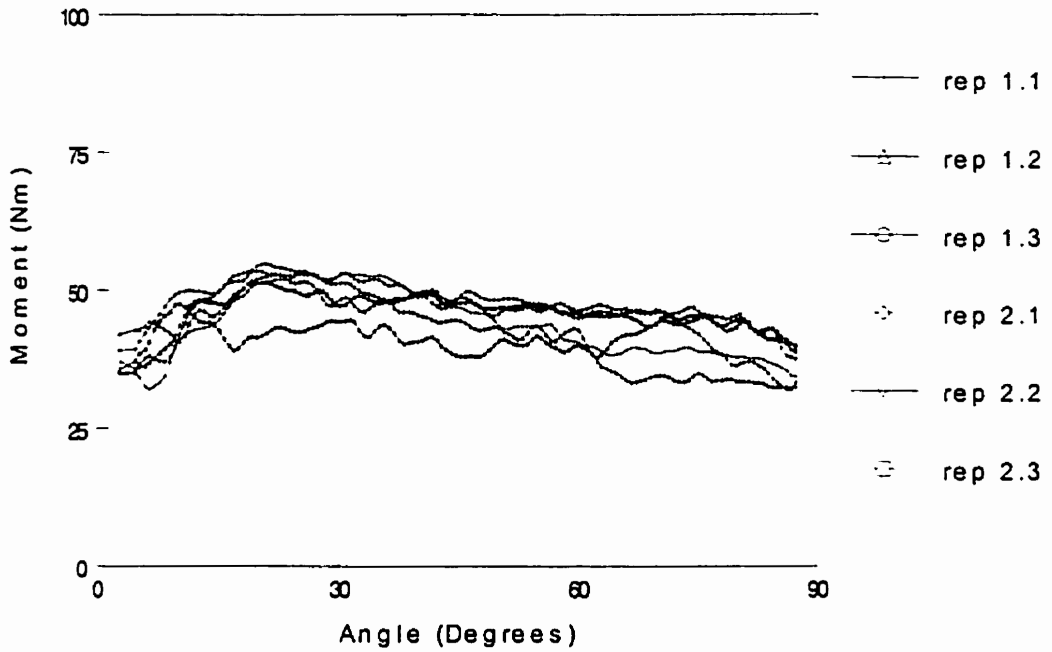
Q10 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants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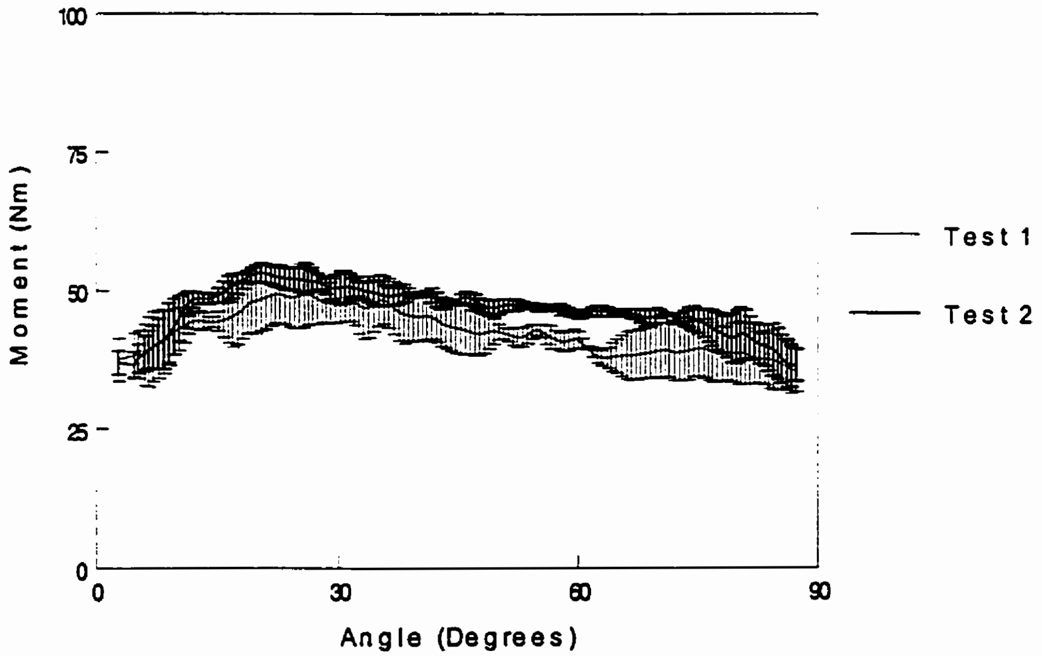
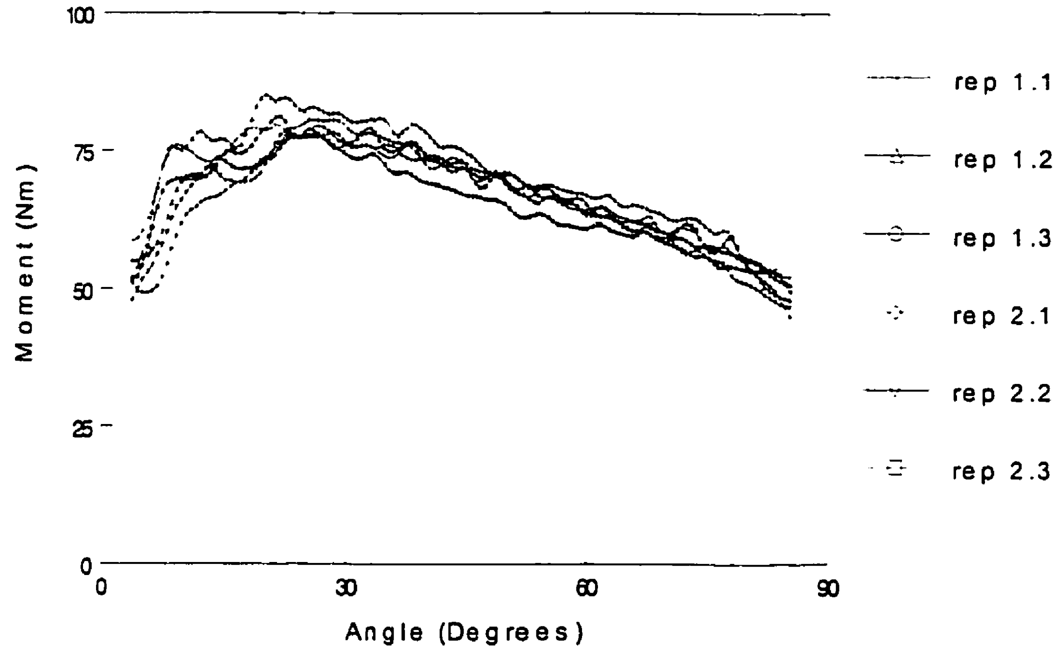
PARAPLEGIC PARTICIPANT P1 – RAW DATA


MEAN FORCE GENERATED DURING GLENOHUMERAL FLEXION
P1 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
PARAPLEGIC PARTICIPANT P2 – RAW DATA



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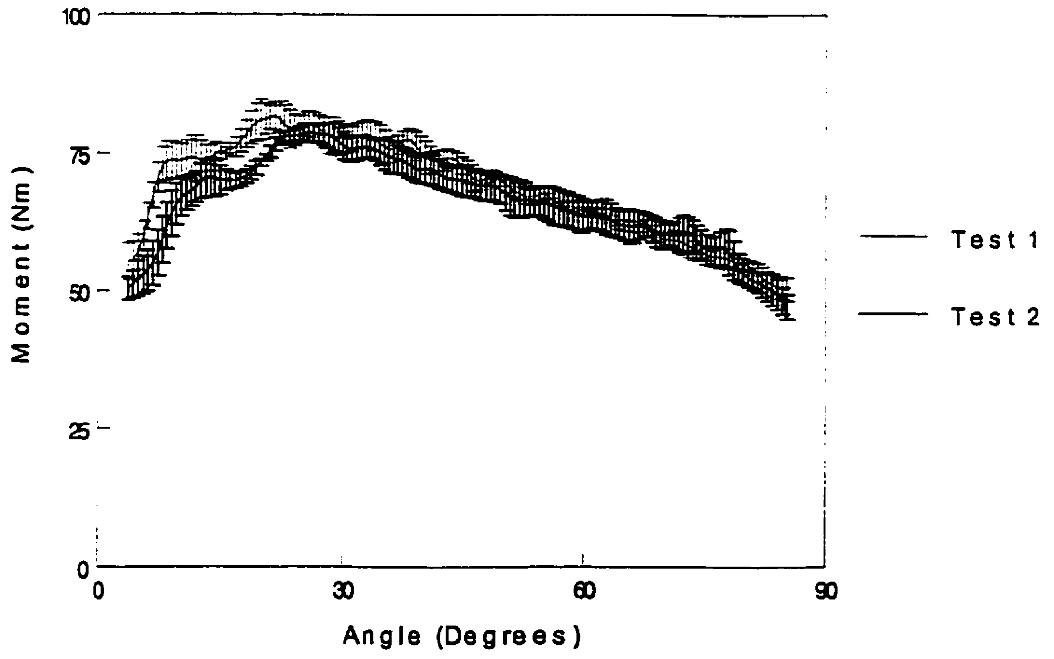
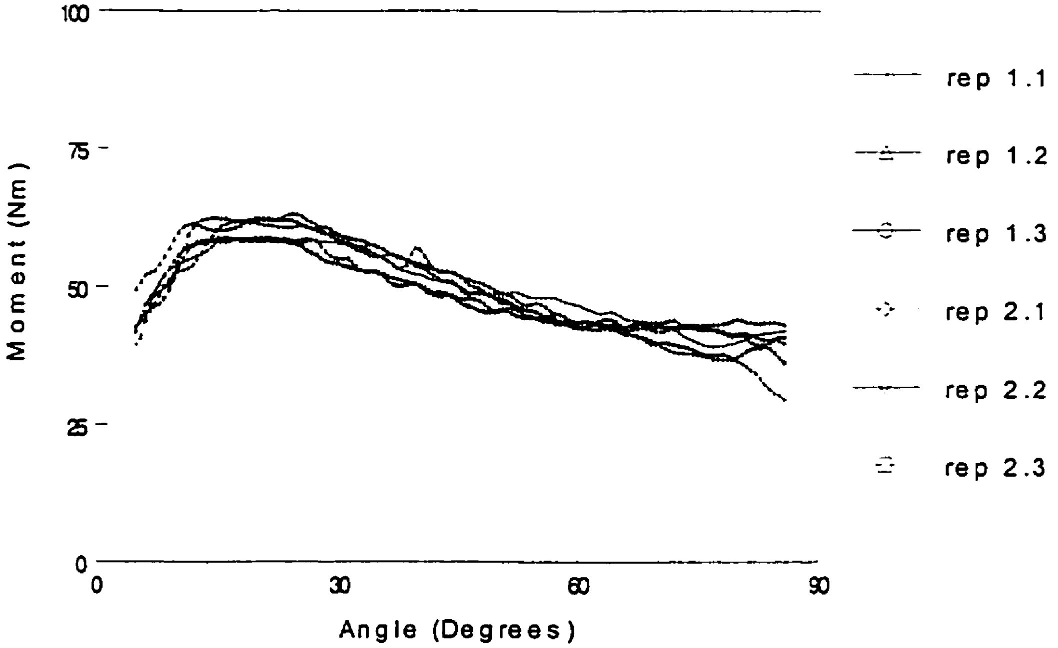
P2 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
PARAPLEGIC PARTICIPANT P3 – RAW DATA



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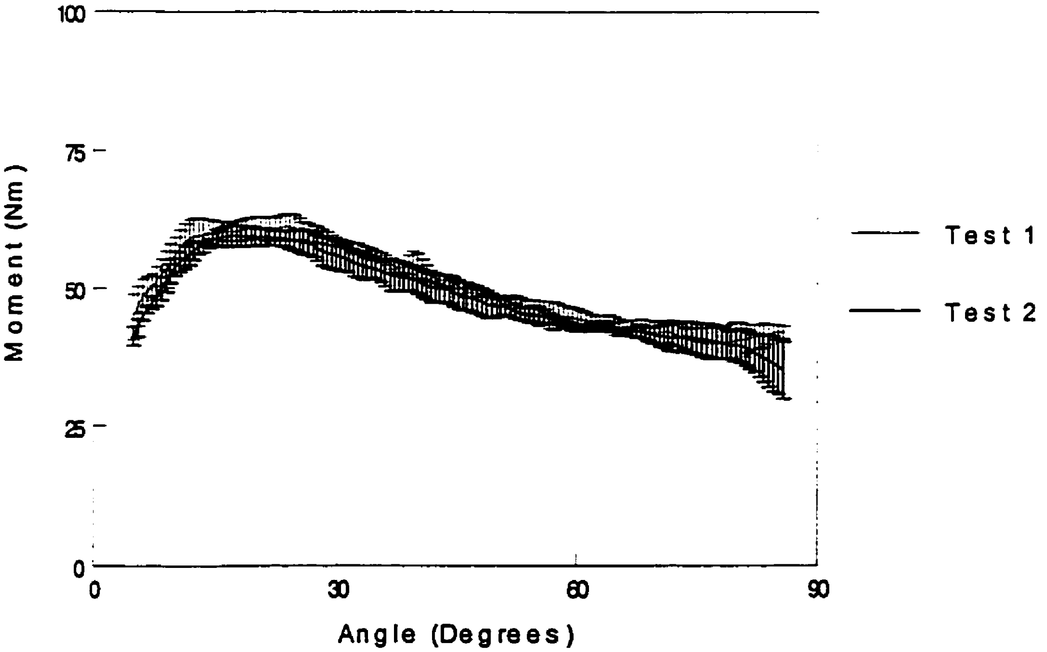
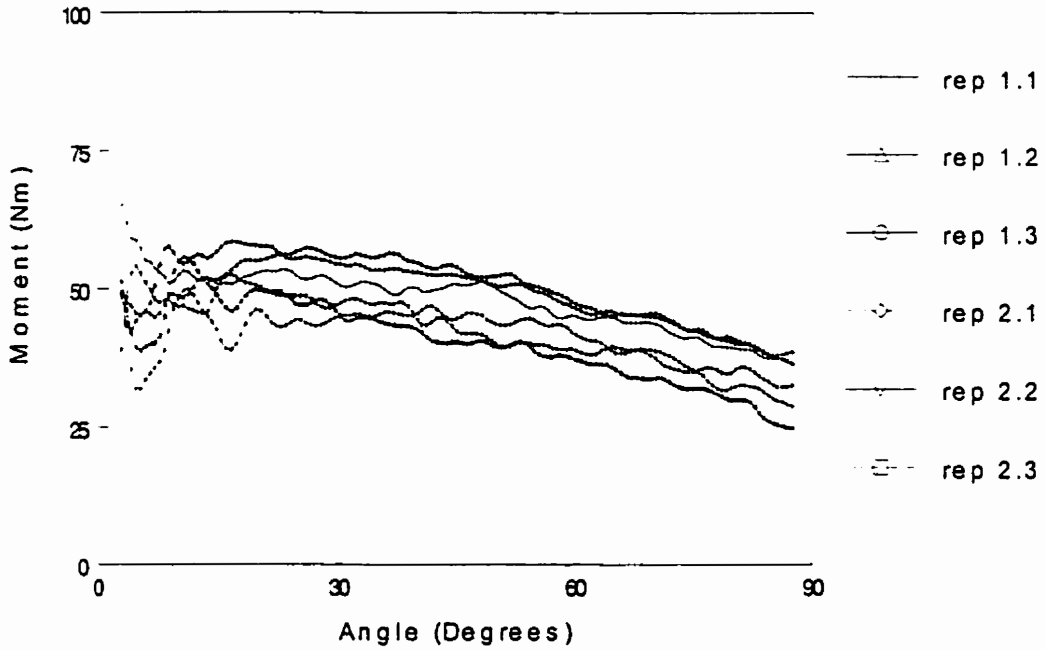
P3 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants



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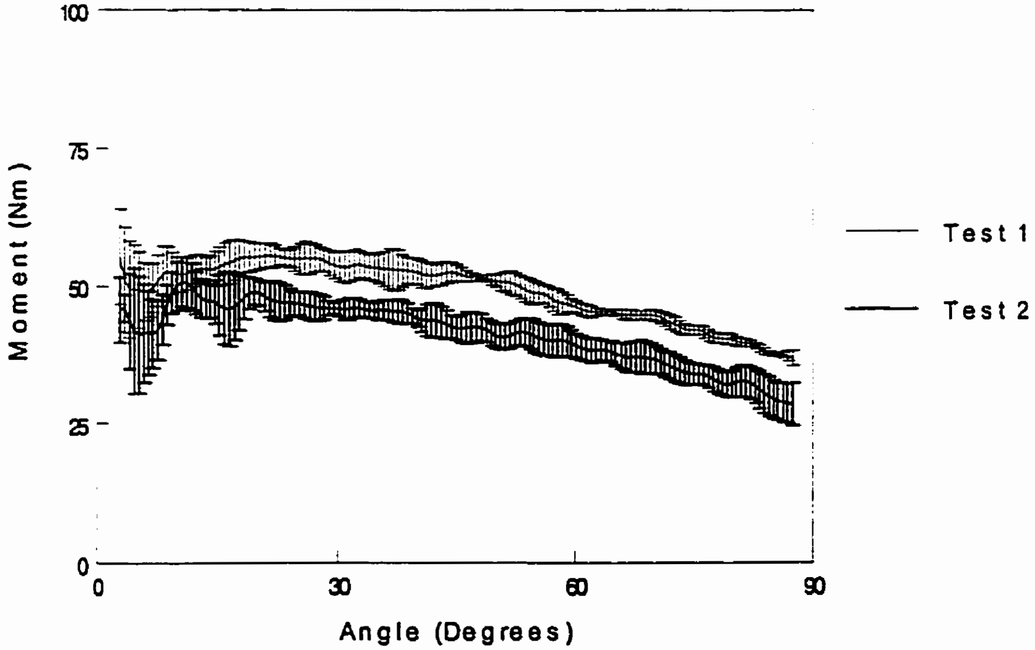
PARAPLEGIC PARTICIPANT P4 – RAW DATA



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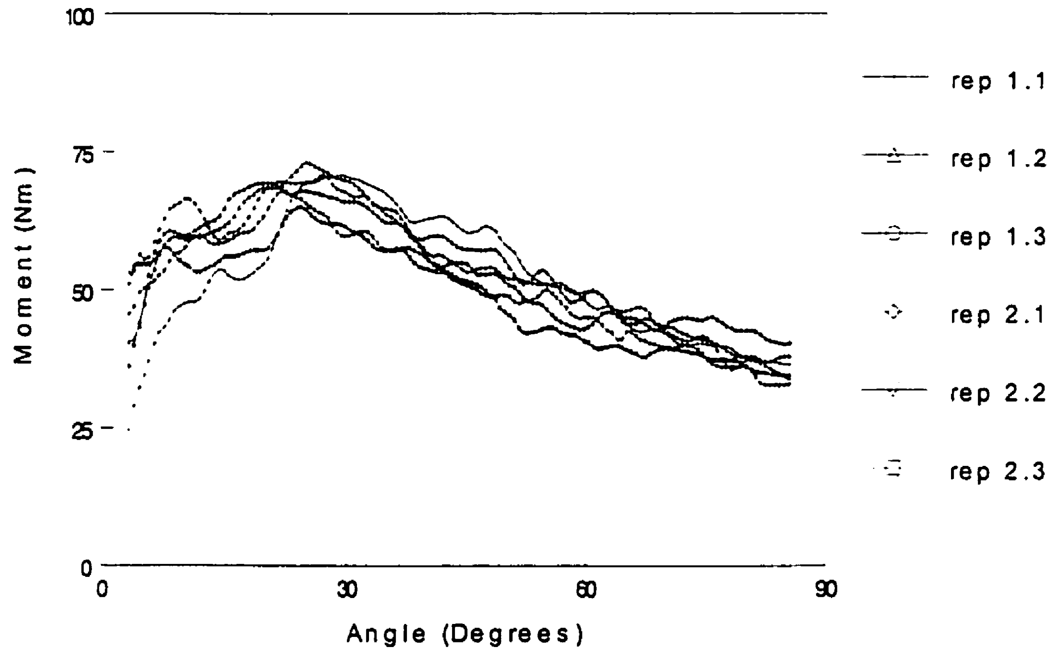
P4 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants



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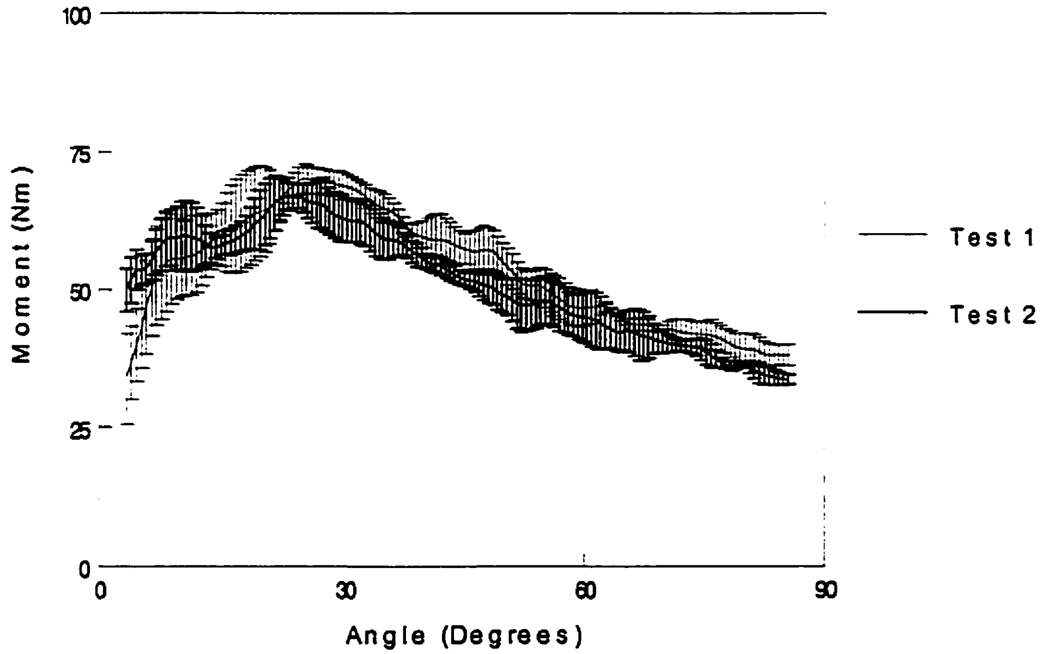
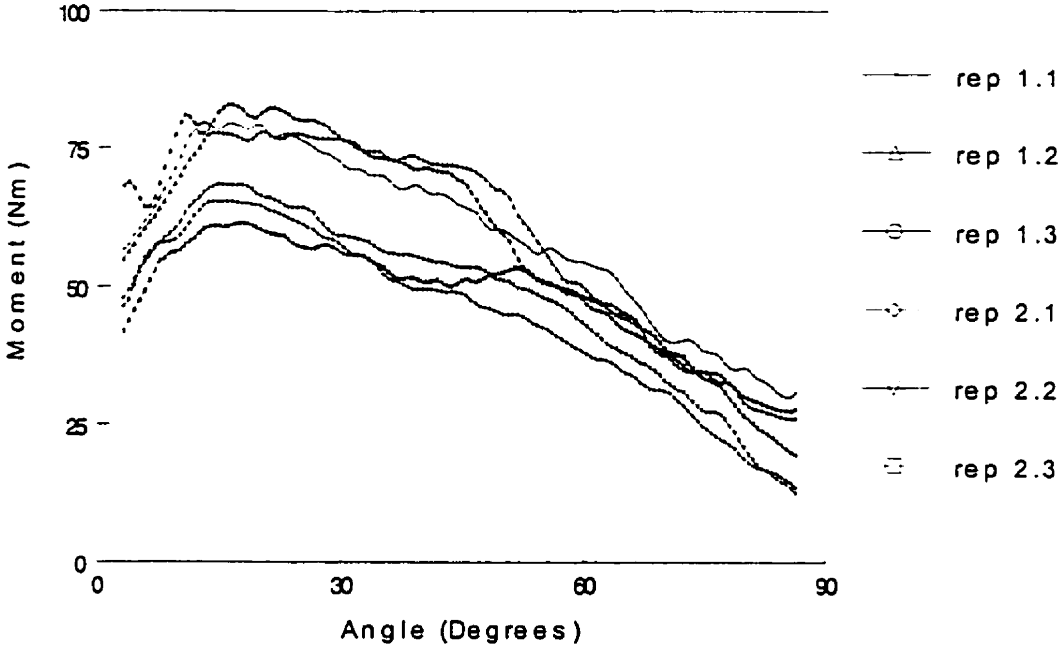
PARAPLEGIC PARTICIPANT P5 – RAW DATA



