Differential Fatigability of the Glenohumeral Rotators

By

Maureen Michelle Walker

A thesis submitted to the
Faculty of Graduate Studies
in partial fulfilment of the requirements
of the degree of

MASTER OF SCIENCE
(Rehabilitation)

School of Medical Rehabilitation
University of Manitoba
April 2002

0-612-80074-1

THE UNIVERSITY OF MANITOBA

FACULTY OF GRADUATE STUDIES

*****

COPYRIGHT PERMISSION PAGE

Differential Fatigability of the Glenohumeral Rotators

BY

Maureen Michelle Walker

A Thesis/Practicum submitted to the Faculty of Graduate Studies of The University of Manitoba in partial fulfillment of the requirements of the degree

of

MASTER OF SCIENCE

Table of Contents

ABSTRACT

Purpose: To provide the first quantification of the magnitude of fatigue of the internal and external rotators while acting as stabilizers of the glenohumeral joint during a repetitive upper limb grip task. To provide the first reported occurrence of differential fatigue between stabilizer muscles groups and between prime mover and stabilizer muscle groups. **Method:** An isovelocity dynamometer (Kin-Com 500H) was used to test shoulder internal and external rotator strength in 15 healthy male subjects. Two maximal effort contractions were performed at 30° and 60° of shoulder external rotation. The identical protocol was repeated after subjects completed a repetitive grip task at 60% of their maximum grip strength with the arm positioned in one of two positions; at the side or elevated to the horizontal. Peak and average moment, as well as IR/ER ratios were derived for each subject using ISOMAP software (Isodyne Inc., Winnipeg). Changes in strength were determined by calculating the difference between the initial and final strength data, and expressed as a percentage values. Dependent T-tests were performed to determine statistical significance. Correlation analyses were performed on age, body mass, initial strength, completed repetitions, grip strength and strength deficits. **Results & Conclusions**: The results demonstrated significant differential fatigue of the internal and external rotator muscles following performance of a standardized gripping task in two distinct positions. The magnitude of fatigue of the internal and external rotators was dependant upon the position of the upper limb during the task and shoulder joint angle. A neuromuscular imbalance was demonstrated that arose from the differential fatigue of the shoulder stabilizers, which could result in increased humeral head migration and ultimately lead to shoulder pathology. Intervention must include an individualized exercise program based upon a thorough assessment of not only strength but also endurance of the internal and external rotator muscles.

ACKNOWLEDGEMENTS

Special thanks go first of all to my advisor Dr. Dean Kriellaars for his unparalleled commitment to my learning process, his amazing ability to provide insight and clarification on any detail, and his contagious enthusiasm for learning. I consider it a privilege to have been given the opportunity to work with him.

I would also like to thank my committee members, Dr. Brent Fedirchuk and Dr. Peter MacDonald for providing insightful comments on both the proposal and final documents, and for their willingness to sacrifice their valuable time.

Thank you to my colleagues at the human dynamics laboratory for providing a positive and encouraging environment. A special thank you to Lynda for her assistance in subject recruitment and her tireless ability to answer questions.

I would like to acknowledge all 15 subjects who willingly volunteered their time to participate in this study. My sincere thanks for their time and patience, it was a pleasure to meet them and an honour to work with them.

Special thanks to my mother and father who provided me with opportunities for quality education, encouraged me to pursue my dreams and instilled in me the confidence that I could succeed. Thank you to my mother who continues to be a constant source of support and encouragement.

Thank you to my wonderful supportive family. To my daughter Laura and my son Scott; your zest for life and boundless energy inspire me daily. Thank you for sharing your mommy with the "outside world". To my wonderful husband and best friend Rex; your love, patience, understanding and willingness to help ("above and beyond the call of duty") were integral to the completion of this thesis.

Finally, this thesis is dedicated to the memory of my father
Charles H. Walker (February 14, 1910 – December 21, 1996),
who taught me by example, about integrity, self-control and love.

LIST OF FIGURES

LIST OF TABLES

OPERATIONAL DEFINITIONS

Prime Mover - The muscle(s) primarily responsible for producing and controlling a movement.

Synergists - A group of muscles that are constrained to work as a unit.

Isometric contraction – A state of muscle activation where the moment generated by the muscle that is equal to the moment produced by the load, resulting in no change in whole muscle length.

Concentric contraction: A state of muscle activation where the moment generated by the muscle that exceeds the moment produced by the load, resulting in shortening of whole muscle length. The force developed by the muscle fibre is dependant both on the length of the fibre and the contraction velocity.

Eccentric contraction: A state