MEAN FORCE GENERATED DURING GLENOHUMERAL FLEXION
P5 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
PARAPLEGIC PARTICIPANT P6 – RAW DATA



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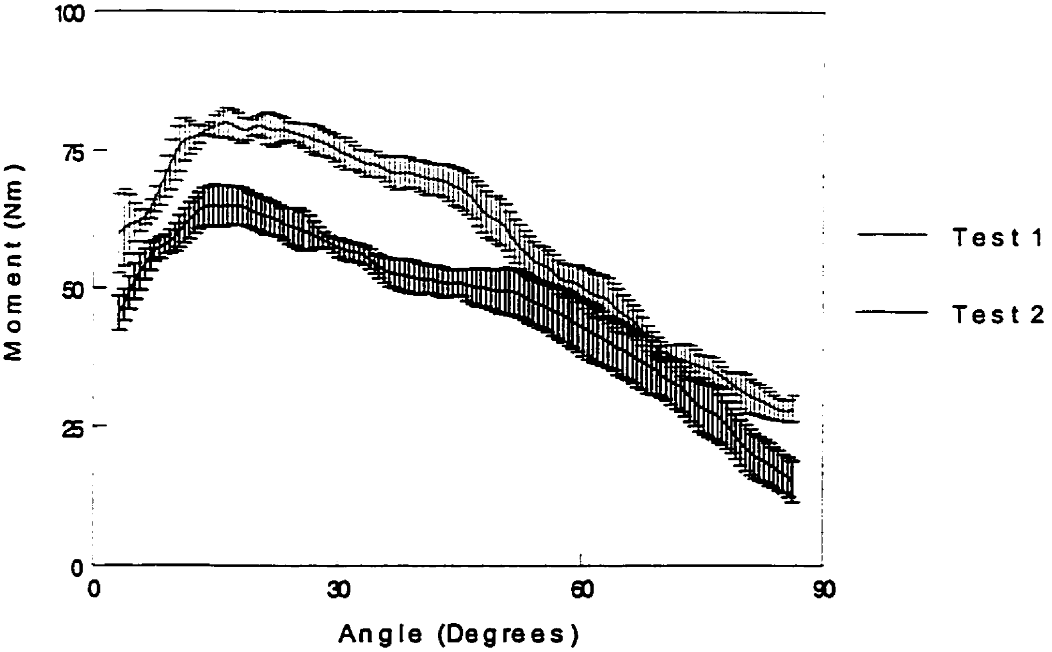
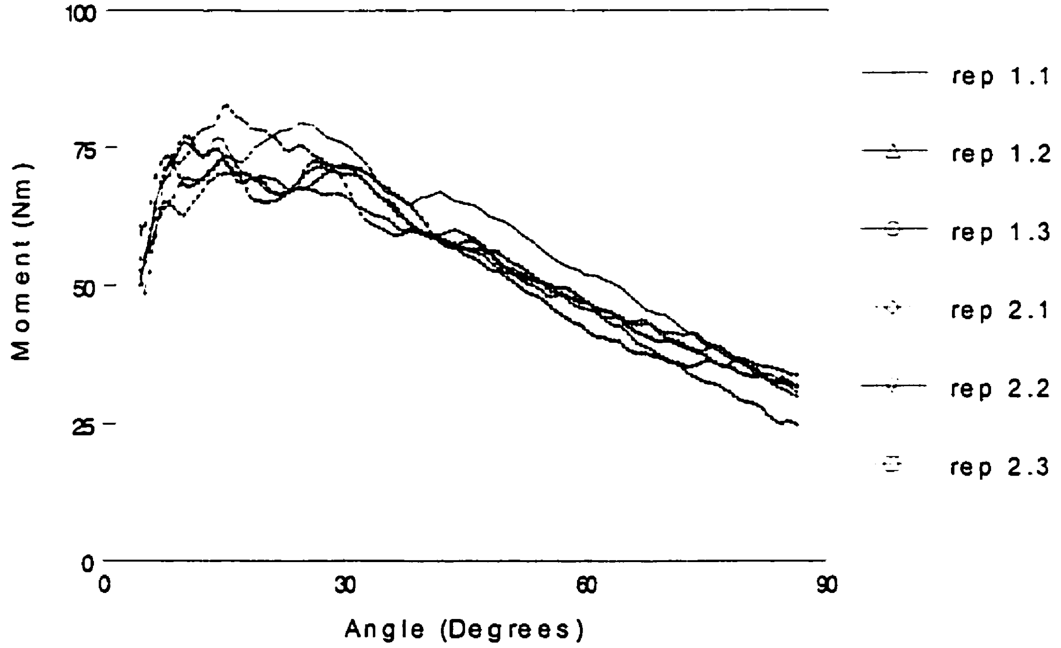
P6 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
PARAPLEGIC PARTICIPANT P7 – RAW DATA



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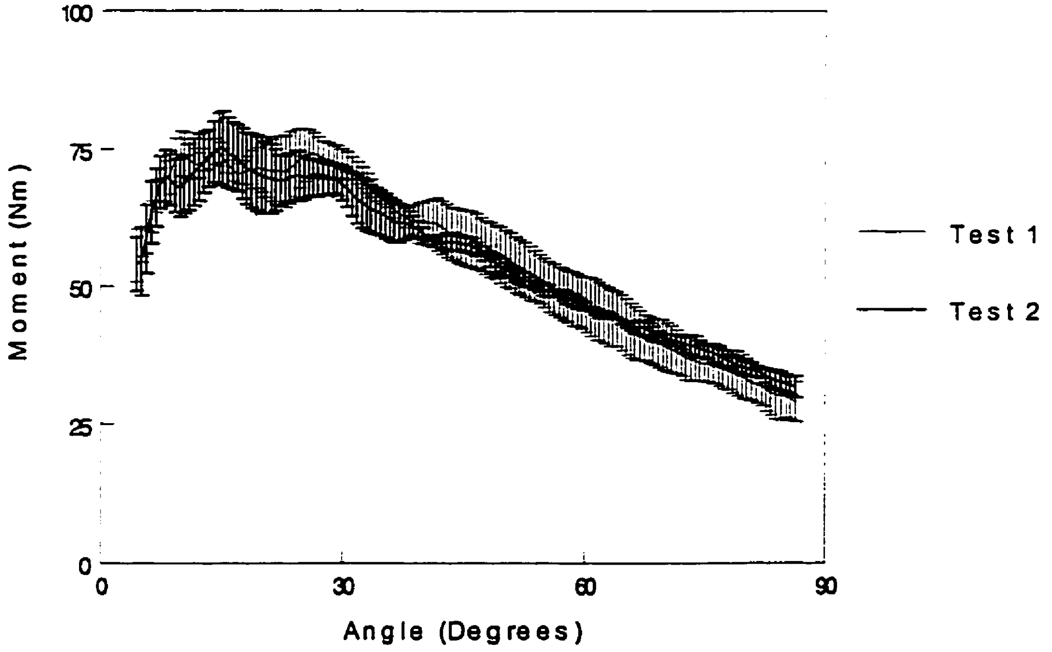
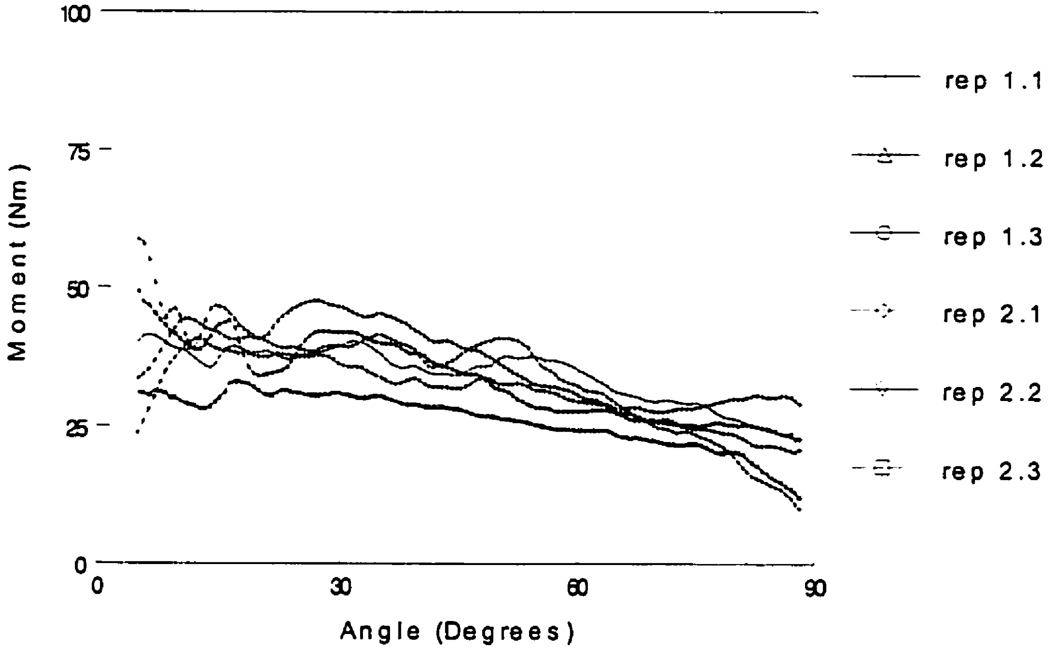
P7 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL FLEXION
PARAPLEGIC PARTICIPANT P8 – RAW DATA



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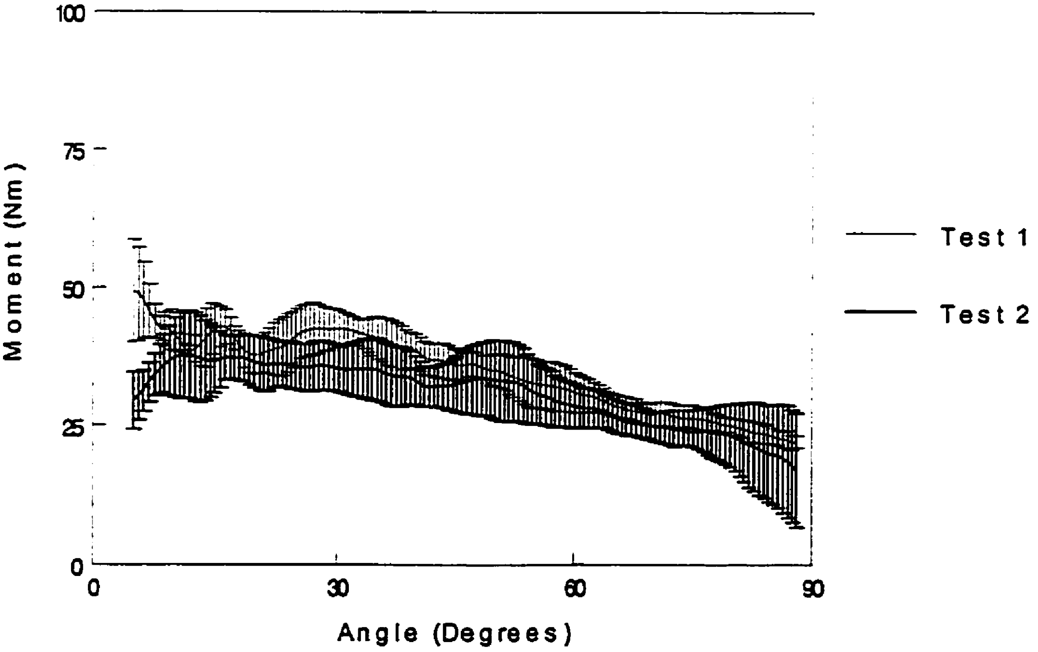
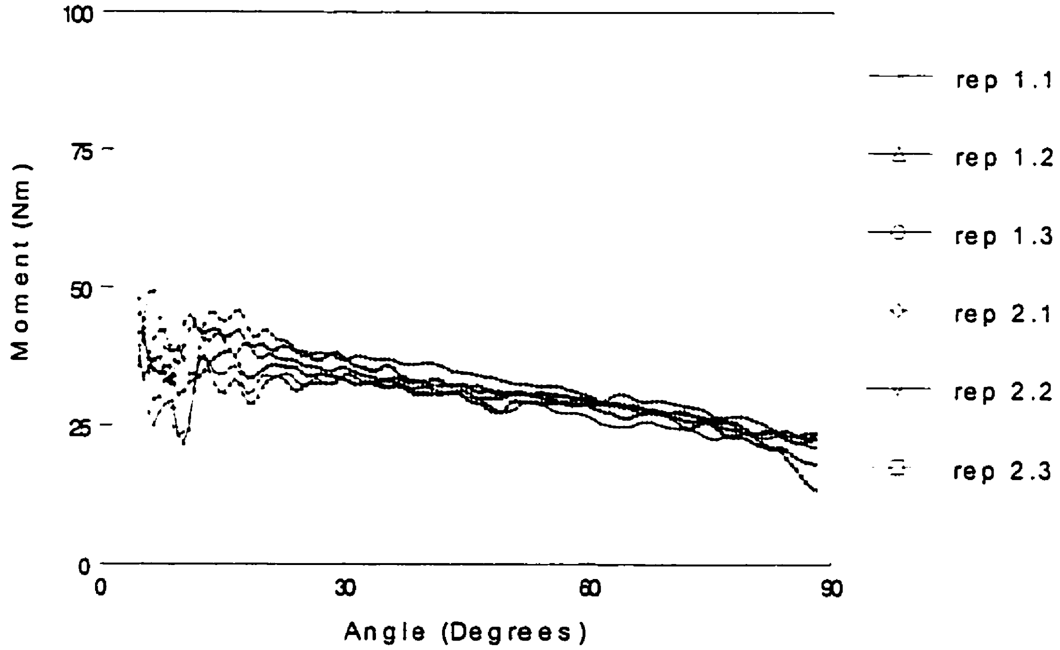
P8 – STANDARD DEVIATION

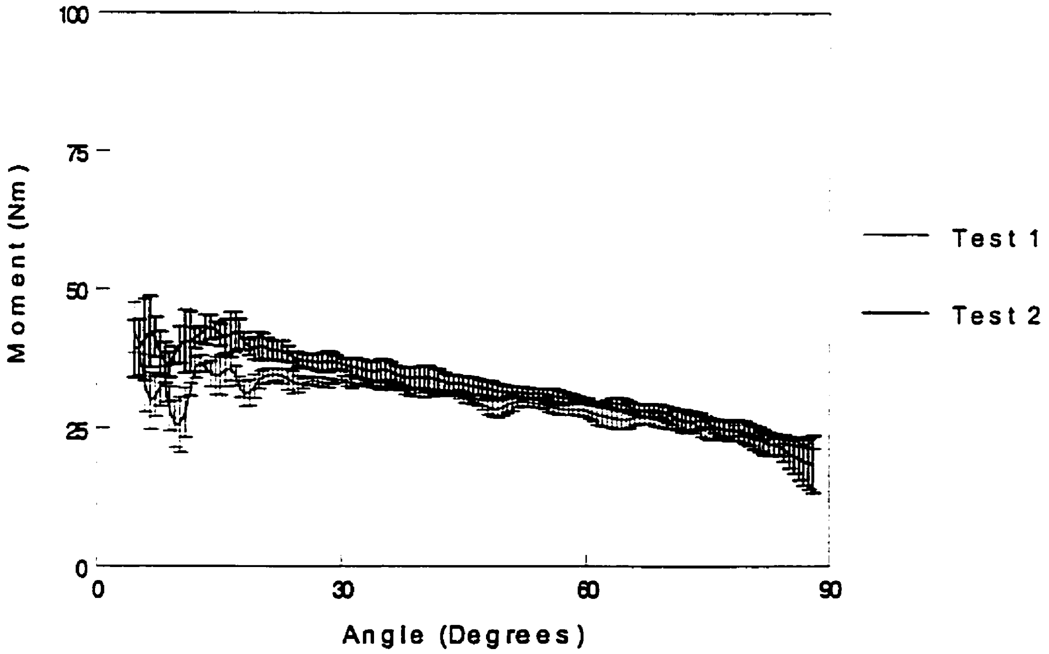


Appendix E – Moment / Angle Graphs for Individual Participants

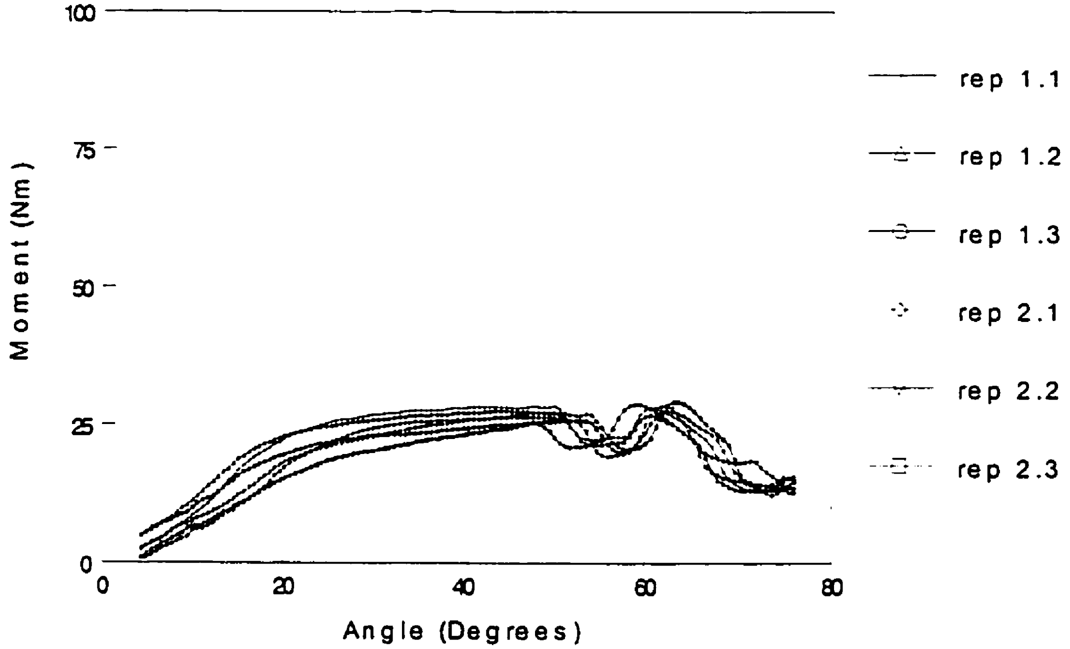
FORCE GENERATED DURING GLENOHUMERAL FLEXION
PARAPLEGIC PARTICIPANT P9 – RAW DATA



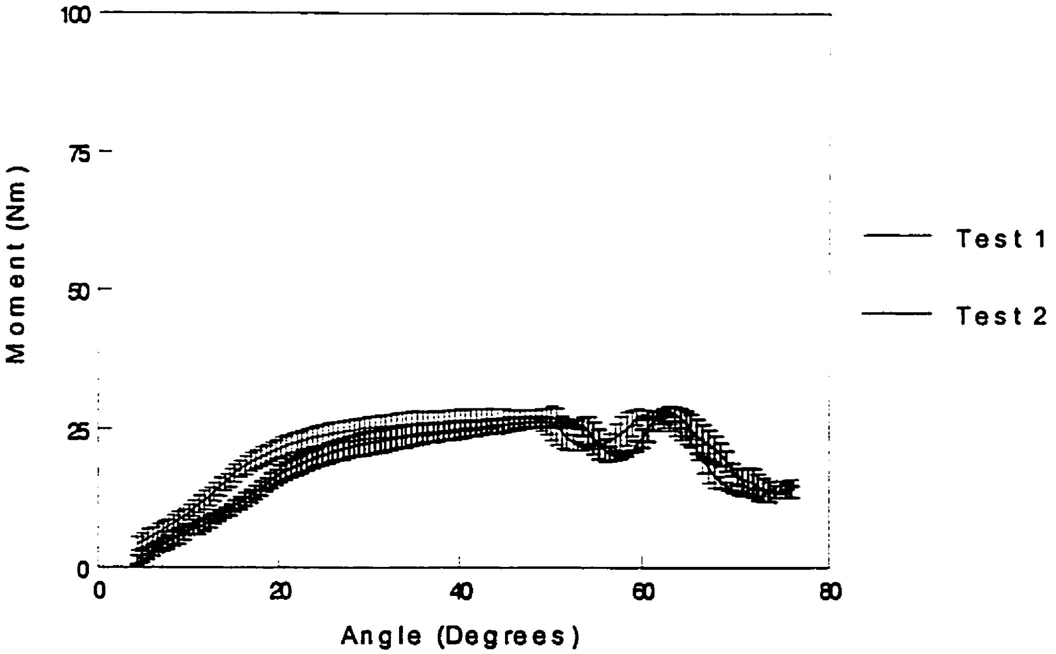
MEAN FORCE GENERATED DURING GLENOHUMERAL FLEXION
P9 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

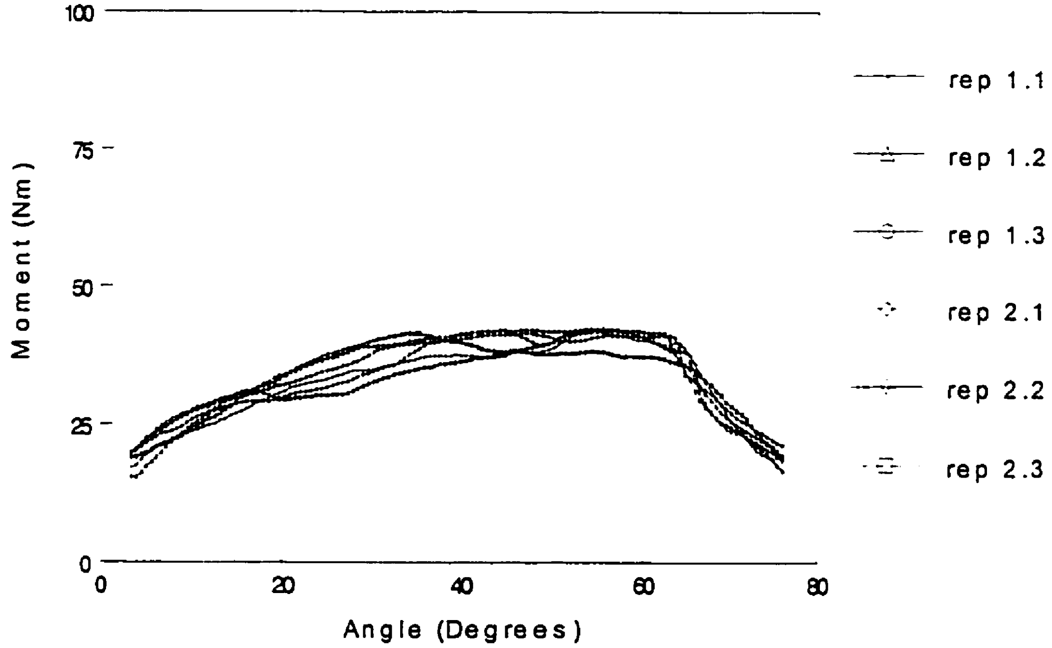


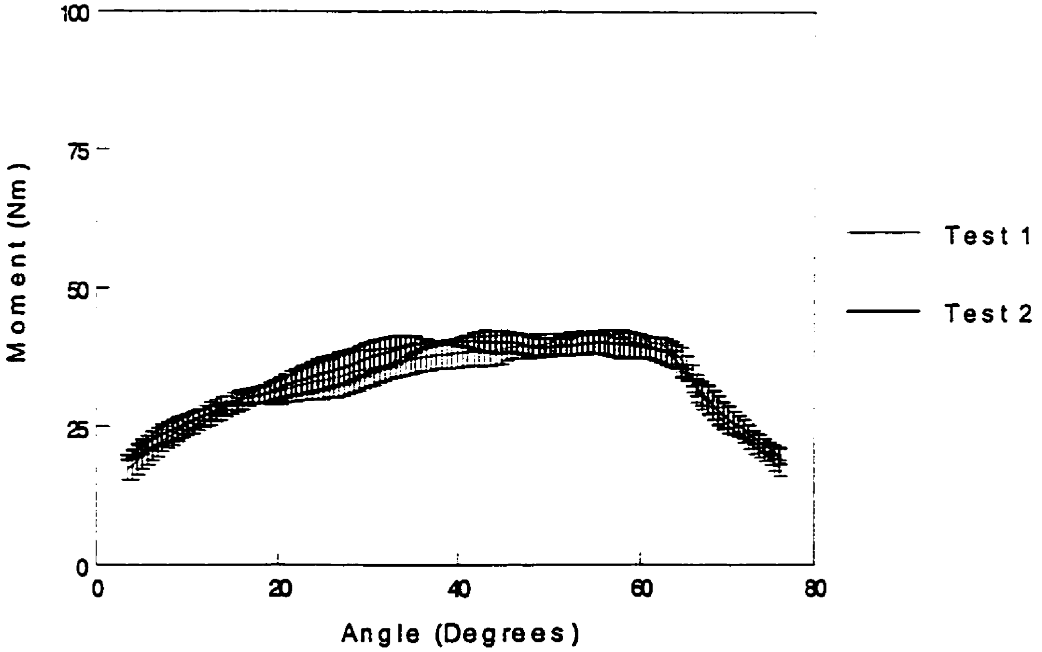
FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q1 – RAW DATA



MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q1 – STANDARD DEVIATION

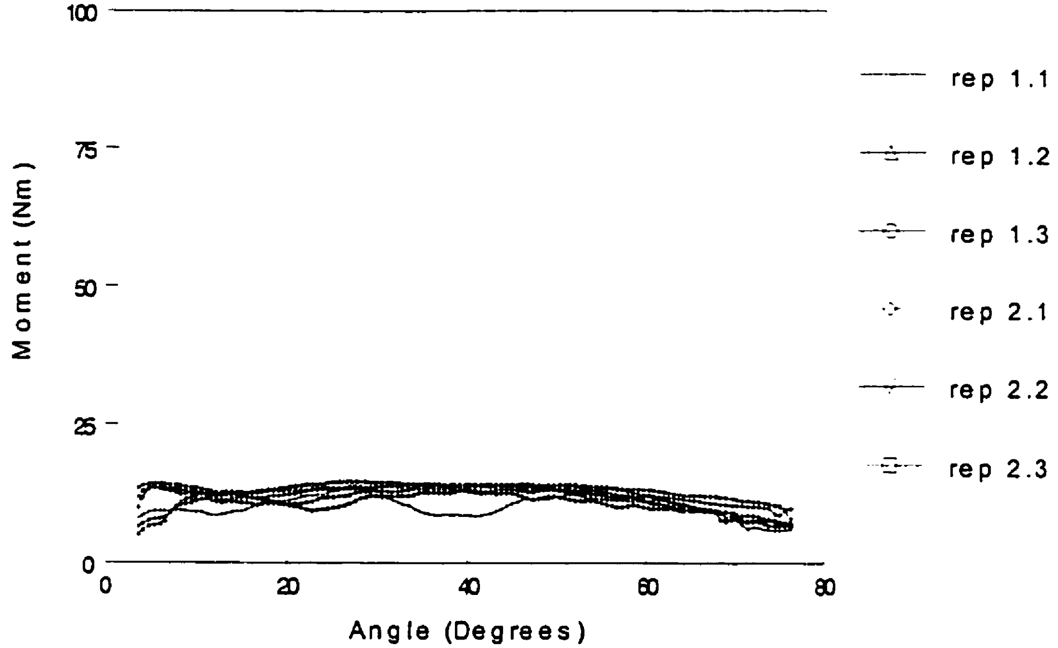
Appendix E – Moment / Angle Graphs for Individual Participants


FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q2 – RAW DATA


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q2 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q3 – RAW DATA



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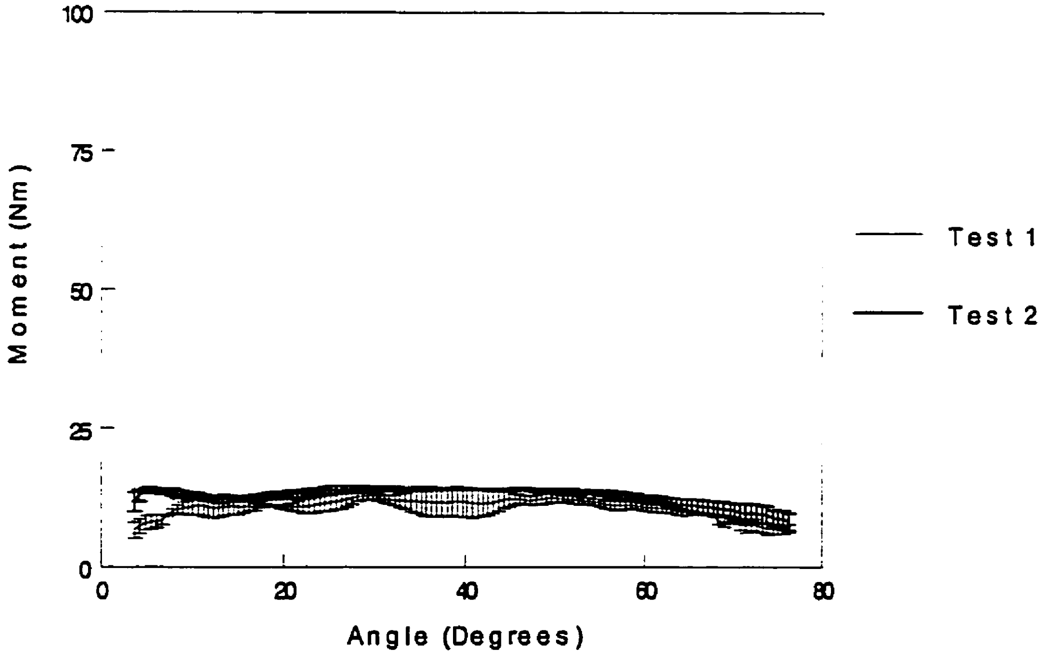
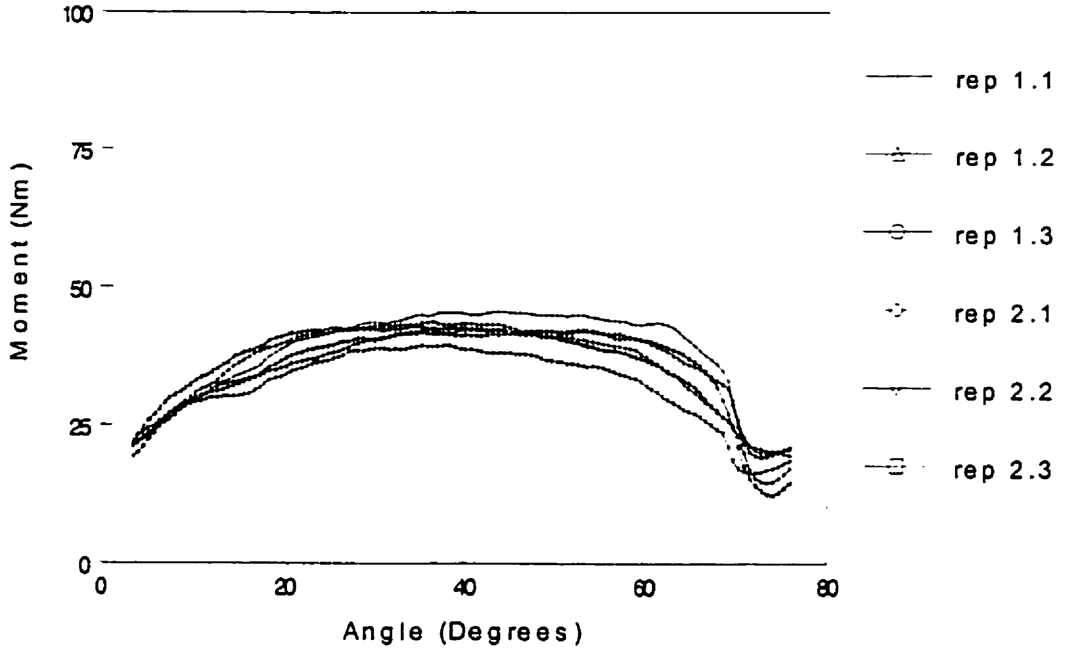
Q3 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q4 – RAW DATA



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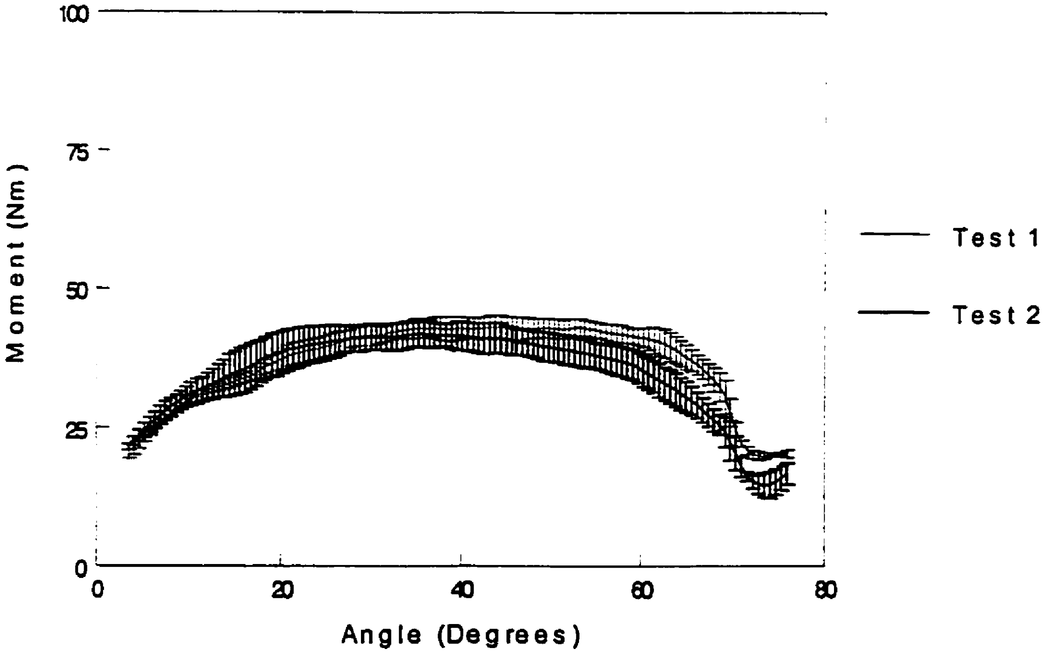
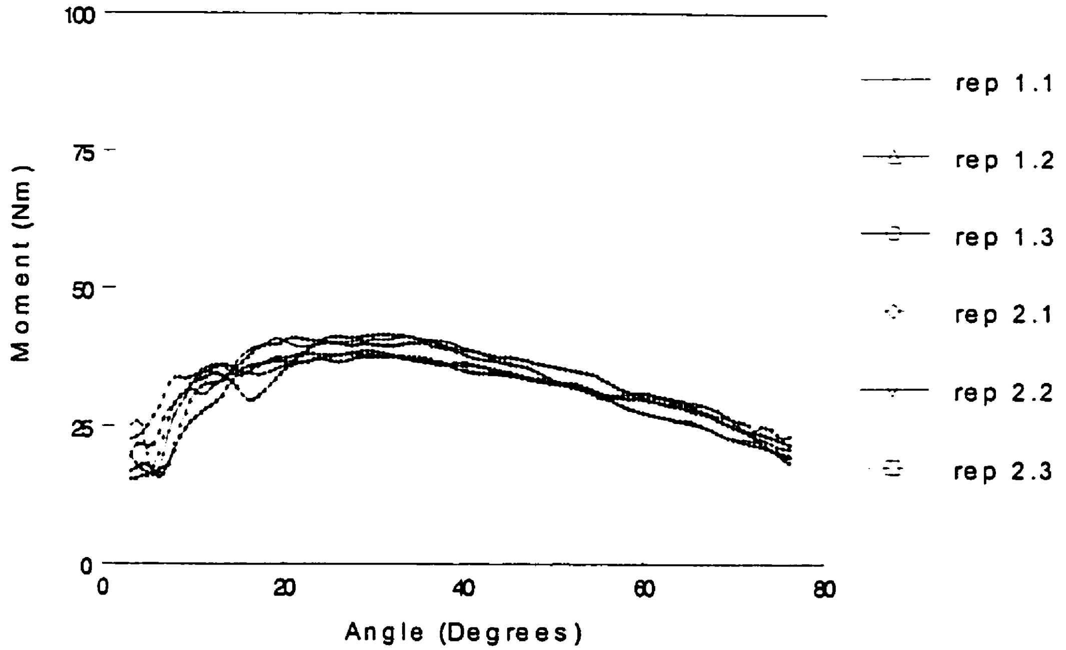
Q4 – STANDARD DEVIATION

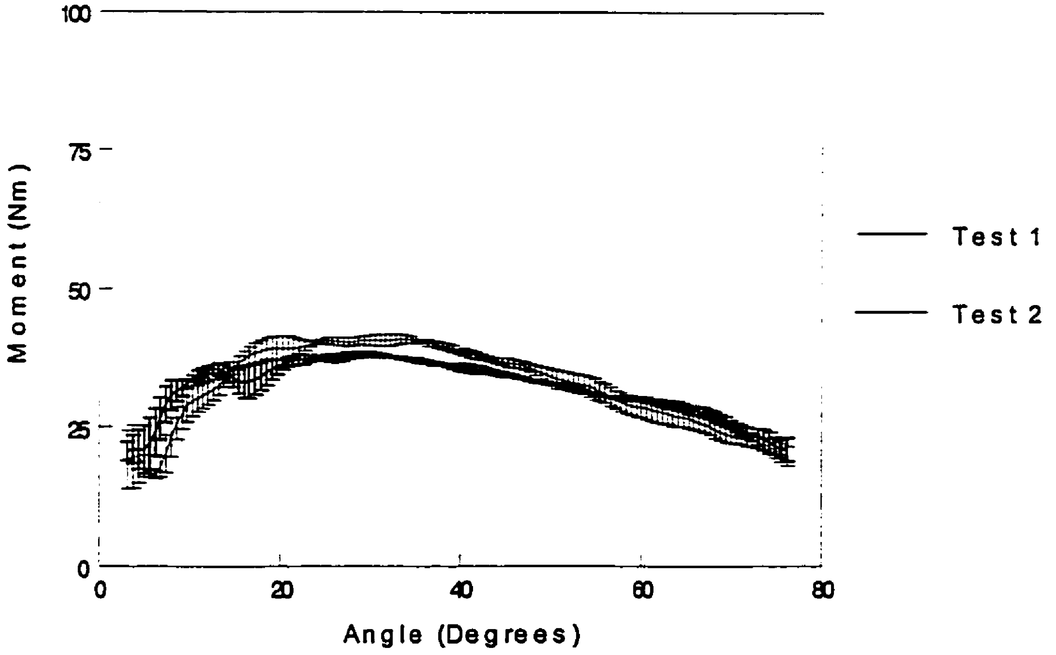


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q5 – RAW DATA

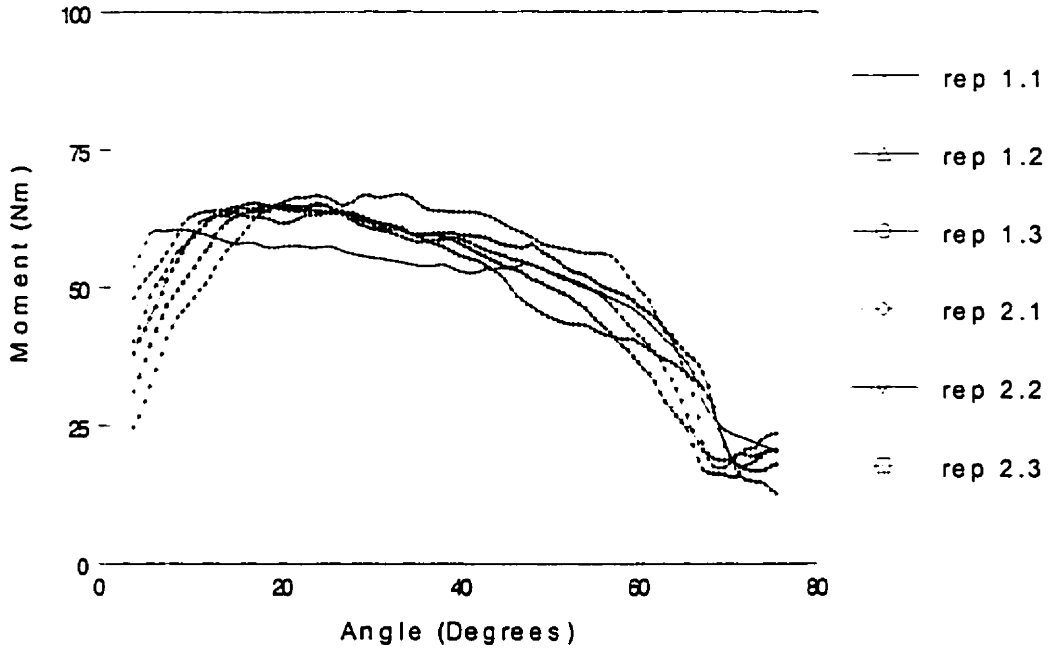


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q5 – STANDARD DEVIATION

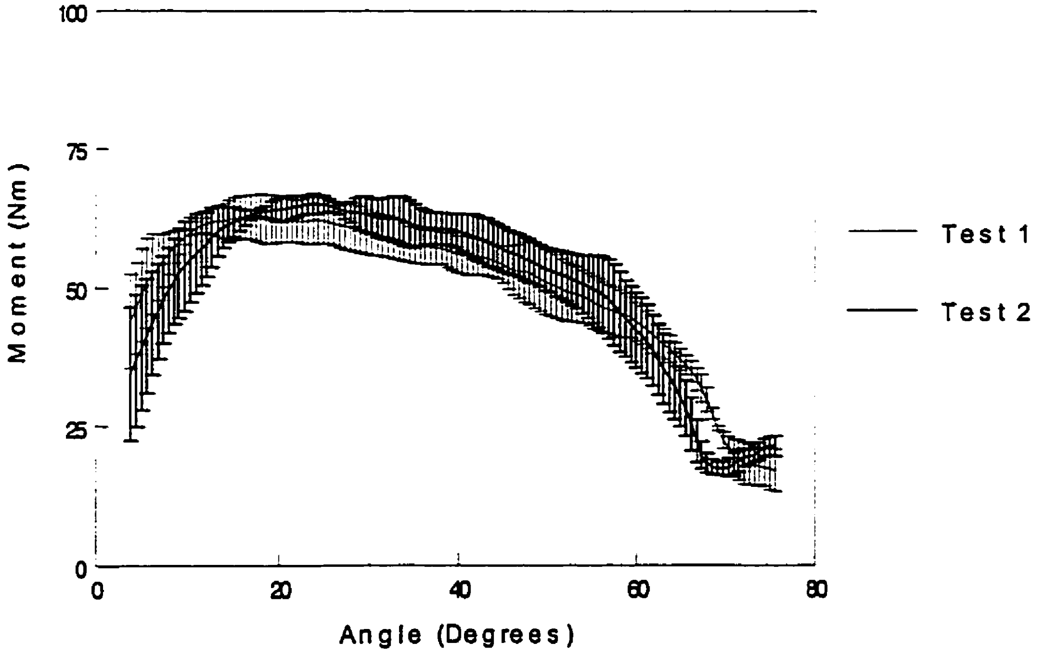


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q6 – RAW DATA

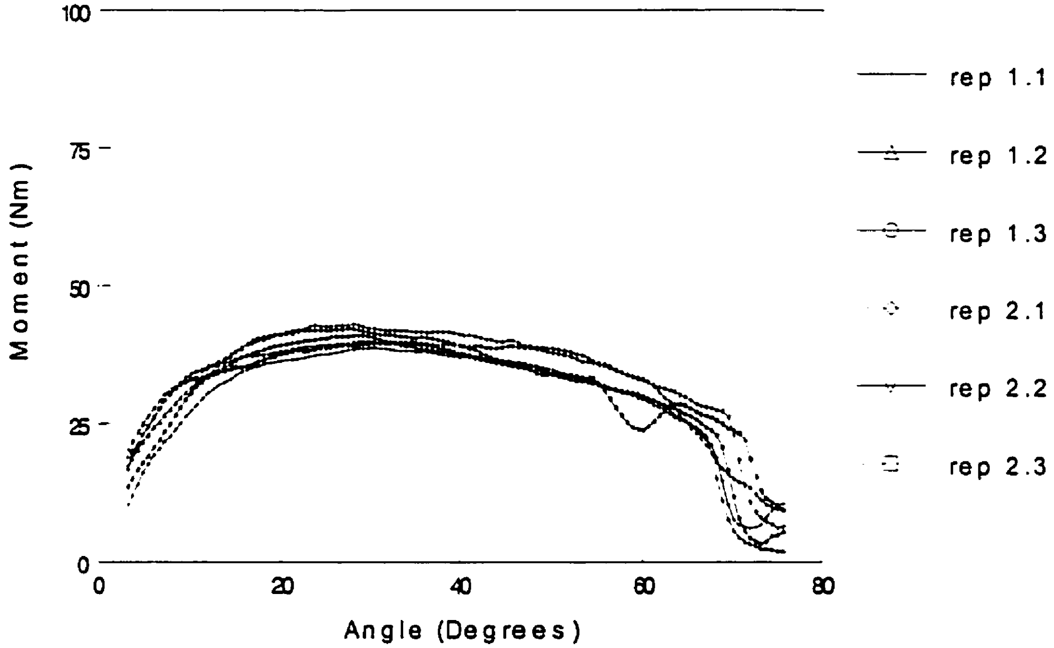


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q6 – STANDARD DEVIATION

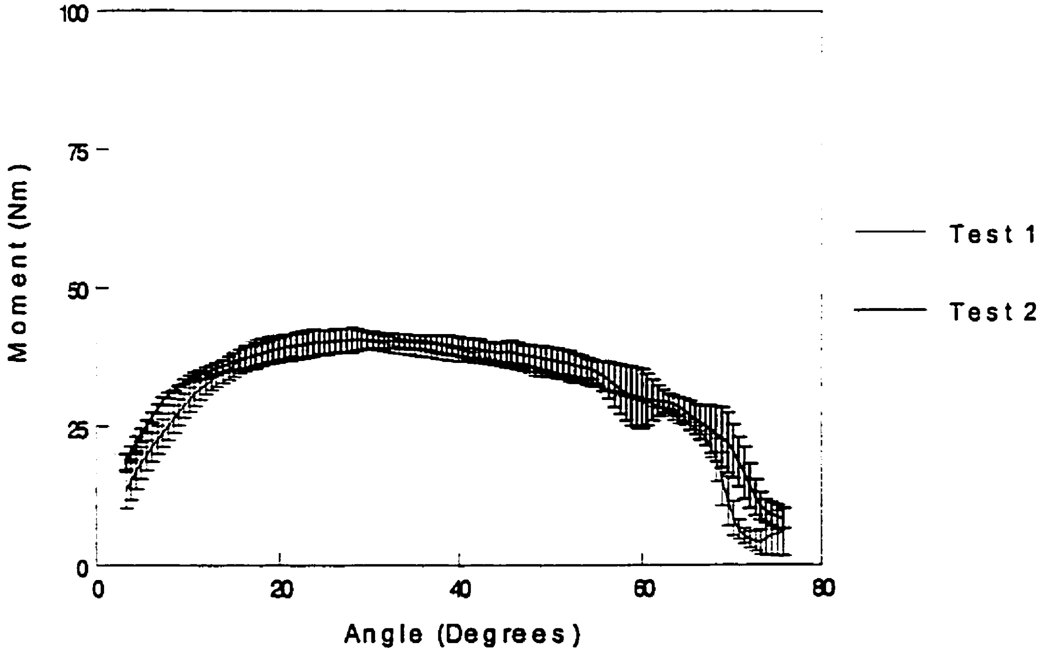


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q7 – RAW DATA

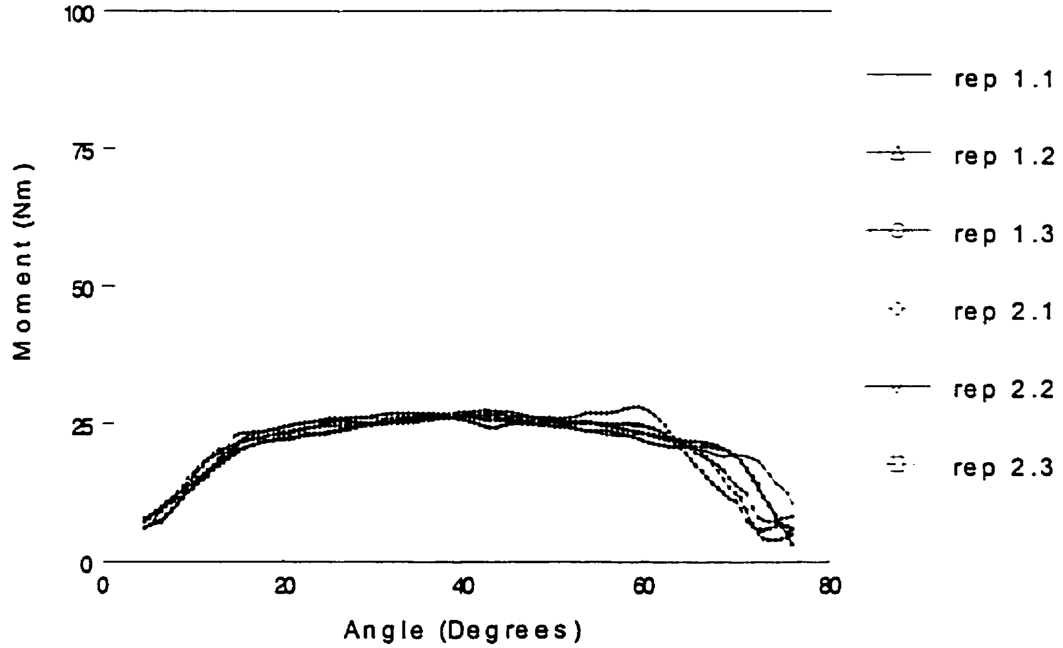


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q7 – STANDARD DEVIATION

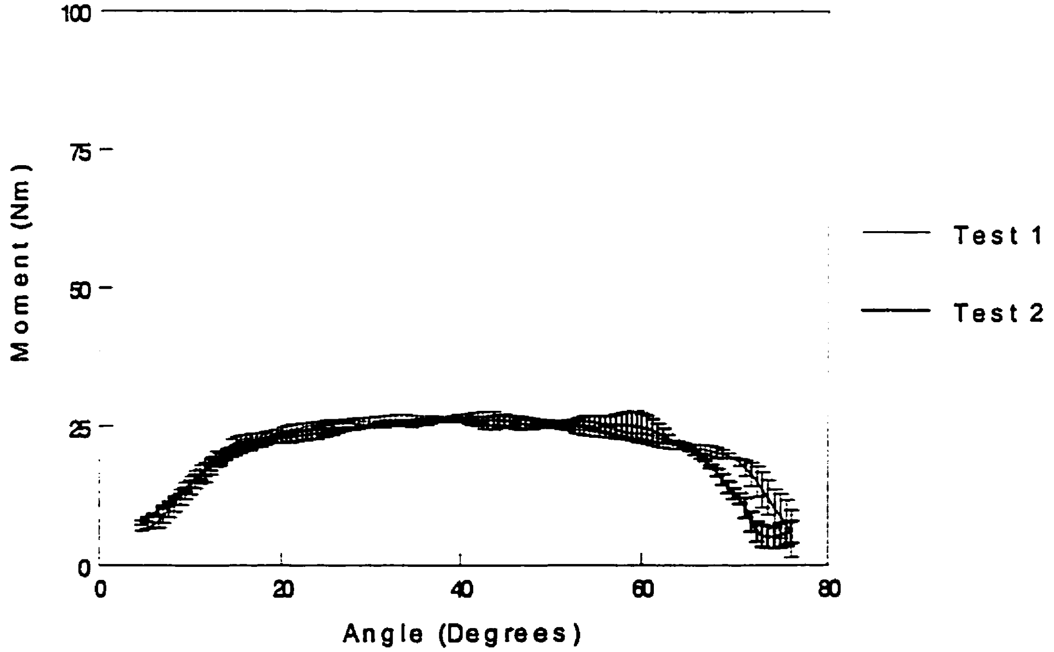


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q8 – RAW DATA

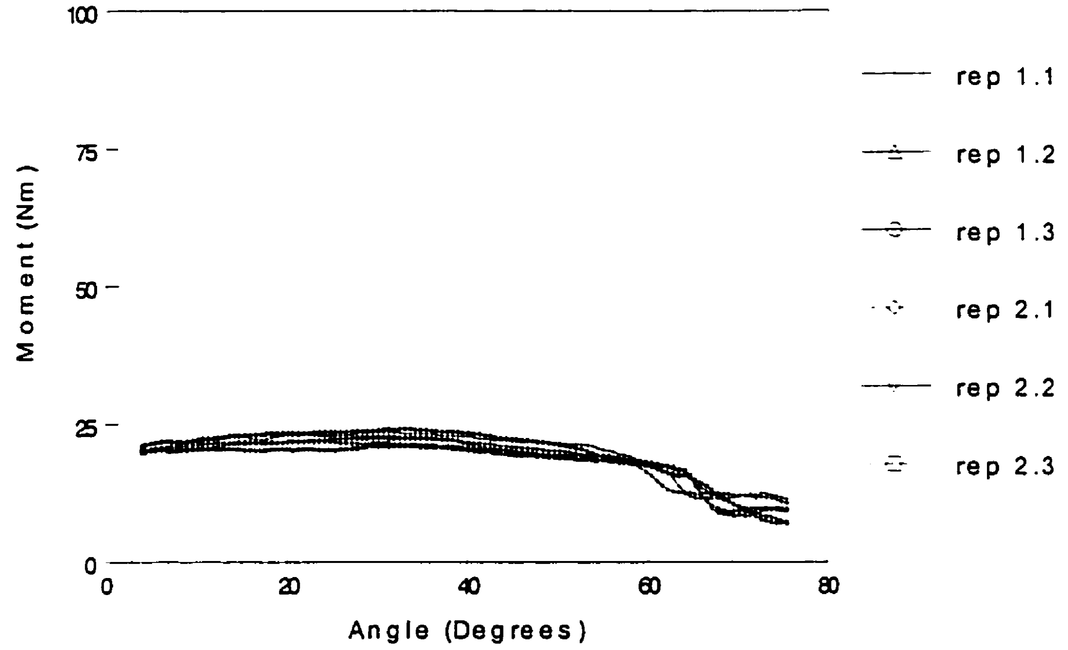


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q8 – STANDARD DEVIATION

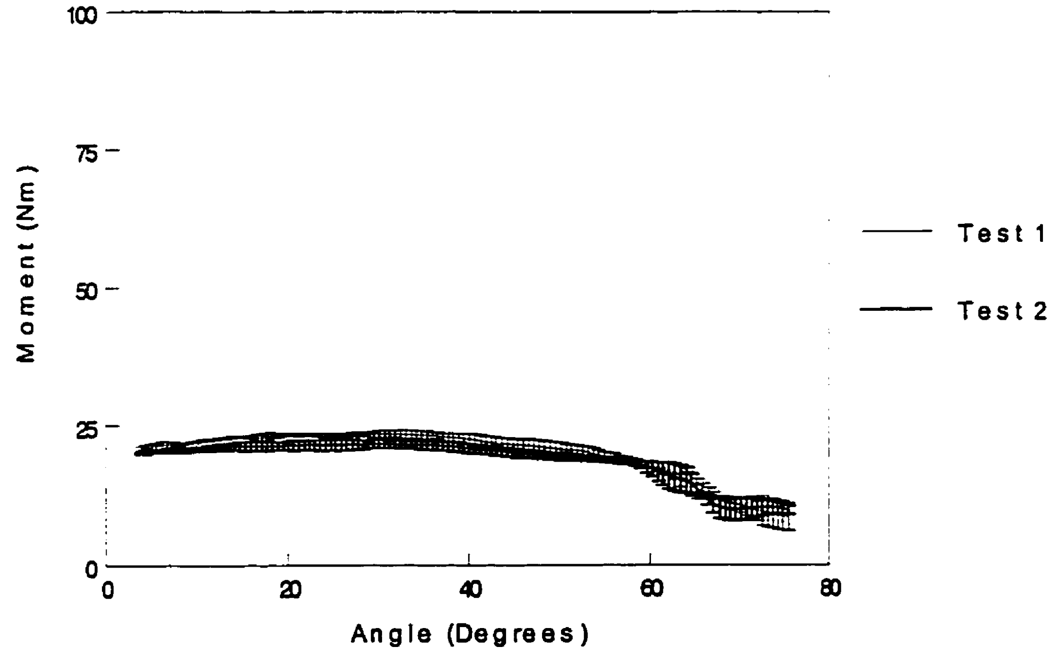


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q9 – RAW DATA

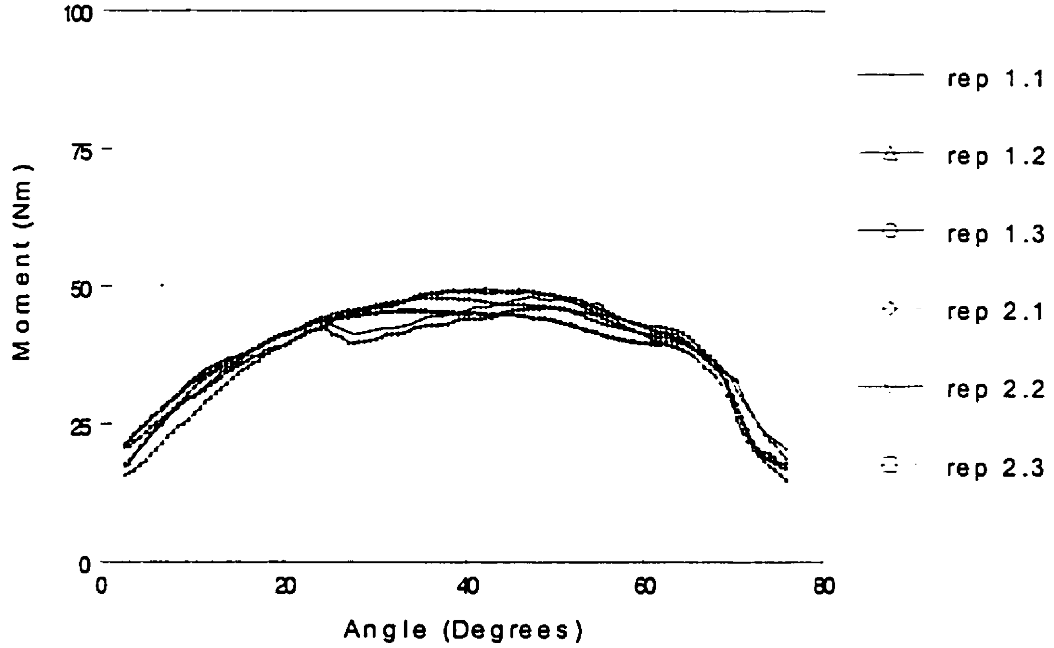


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q9 – STANDARD DEVIATION

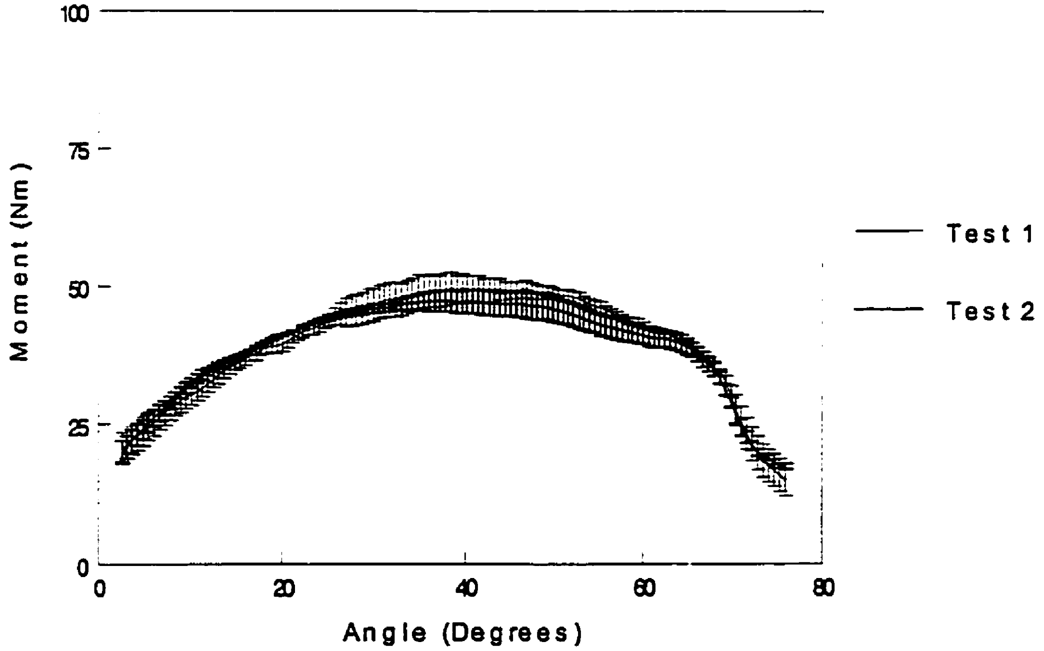


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
QUADRIPLEGIC PARTICIPANT Q10 – RAW DATA

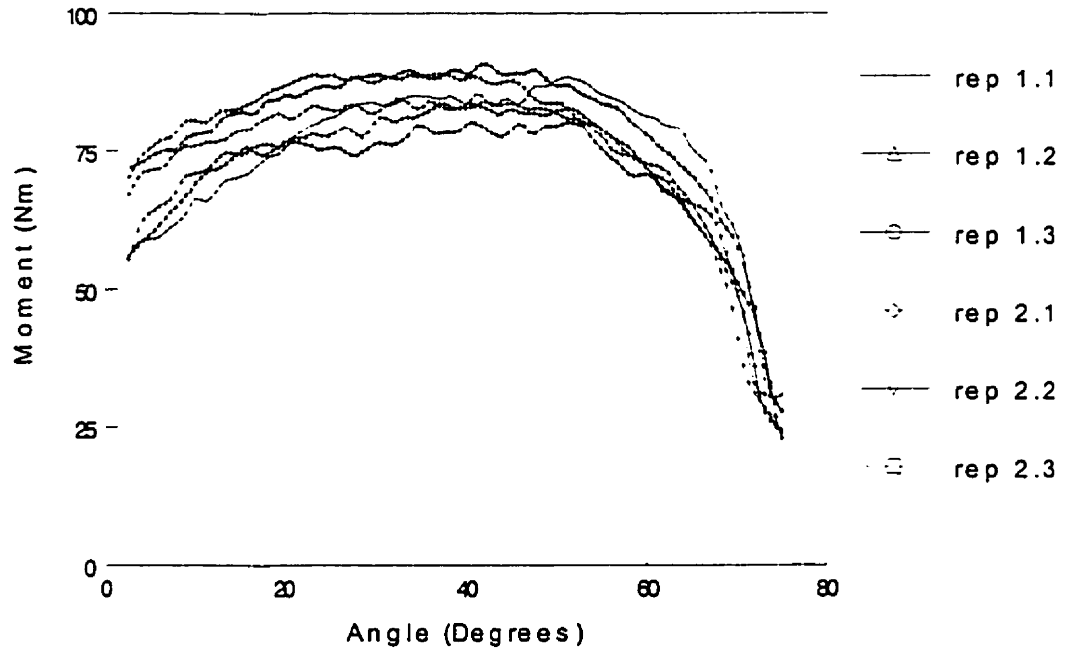


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
Q10 – STANDARD DEVIATION

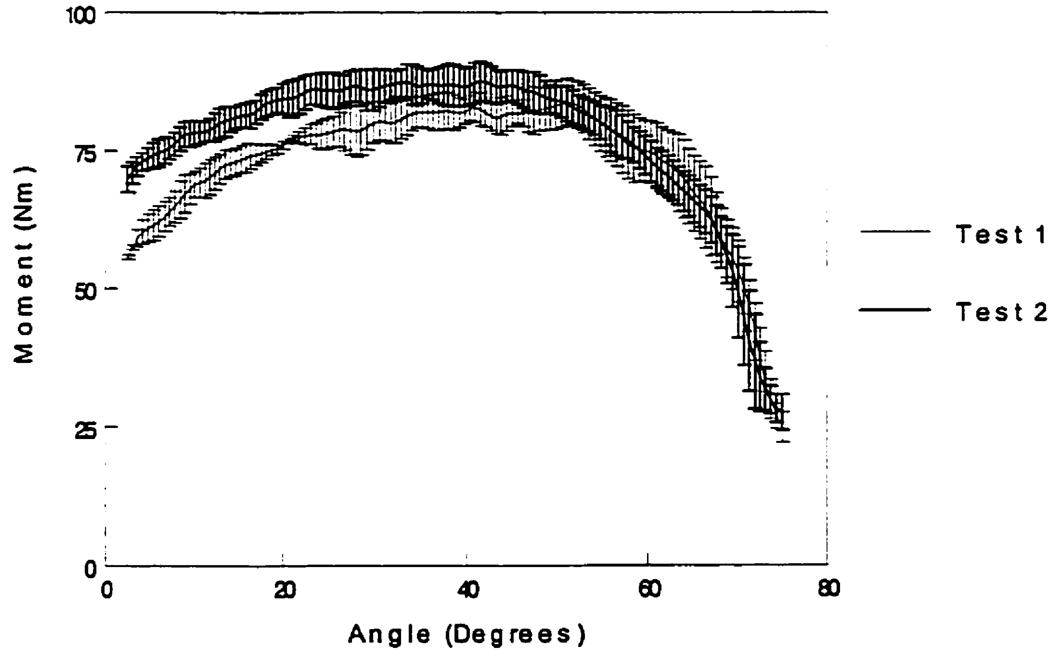


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P1 – RAW DATA

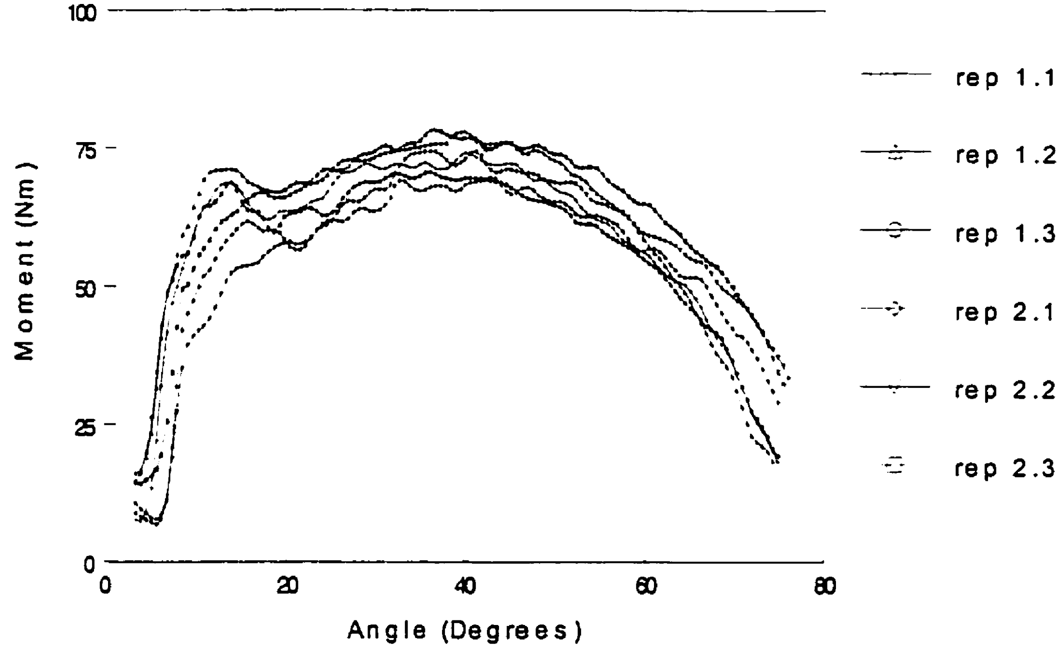


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P1 – STANDARD DEVIATION

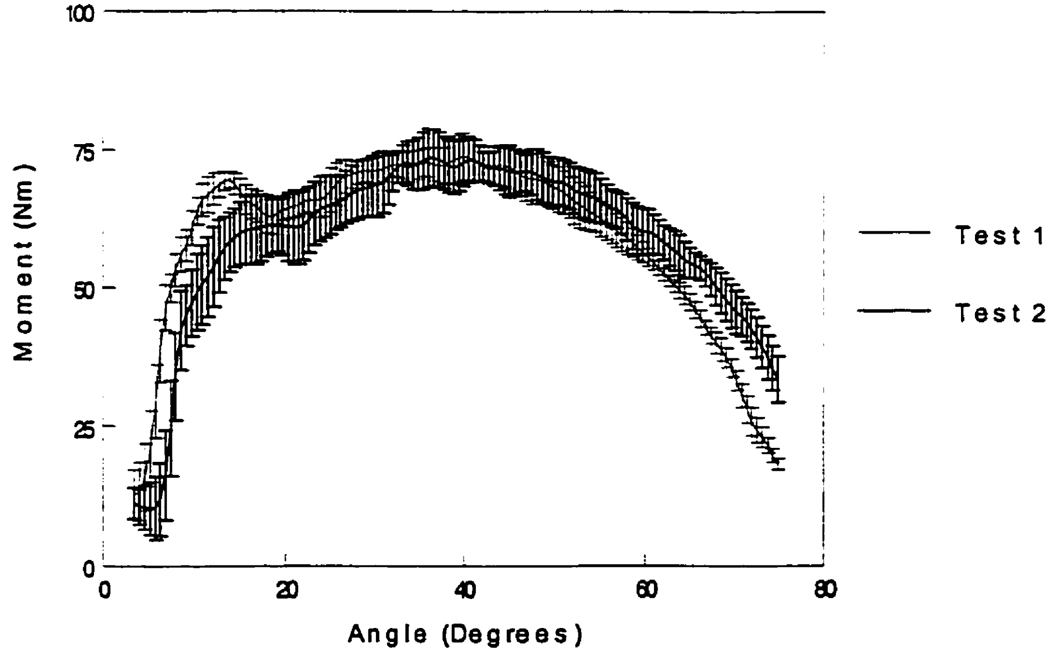


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P2 – RAW DATA

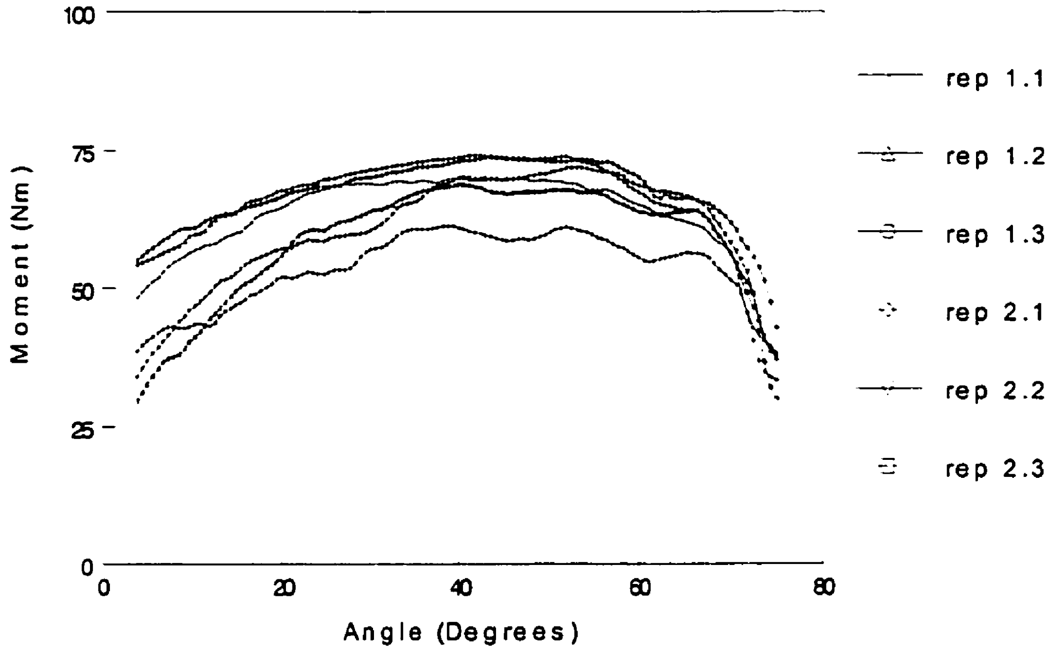


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P2 – STANDARD DEVIATION

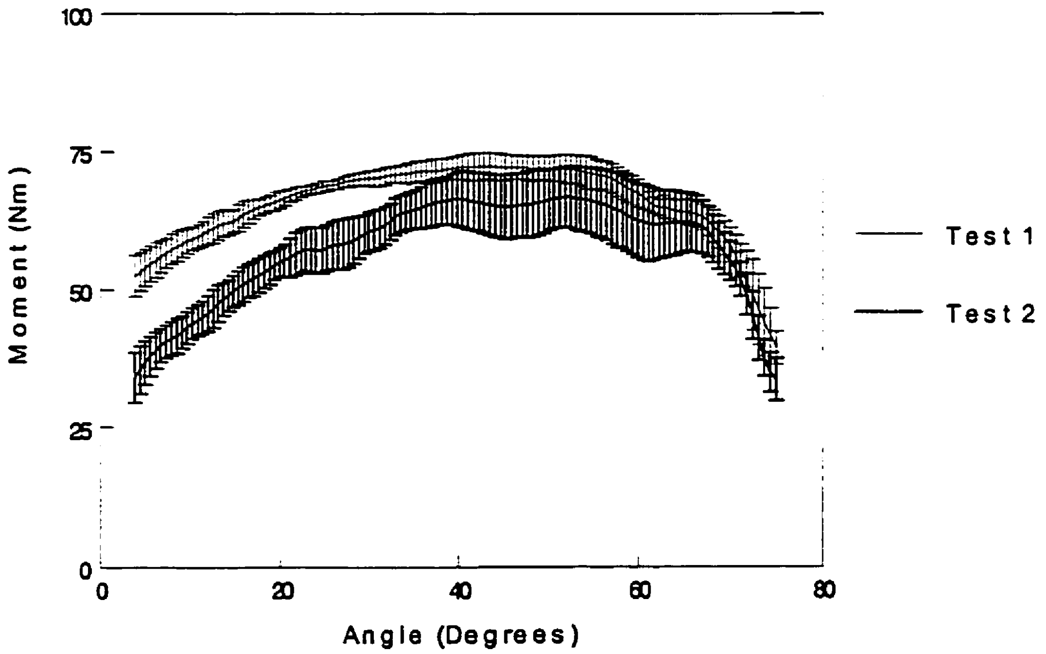


Appendix E – Moment / Angle Graphs for Individual Participants

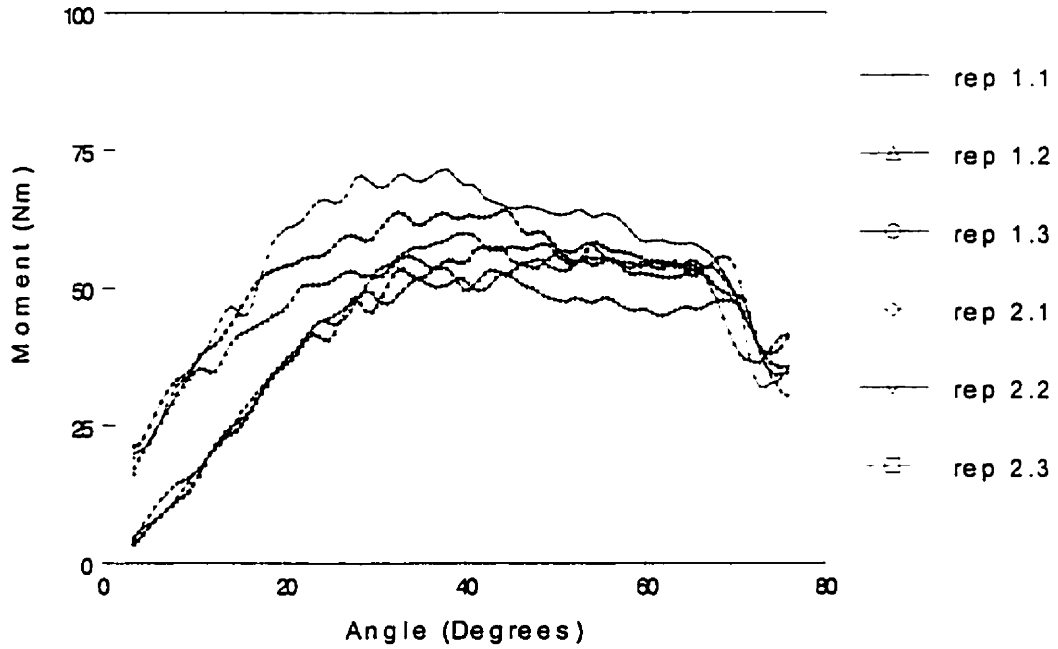
FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P3 – RAW DATA



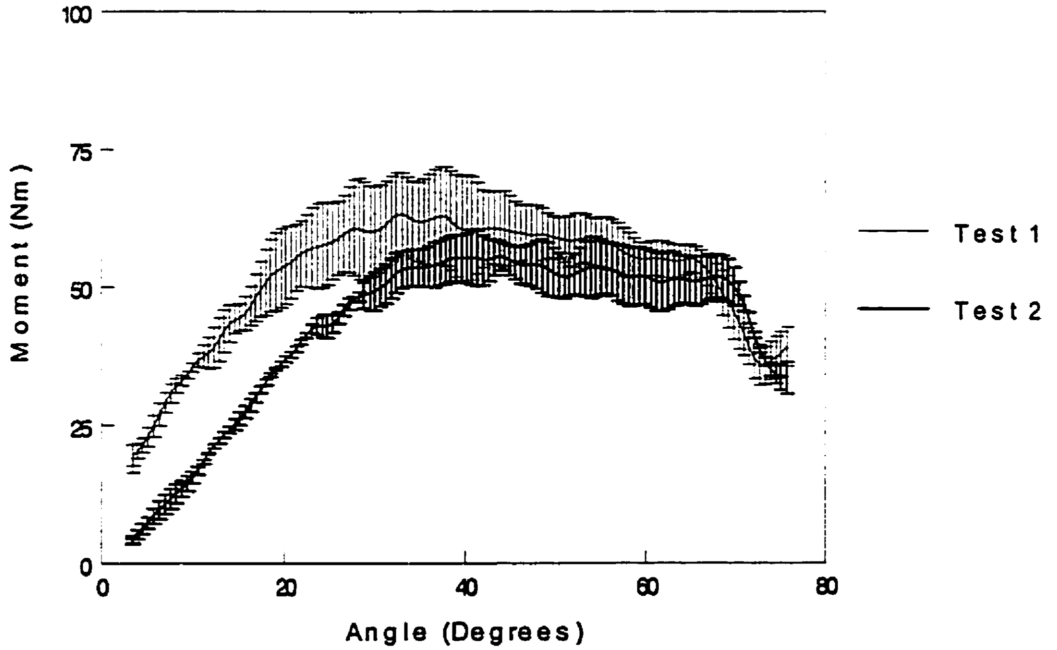
MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P3 – STANDARD DEVIATION



Appendix E – Moment / Angle Graphs for Individual Participants

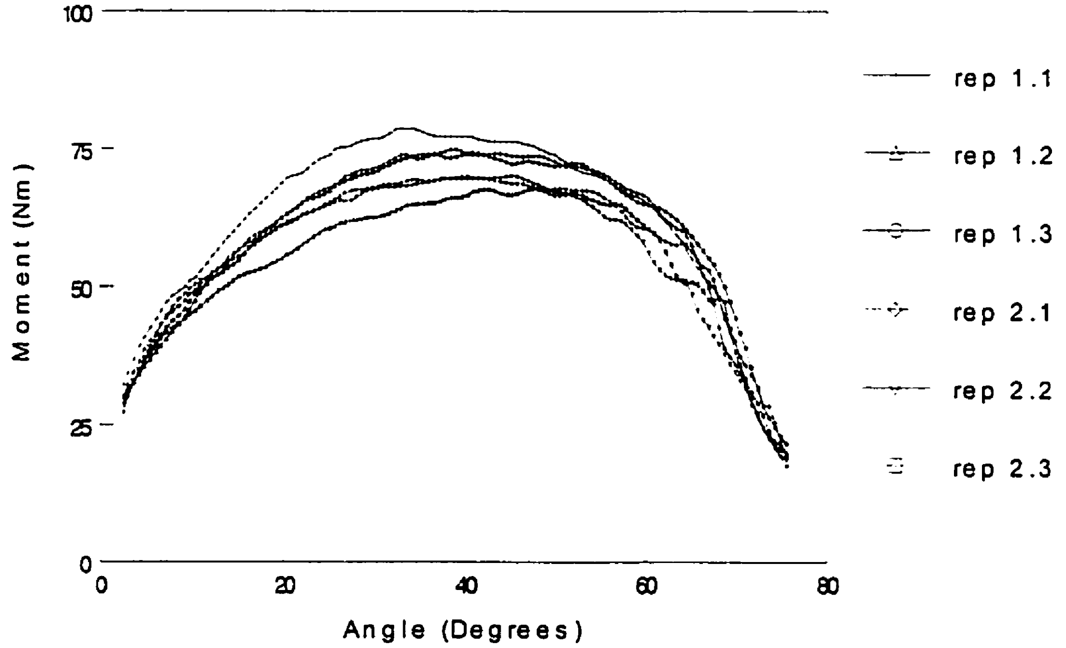


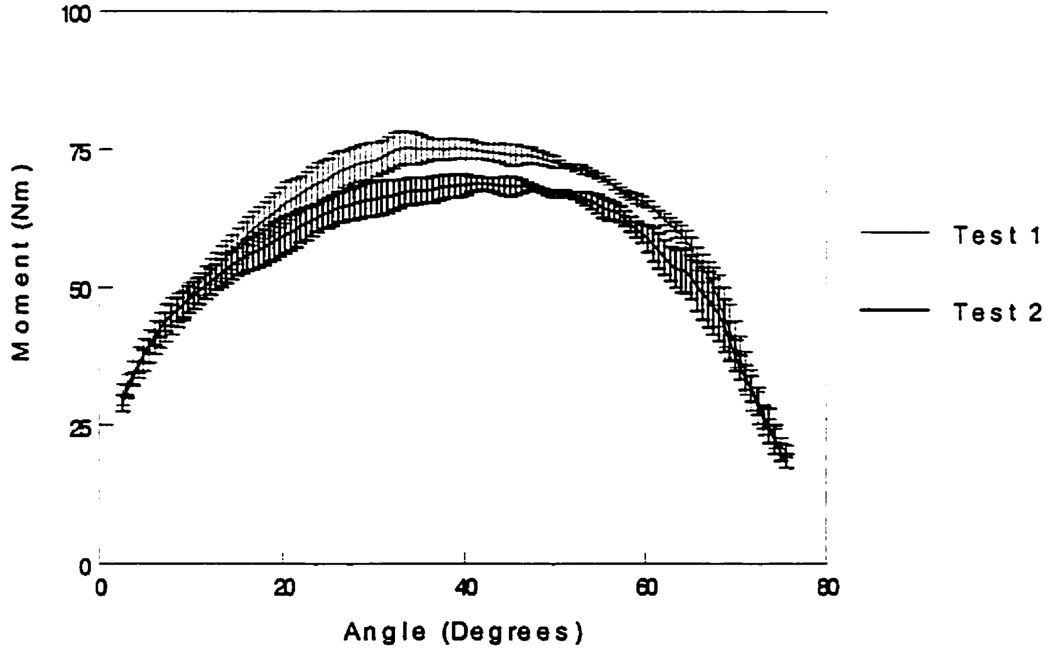
FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P4 – RAW DATA



MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P4 – STANDARD DEVIATION

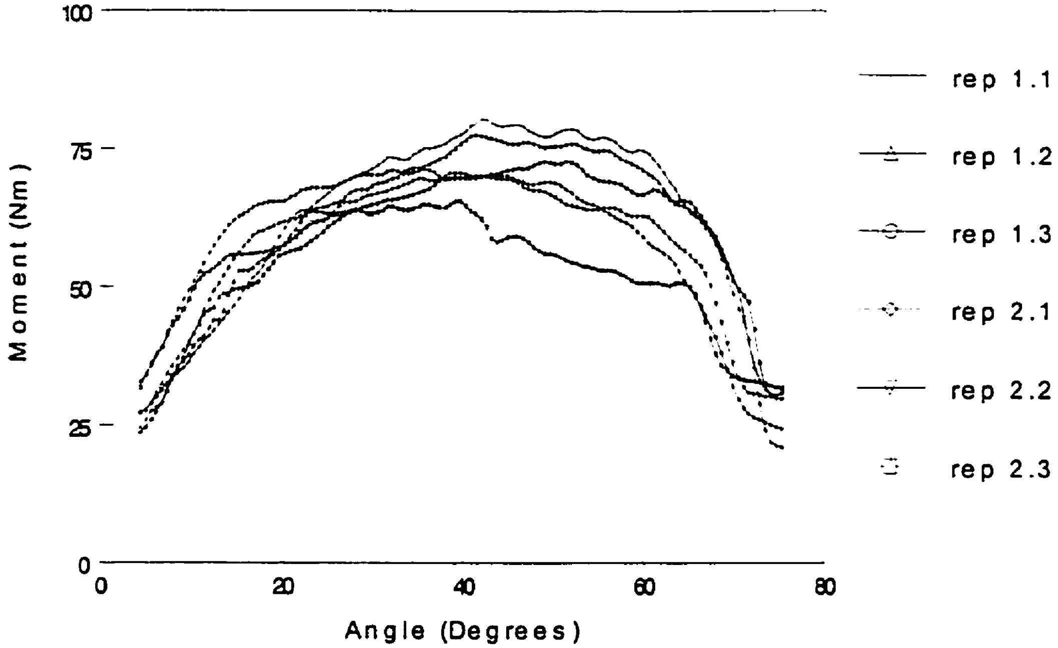
Appendix E – Moment / Angle Graphs for Individual Participants


FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P5 – RAW DATA

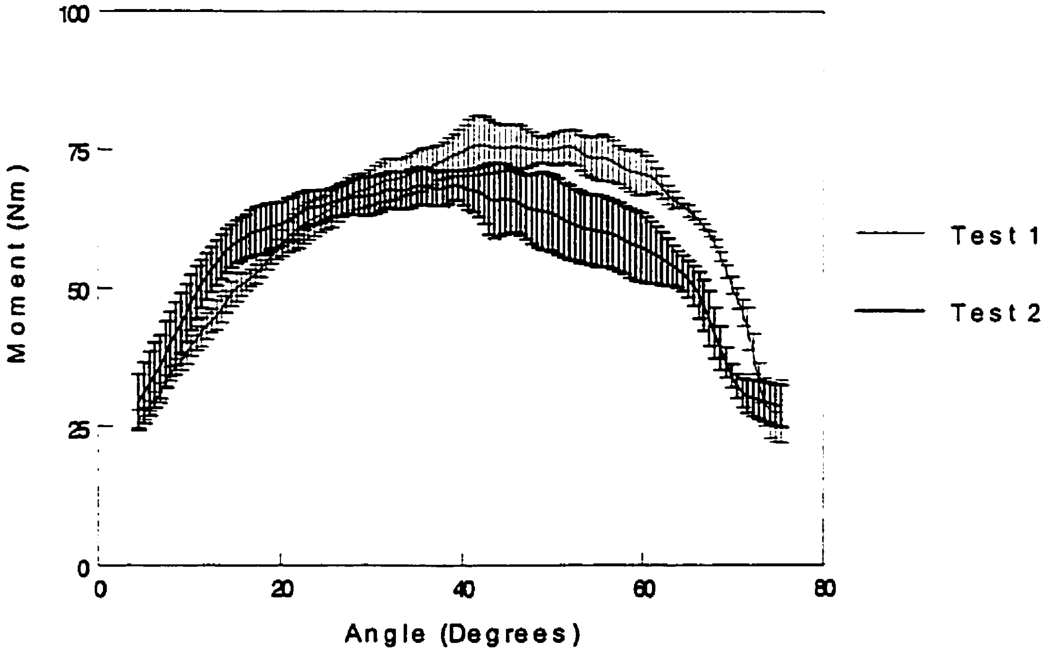

MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P5 – STANDARD DEVIATION

Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P6 – RAW DATA

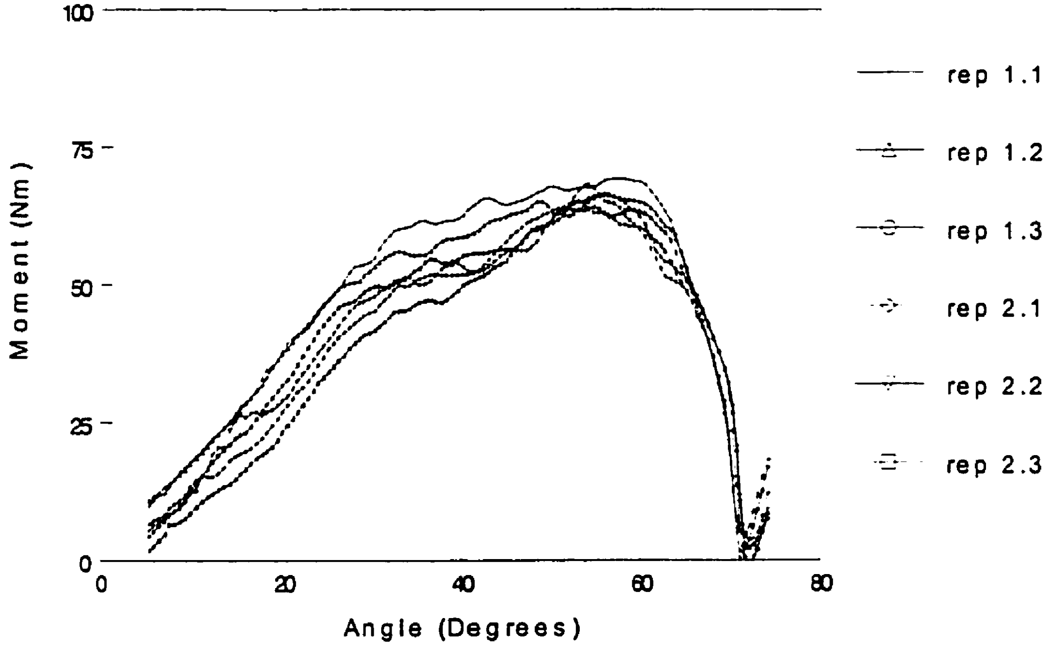


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P6 – STANDARD DEVIATION

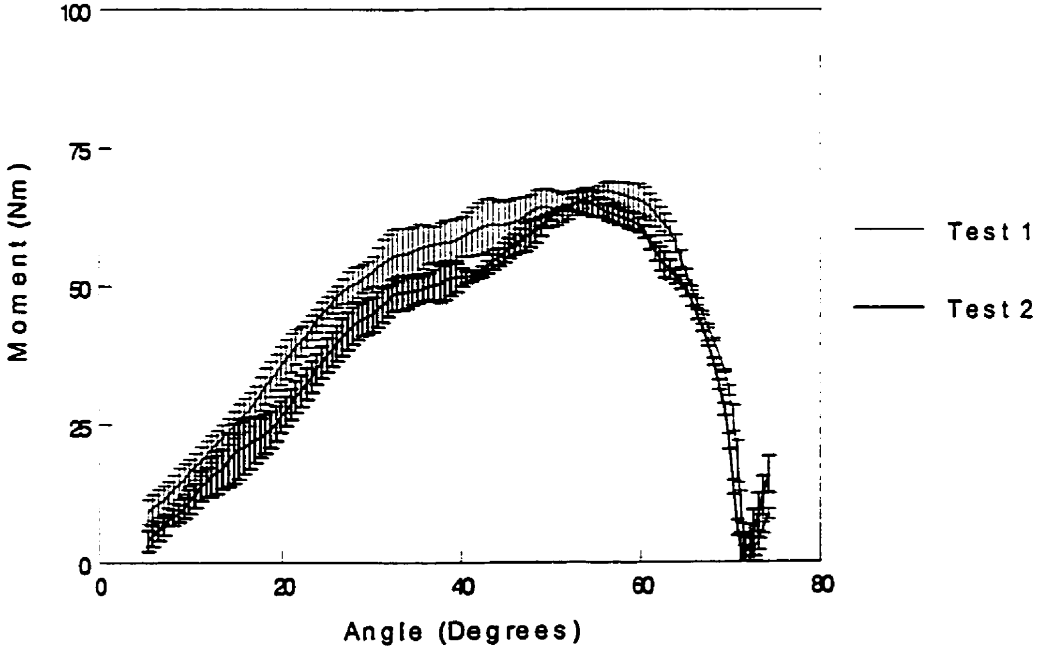


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P7 – RAW DATA

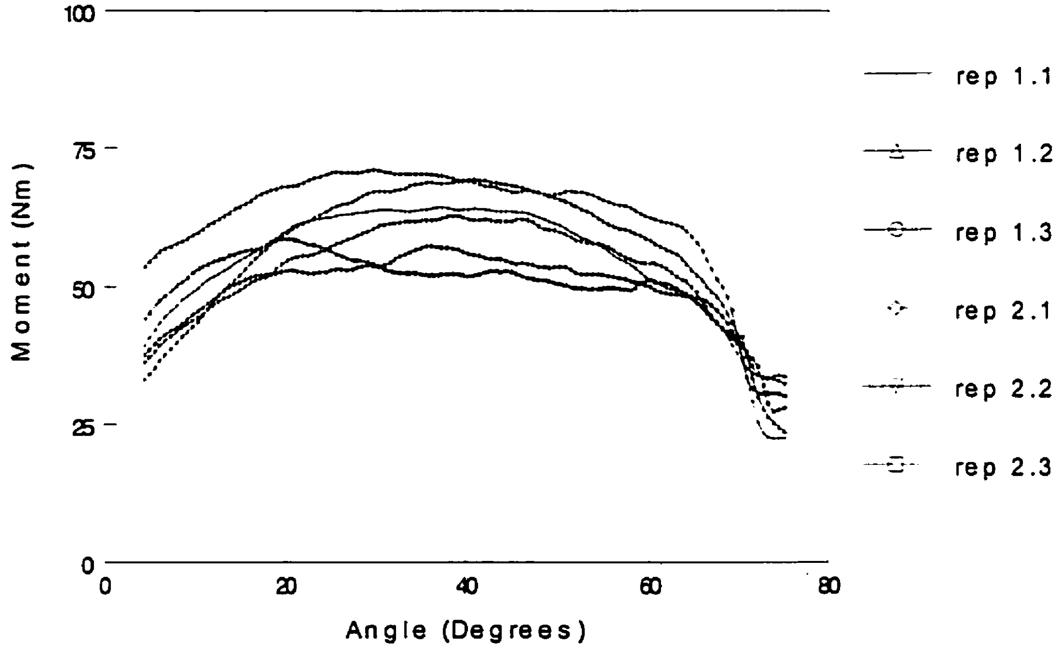


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P7 – STANDARD DEVIATION

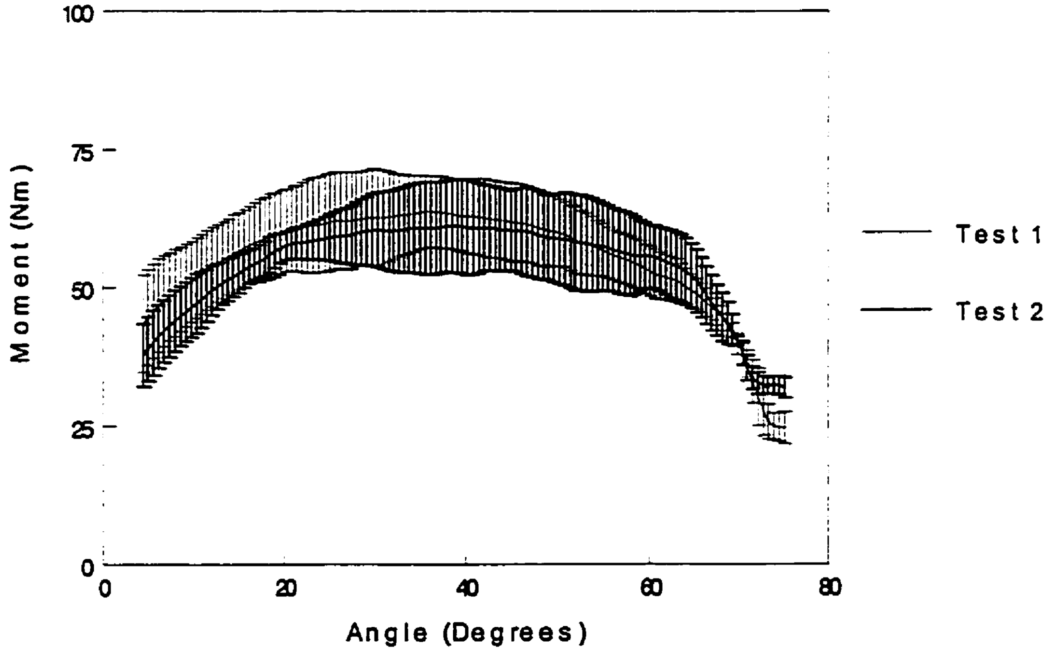


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P8 – RAW DATA

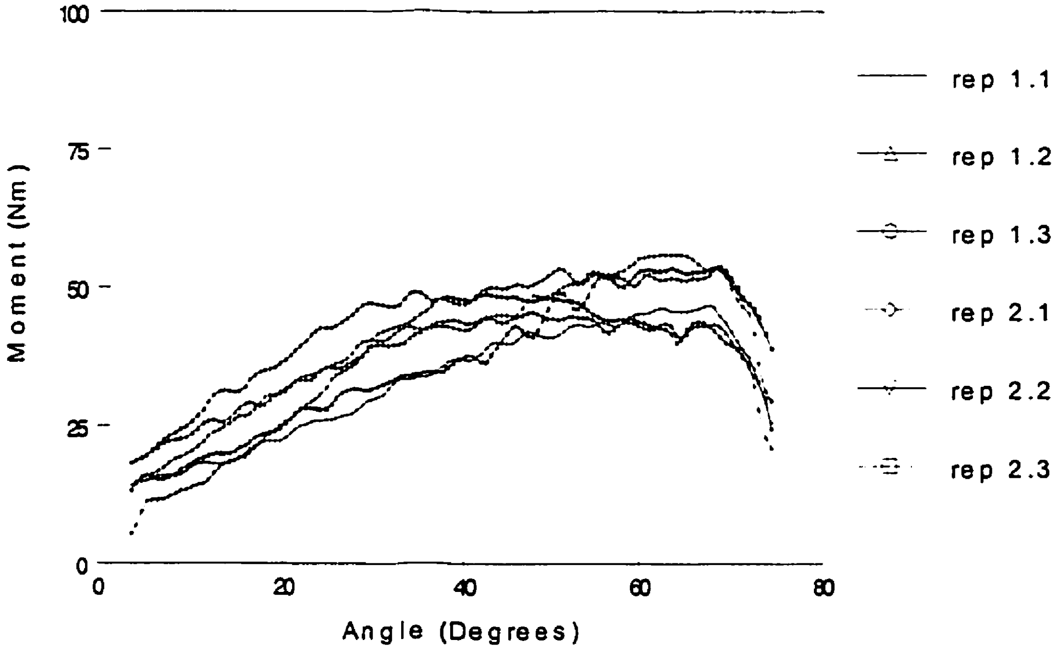


MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P8 – STANDARD DEVIATION

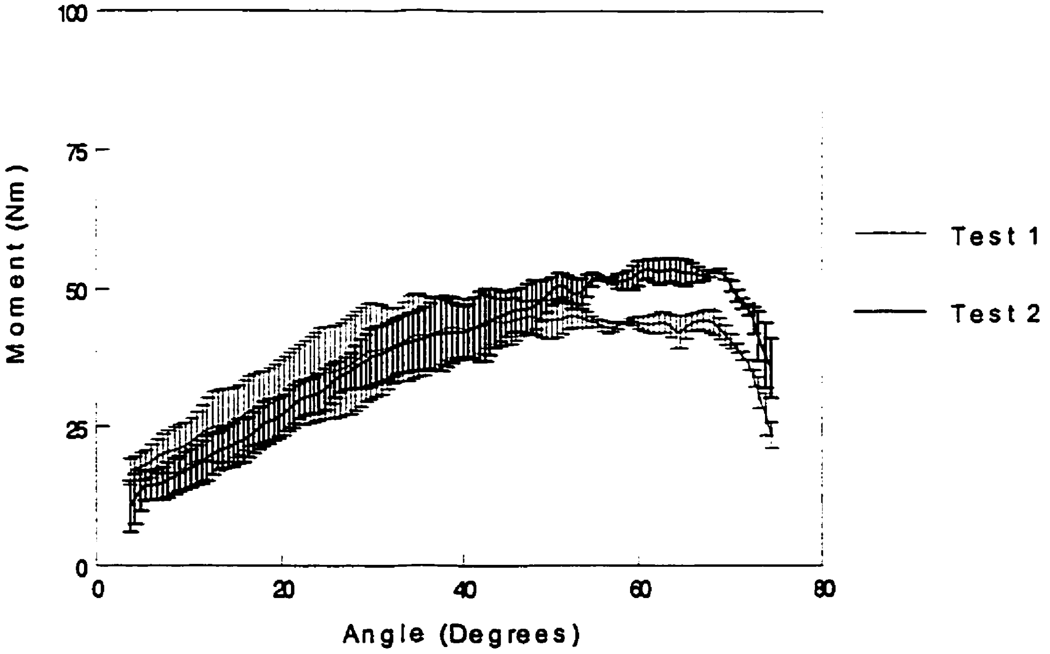


Appendix E – Moment / Angle Graphs for Individual Participants

FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
PARAPLEGIC PARTICIPANT P9 – RAW DATA



MEAN FORCE GENERATED DURING GLENOHUMERAL ADDUCTION
P9 – STANDARD DEVIATION



Appendix F - Area Under the Curve Values for All Participants

<table>
<tr><td colspan="5">Hypothesis
No change in isovelocity moment generated when test 1 is compared to test 2 in each test movement</td></tr>
<tr><th>Elevation</th><th>Test 1 AUC value</th><th>Test 2 AUC value</th><th>Linear Regression value</th><th>ICC Value</th></tr>
<tr><td>Quadriplegic Participants</td><td>1822.05</td><td>1603.68</td><td>.76</td><td>.903</td></tr>
<tr><td></td><td>1012.33</td><td>1813.17</td><td></td><td></td></tr>
<tr><td></td><td>1660.96</td><td>1308.29</td><td></td><td></td></tr>
<tr><td></td><td>3046.60</td><td>3694.46</td><td></td><td></td></tr>
<tr><td></td><td>1498.25</td><td>1177.36</td><td></td><td></td></tr>
<tr><td></td><td>2769.89</td><td>3840.01</td><td></td><td></td></tr>
<tr><td></td><td>3007.05</td><td>3088.63</td><td></td><td></td></tr>
<tr><td></td><td>1818.58</td><td>1859.95</td><td></td><td></td></tr>
<tr><td></td><td>2048.16</td><td>2064.42</td><td></td><td></td></tr>
<tr><td></td><td>1974.24</td><td>2343.22</td><td></td><td></td></tr>
<tr><td>Paraplegic Participants</td><td>4114.55</td><td>5478.78</td><td>.94</td><td>.984</td></tr>
<tr><td></td><td>5634.05</td><td>6037.81</td><td></td><td></td></tr>
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Appendix F – Area Under the Curve Values for All Participants

<table>
<tr><td colspan="5">Hypothesis
No change in isovelocity moment generated when test 1 is compared to test 2 in each test movement</td></tr>
<tr><th>Depression</th><th>Test 1 AUC value</th><th>Test 2 AUC value</th><th>Linear Regression value</th><th>ICC Value</th></tr>
<tr><td>Quadriplegic Participants</td><td>2115.56</td><td>1987.16</td><td>.98</td><td>.994</td></tr>
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<tr><td></td><td>2029.67</td><td>2127.04</td><td></td><td></td></tr>
<tr><td>Paraplegic Participants</td><td>2030.23</td><td>2170.35</td><td>.94</td><td>.981</td></tr>
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</table>

Appendix F – Area Under the Curve Values for All Participants

Hypothesis No change in isovelocity moment generated when test 1 is compared to test 2 in each test movement				
Flexion	Test 1 AUC value	Test 2 AUC value	Linear Regression value	ICC Value
Quadriplegic Participants	4115.46	4424.08	.95	.986
	4401.25	4737.14		
	3052.95	2277.91		
	5804.58	5910.77		
	5340.74	5003.77		
	3126.21	2955.28		
	6036.01	5882.65		
	4985.40	5081.22		
	1862.55	1852.17		
	3025.99	3137.07		
Paraplegic Participants	6012.99	6573.28	.83	.952
	9597.00	9267.09		
	7033.83	6802.12		
	6861.39	5783.45		
	7507.09	7182.94		
	8295.59	6582.40		
	7778.53	7692.01		
	4839.46	4300.59		
	4131.71	4525.00		

Appendix F – Area Under the Curve Values for All Participants

Hypothesis No change in isovelocity moment generated when test 1 is compared to test 2 in each test movement				
Adduction	Test 1 AUC value	Test 2 AUC value	Linear Regression value	ICC Value
Quadriplegic Participants	2593.92	2389.44	.98	.996
	4140.50	4215.31		
	1336.23	1592.51		
	4554.84	4306.08		
	4029.14	3981.02		
	6138.88	6122.53		
	3726.04	4007.32		
	2714.64	2622.18		
	2436.30	2296.97		
	4931.63	4844.88		
Paraplegic Participants	8840.55	9381.04	.86	.960
	7322.28	7261.74		
	8012.46	7012.09		
	6383.94	5253.91		
	7430.75	6879.58		
	7413.55	6897.45		
	5465.17	4801.81		
	6711.17	6553.67		
	4476.76	4678.02		

Appendix G - Statistical Results with ICC Confidence Intervals

Hypothesis No difference between test-retest coefficients between groups	Linear regression R-Squared Value	ICC	Confidence interval	
			Upper	Lower
Elevation -				
Quadriplegic Participants	.76	.903	.976	.606
Paraplegic Participants	.94	.984	.996	.928
Depression -				
Quadriplegic Participants	.98	.994	.999	.977
Paraplegic Participants	.94	.981	.996	.913
Flexion -				
Quadriplegic Participants	.95	.986	.997	.945
Paraplegic Participants	.83	.952	.989	.785
Adduction -				
Quadriplegic Participants	.98	.996	.999	.983
Paraplegic Participants	.86	.960	.991	.822

Appendix G – Statistical Results with ICC Confidence Intervals

Hypothesis No difference between test-retest AUC values between group with active sternal head (ASH) of pectoralis major, and group with non-active sternal head (NASH) of pectoralis major	Linear regression R-Squared Value	ICC	Confidence interval	
			Upper	Lower
Elevation -				
ASH Participants	.82	.893	.988	.295
NASH Participants	.94	.984	.996	.928
Depression -				
ASH Participants	.99	.992	.999	.927
NASH Participants	.97	.956	.995	.646
Flexion -				
ASH Participants	.99	.993	.999	.966
NASH Participants	.92	.946	.994	.732
Adduction -				
ASH Participants	.99	.996	.999	.962
NASH Participants	.96	.979	.998	.814

Appendix G – Statistical Results with ICC Confidence Intervals

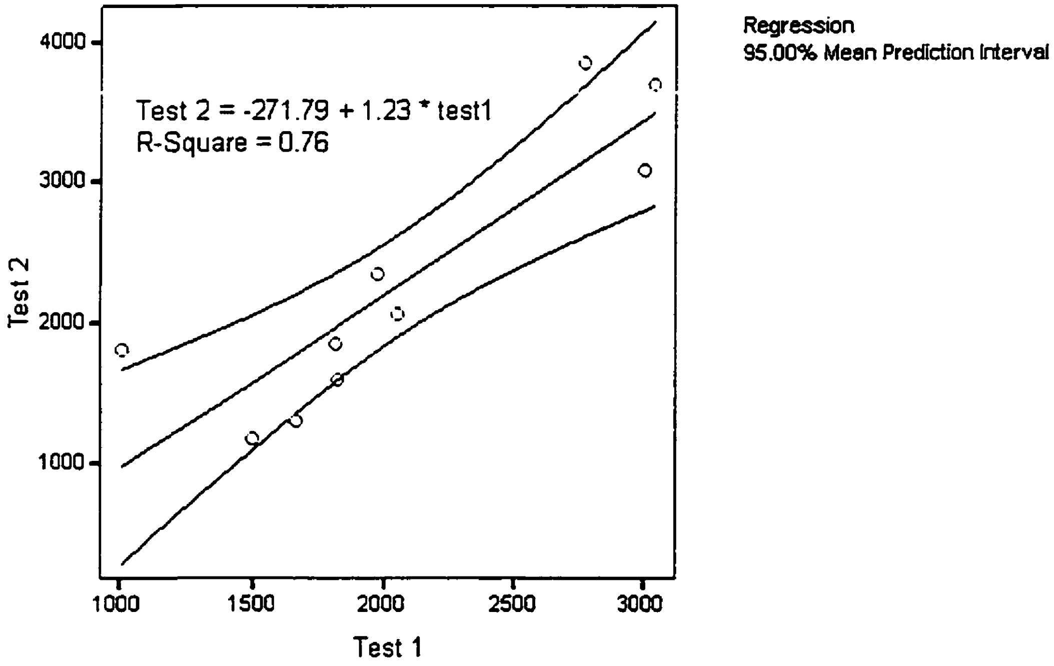
Hypothesis No difference between test-retest AUC values between group with greater than two days (2+) of recreational activity in one week, and group with 2 or fewer days (2-) of recreational activity in one week	Linear regression R-Squared Value	ICC	Confidence interval	
			Upper	Lower
Elevation -				
2+ Participants	1.00	.997	.999	.971
2- Participants	.95	.977	.993	.930
Depression -				
2+ Participants	.98	.992	.999	.923
2- Participants	.96	.982	.994	.946
Flexion -				
2+ Participants	.92	.959	.996	.664
2- Participants	.94	.966	.989	.897
Adduction -				
2+ Participants	.98	.975	.997	.779
2- Participants	.96	.980	.994	.939

Appendix H - Statistical test for age difference

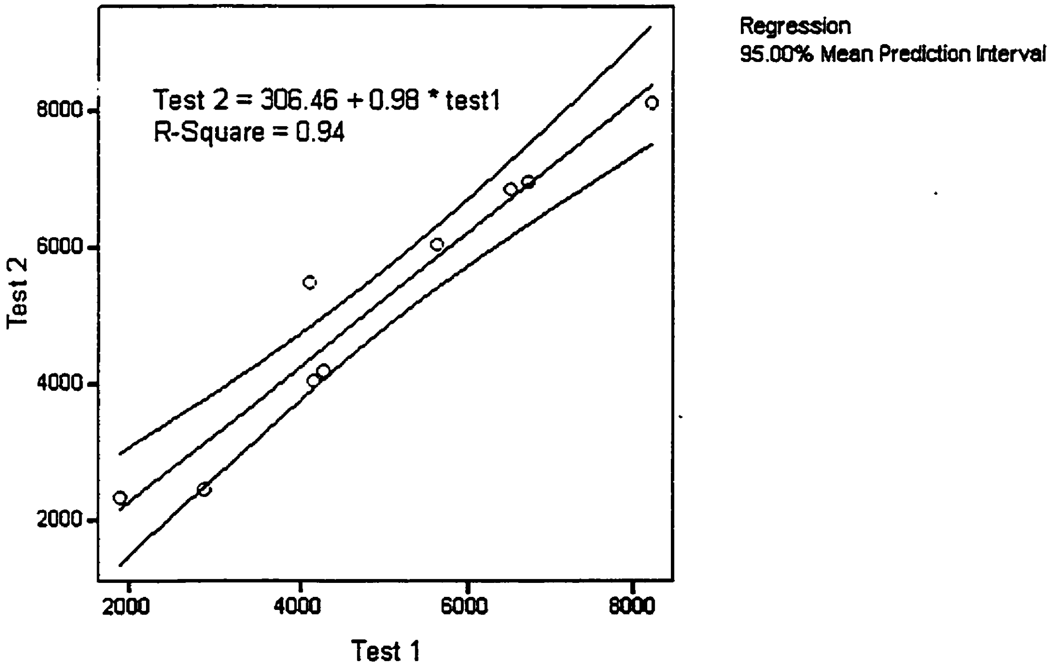
Participant Number	Participants with Quadriplegia (Group Q)	Participants with Paraplegia (Group P)	Mean and SD Group (Q)	Mean and SD Group (P)
1	40	31	33.2 ± 8.4	38.1 ± 8.8*
2	30	43		
3	25	31		
4	24	52		
5	24	21		
6	41	42		
7	46	41		
8	26	35		
9	41	45		
10	35	40		

*$p \leq 0.01$

Appendix I - Linear Regression Graphs

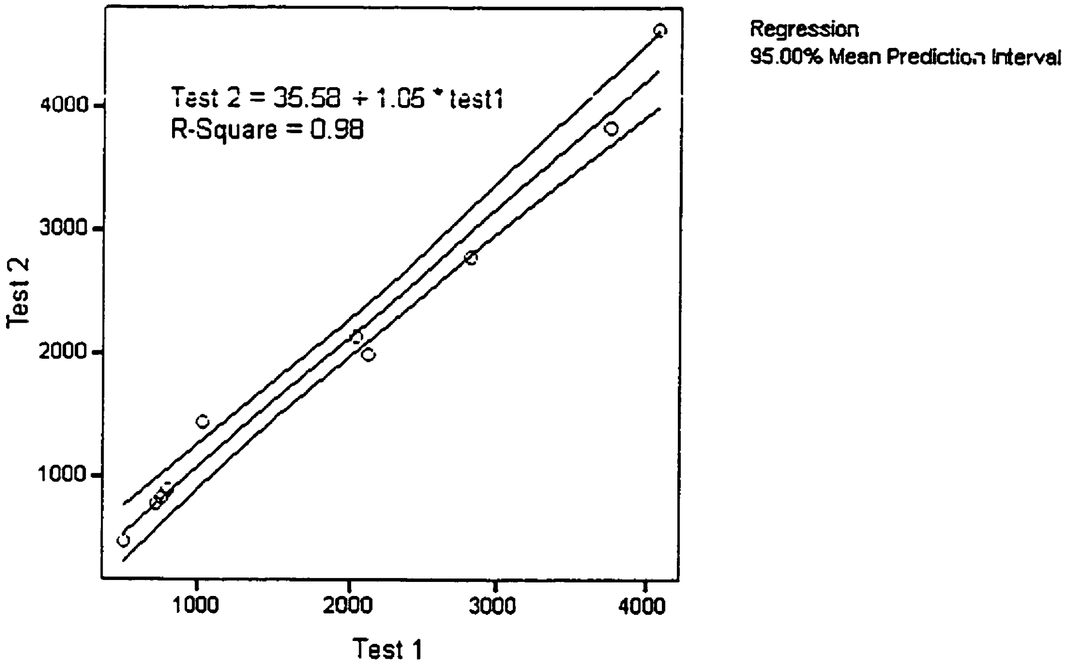


Linear Regression - Elevation
Quadriplegic Participants



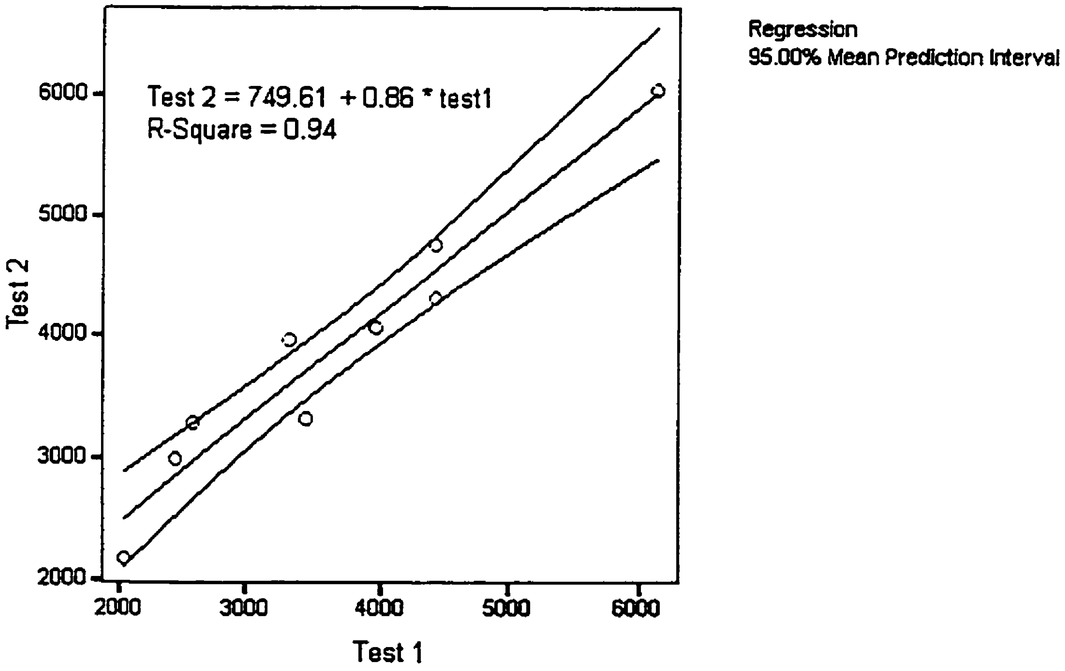
Linear Regression - Elevation
Paraplegic Participants

Appendix I (Continued) – Linear Regression Graphs



Linear Regression - Depression

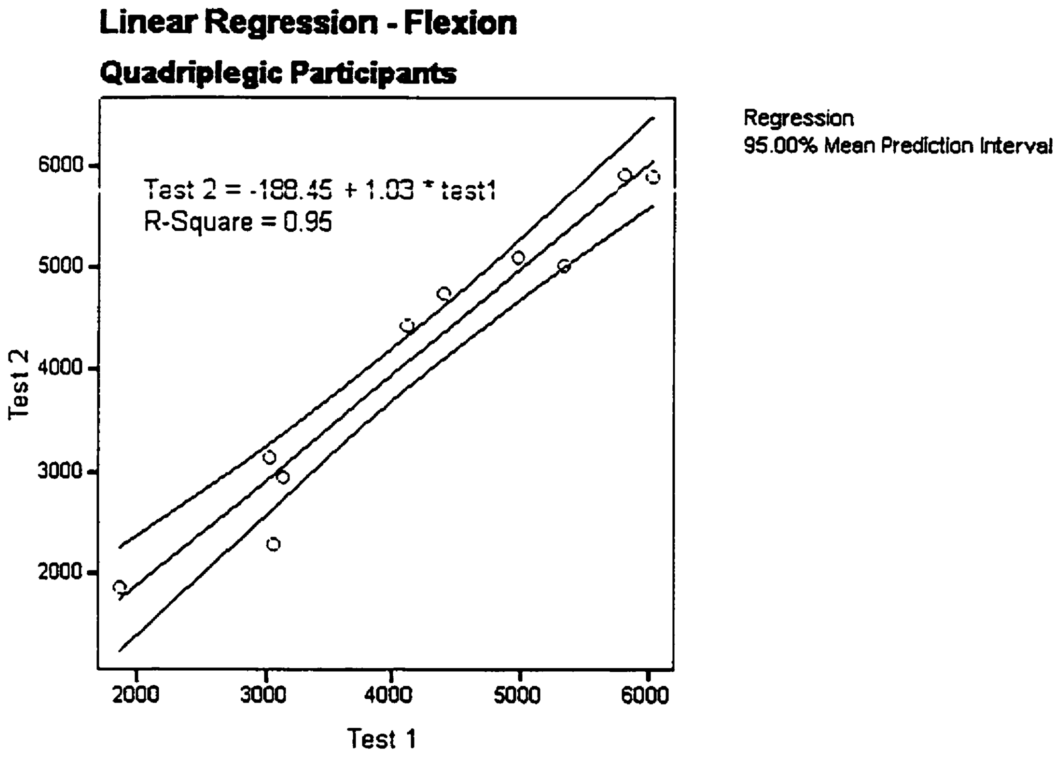
Quadriplegic Participants

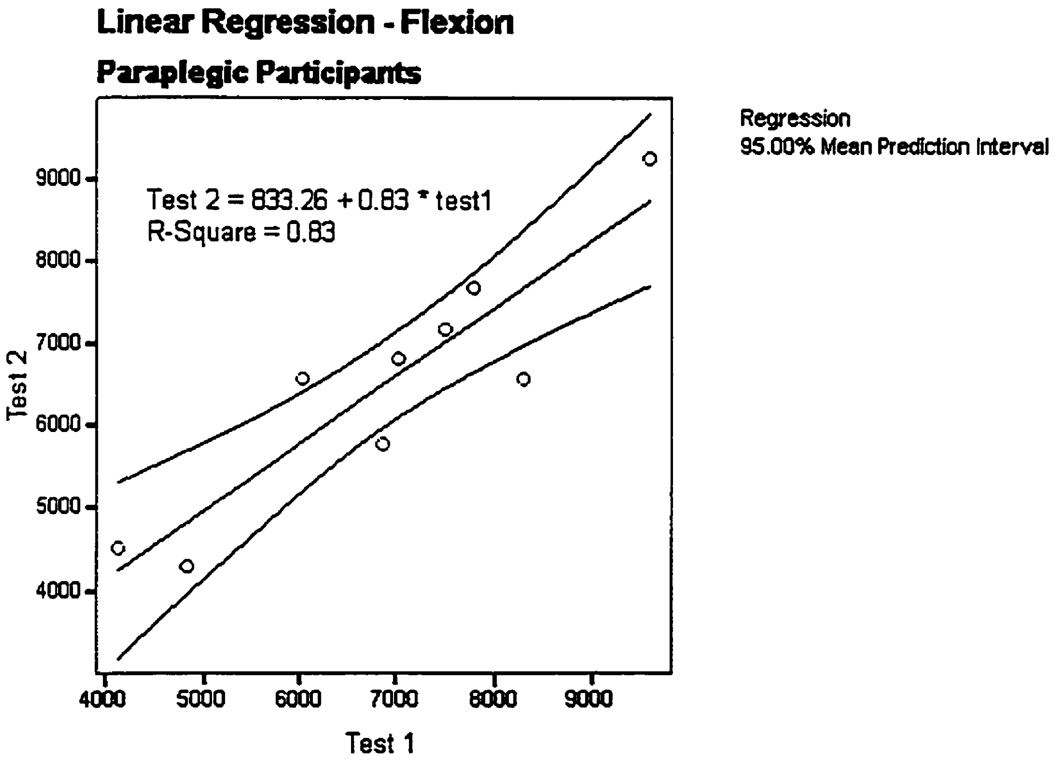


Linear Regression - Depression

Paraplegic Participants

Appendix I (Continued) – Linear Regression Graphs

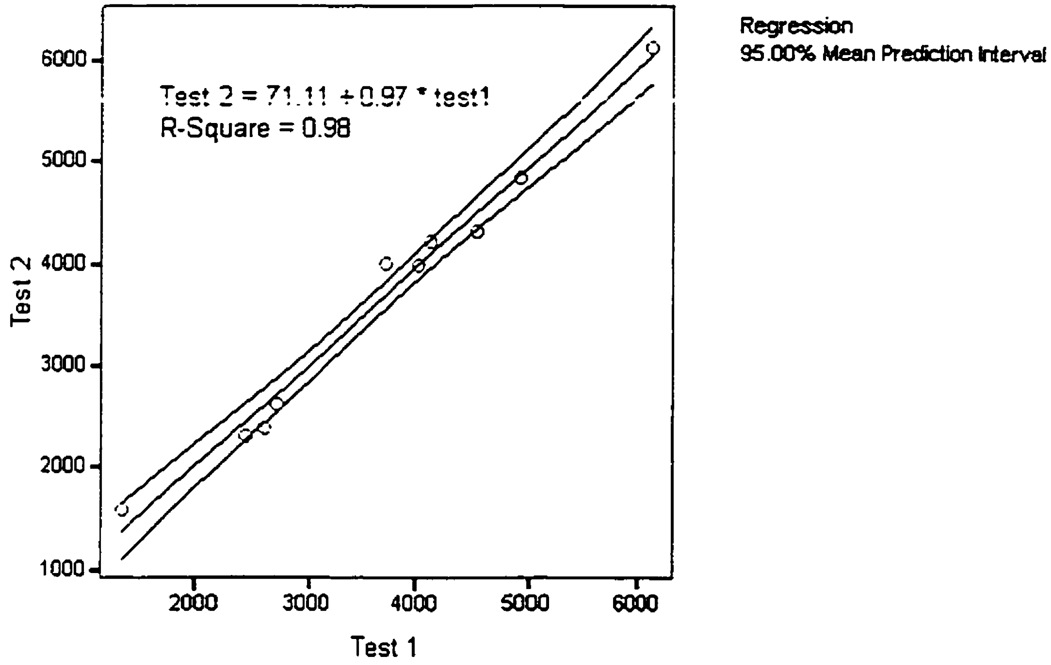




Appendix I (Continued) – Linear Regression Graphs

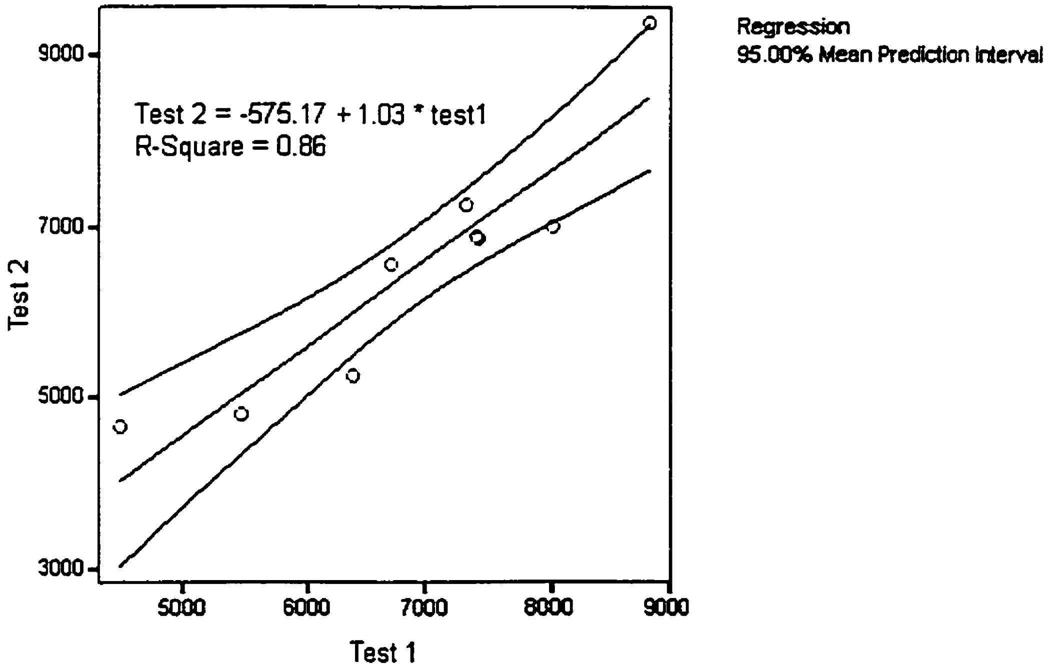
Linear Regression - Adduction

Quadriplegic Participants



Linear Regression - Adduction

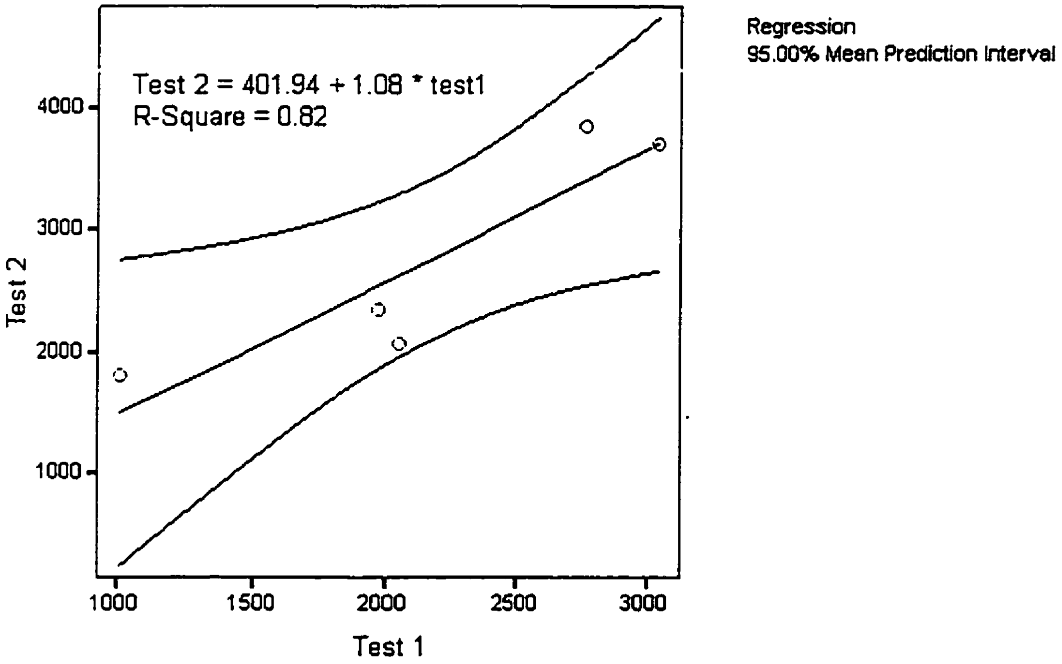
Paraplegic Participants



Appendix I (Continued) – Linear Regression Graphs

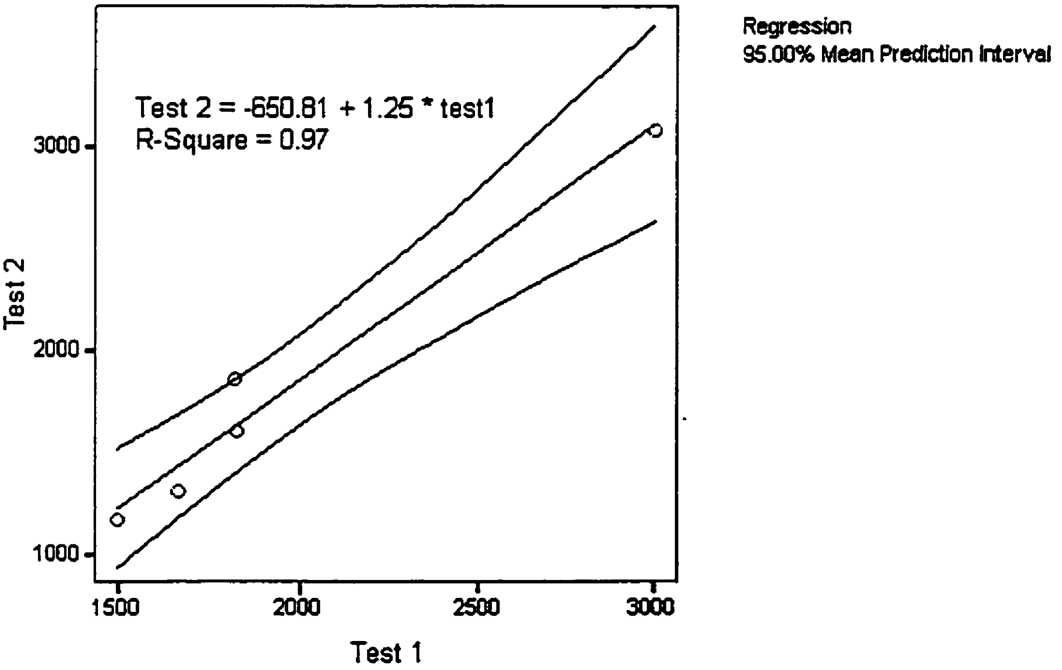
Linear Regression - Elevation

Quadriplegic Participants
Active Sternal Head of Pectoralis Major



Linear Regression - Elevation

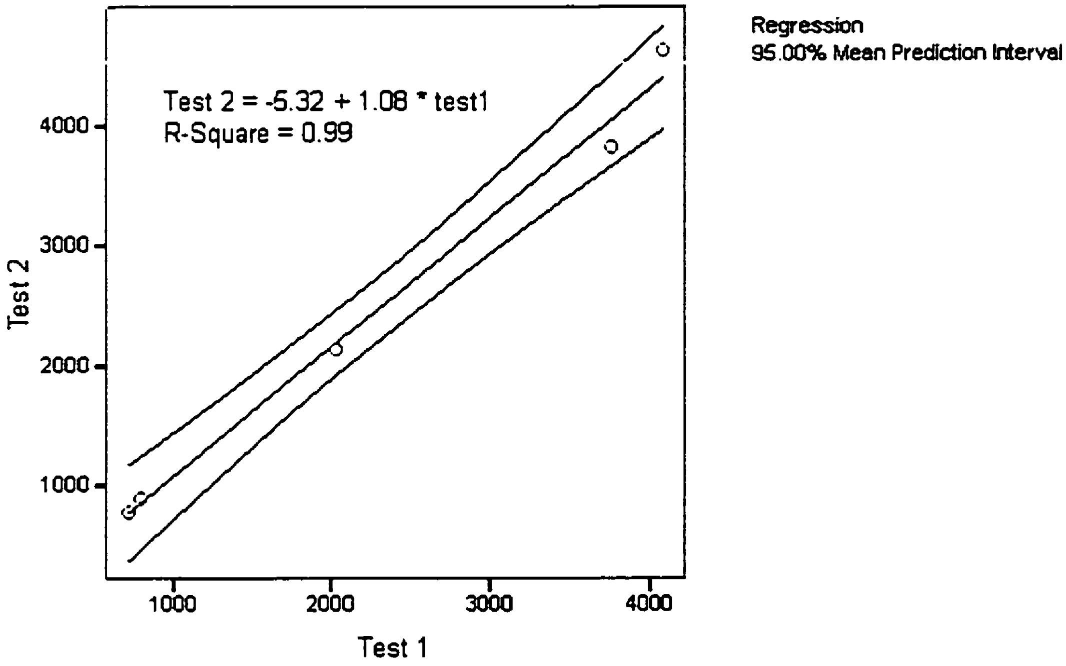
Quadriplegic Participants
Non-active Sternal Head of Pectoralis Major



Appendix I (Continued) – Linear Regression Graphs

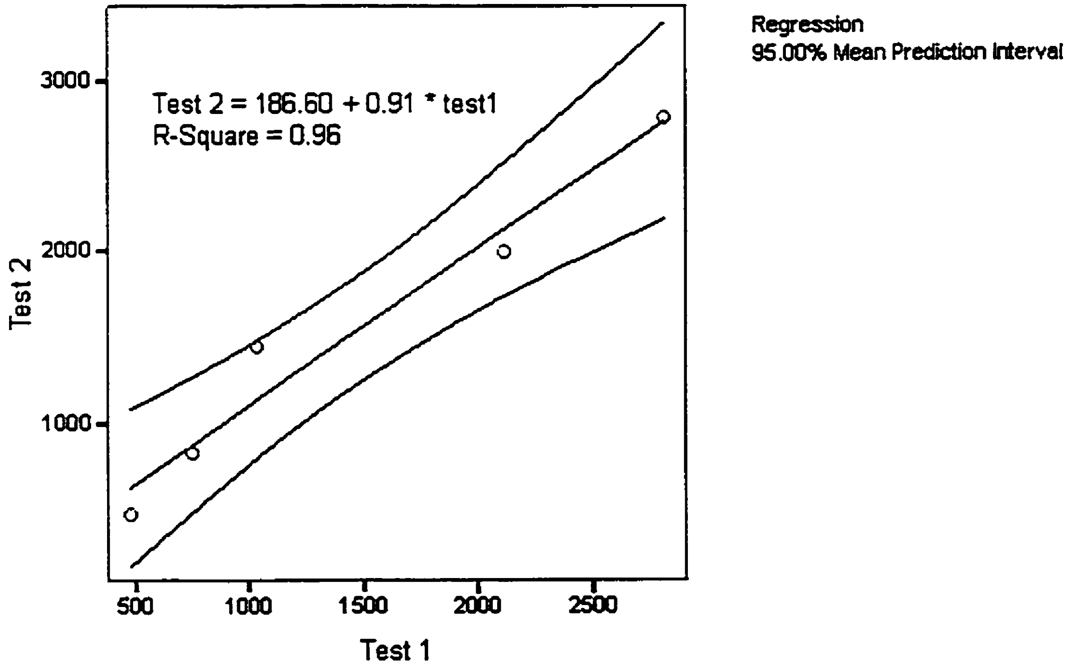
Linear Regression - Depression

Quadriplegic Participants
Active Sternal Head of Pectoralis Major



Linear Regression - Depression

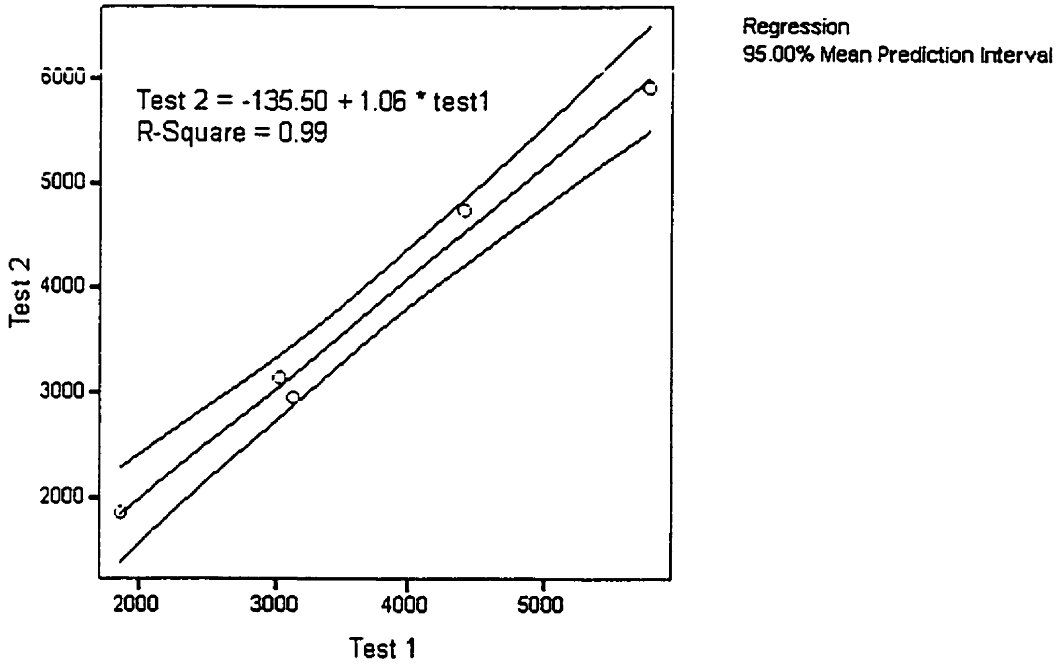
Quadriplegic Participants
Non-active Sternal Head of Pectoralis Major



Appendix I (Continued) – Linear Regression Graphs

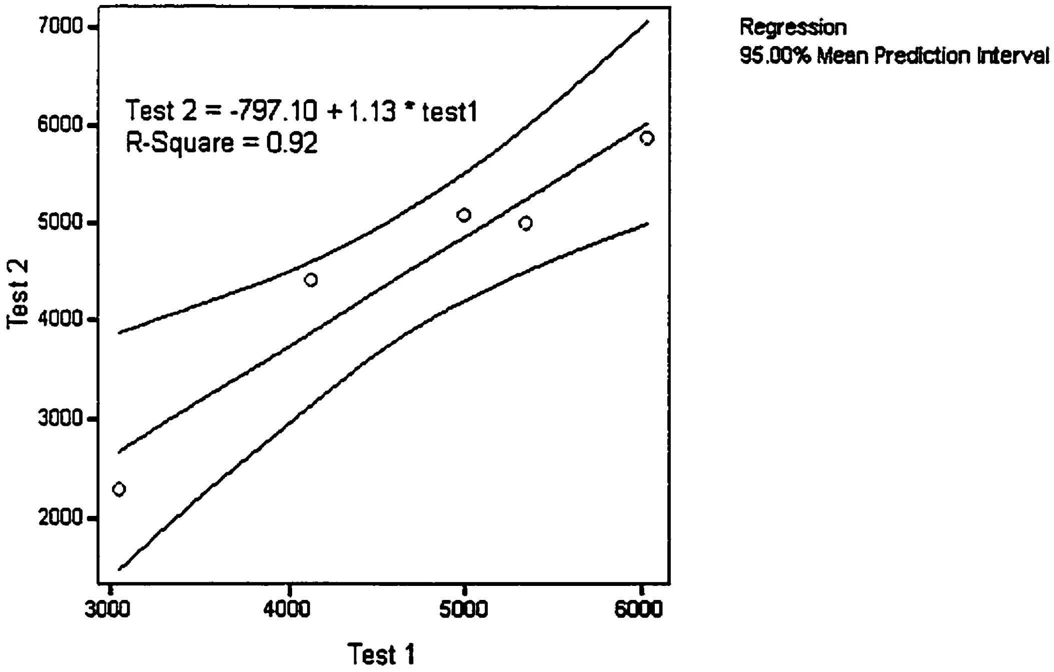
Linear Regression - Flexion

Quadriplegic Participants
Active Sternal Head of Pectoralis Major



Linear Regression - Flexion

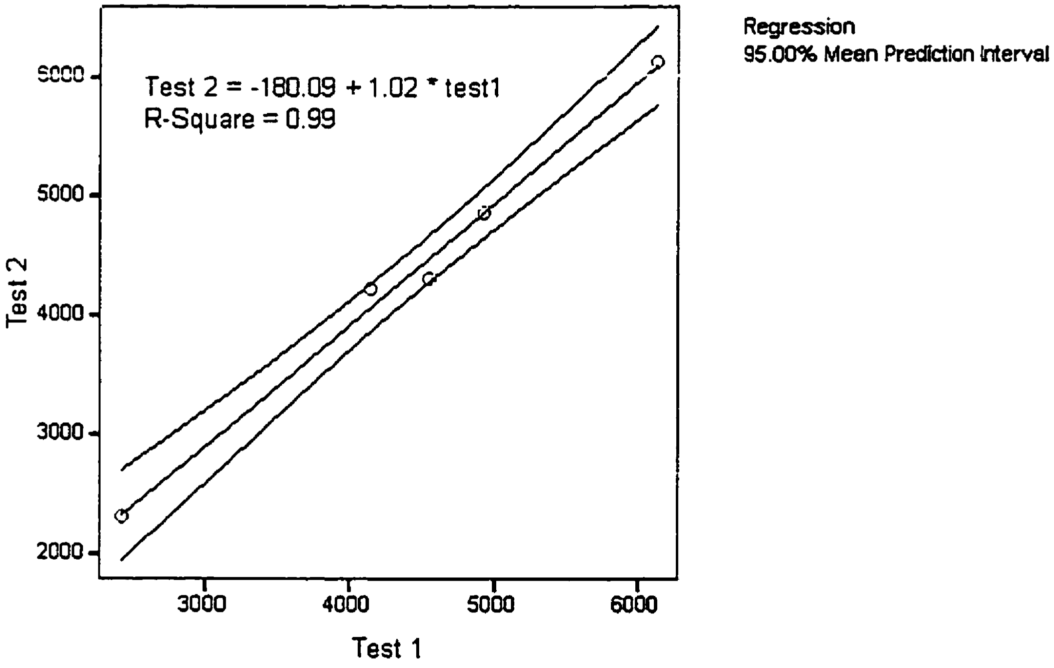
Quadriplegic Participants
Non-active Sternal Head of Pectoralis Major



Appendix I (Continued) – Linear Regression Graphs

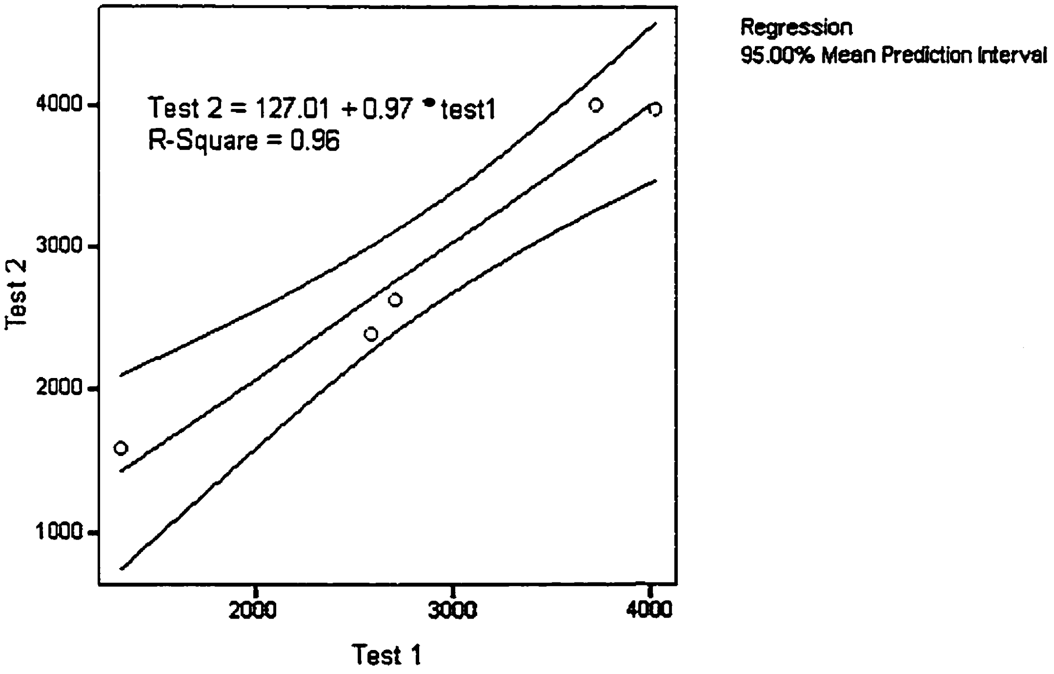
Linear Regression - Adduction

Quadriplegic Participants
Active Sternal Head of Pectoralis Major



Linear Regression - Adduction

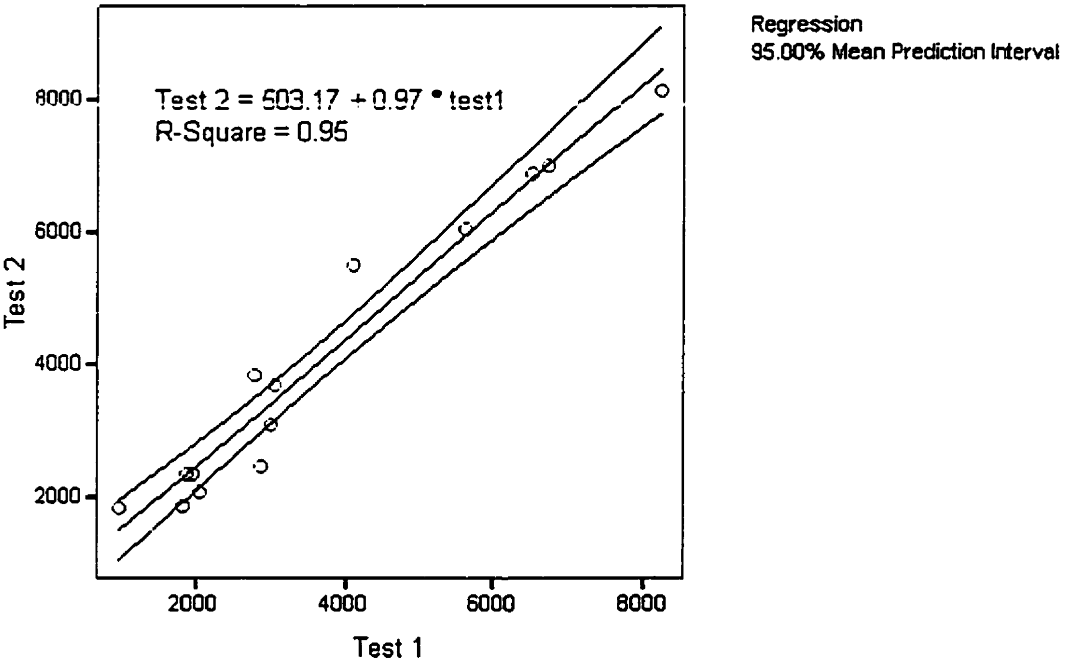
Quadriplegic Participants
Non-active Sternal Head of Pectoralis Major



Appendix I (Continued) – Linear Regression Graphs

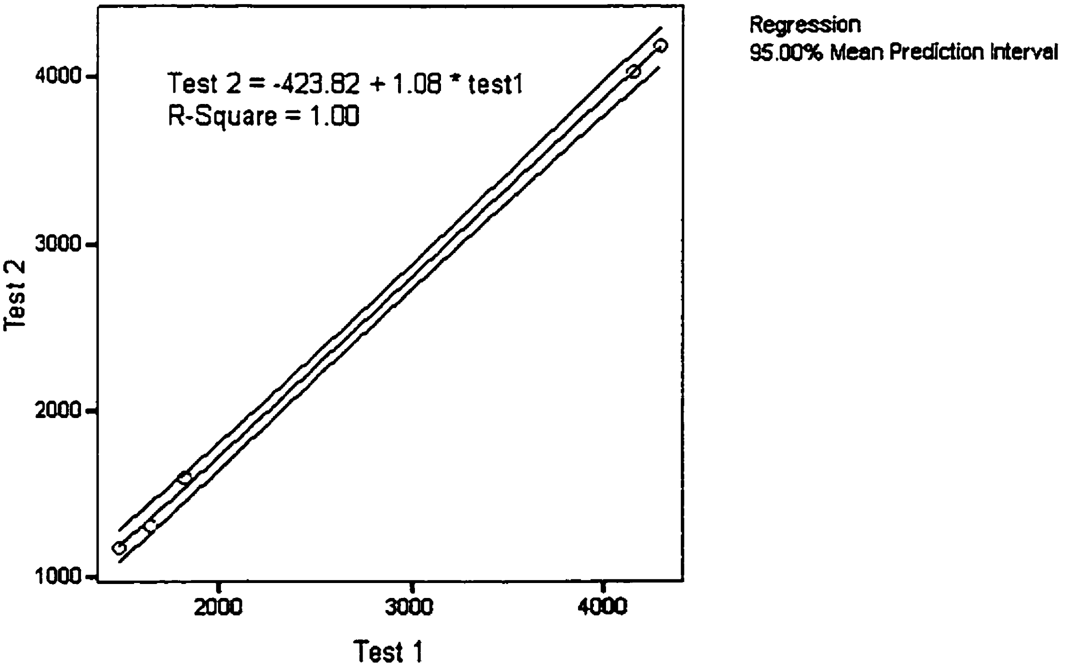
Linear Regression - Elevation

2- Recreational Group



Linear Regression - Elevation

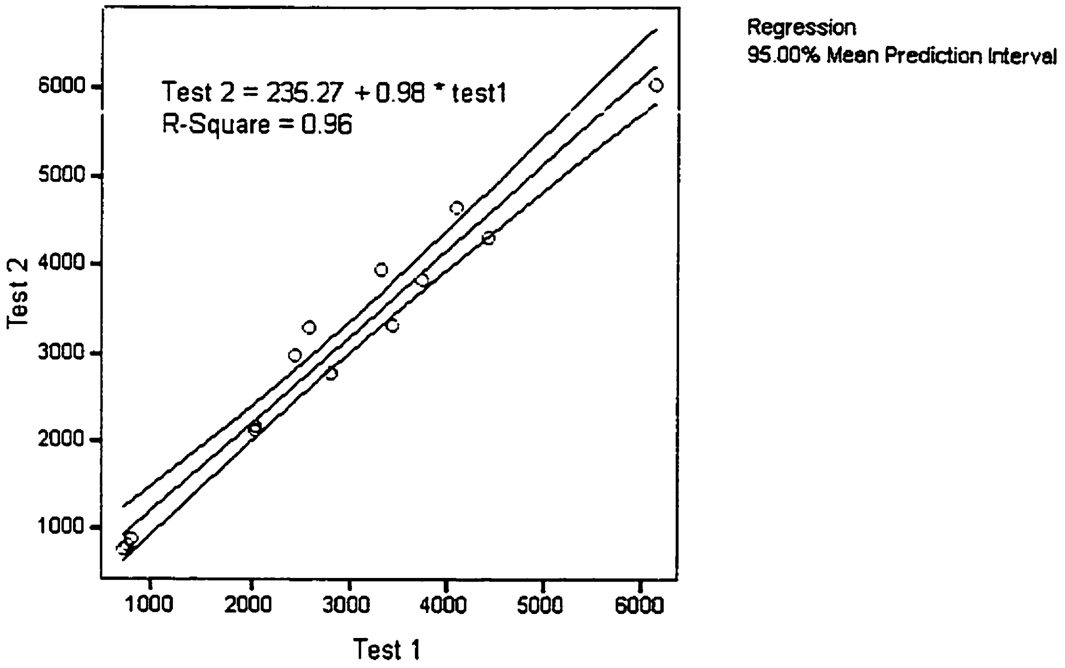
2+ Recreational Group



Appendix I (Continued) – Linear Regression Graphs

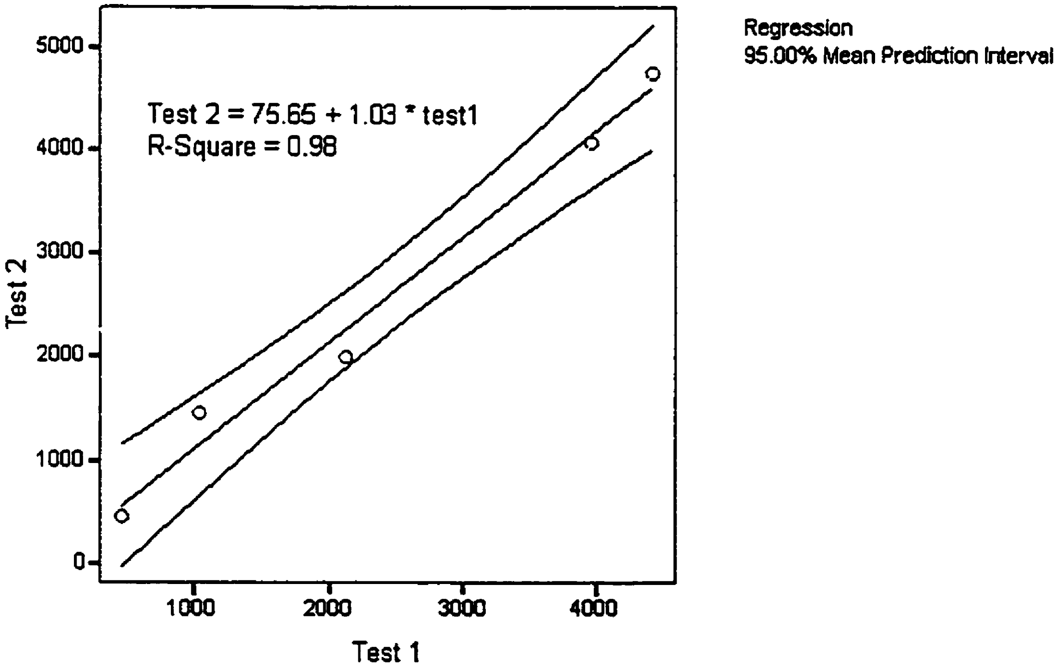
Linear Regression - Depression

2- Recreational Group



Linear Regression - Depression

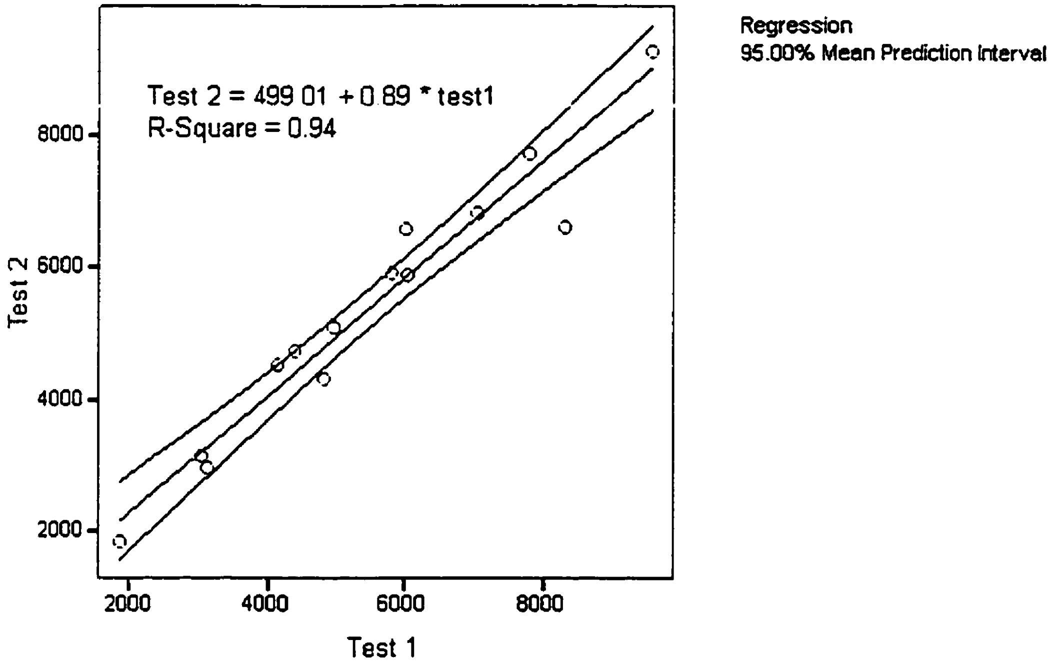
2+ Recreational Group



Appendix I (Continued) – Linear Regression Graphs

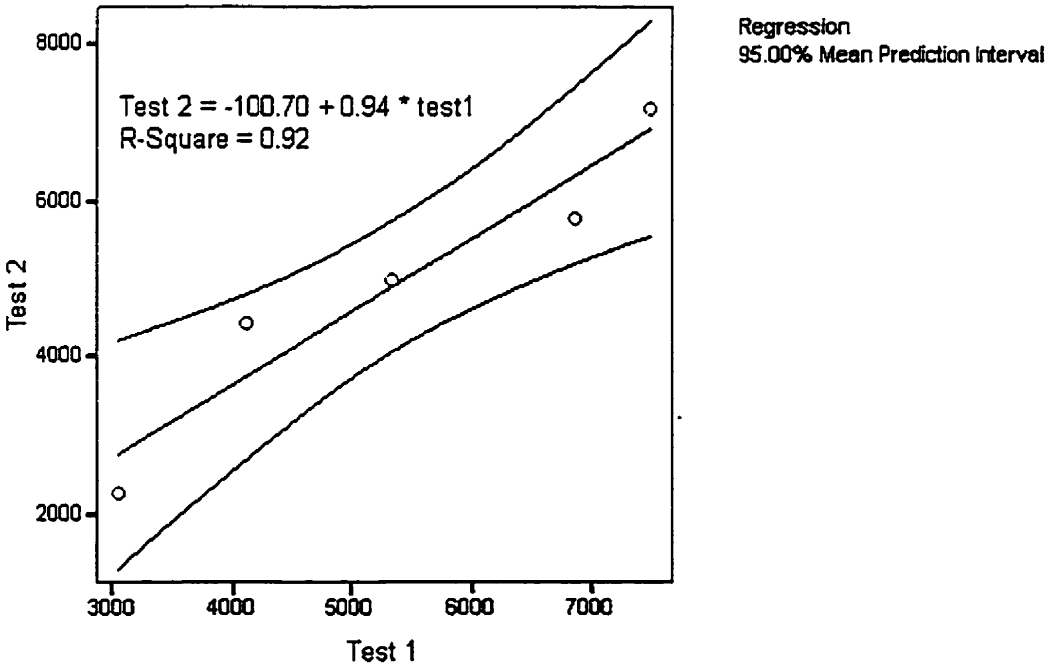
Linear Regression - Flexion

2- Recreational Group



Linear Regression - Flexion

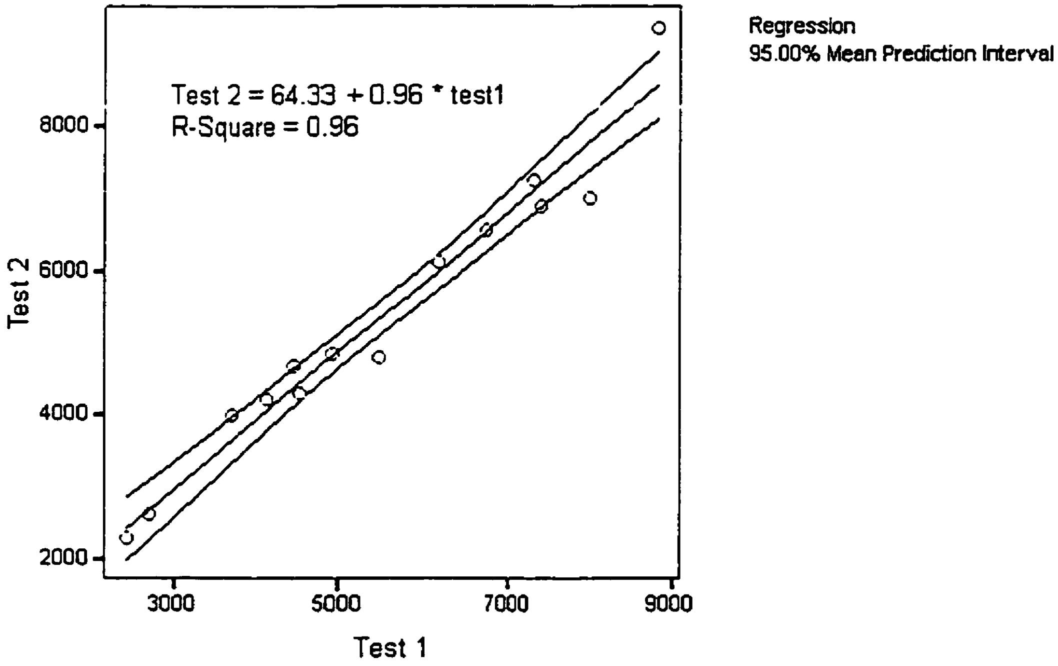
2+ Recreational Group



Appendix I (Continued) – Linear Regression Graphs

Linear Regression - Adduction

2- Recreational Group



Linear Regression - Adduction

2+ Recreational Group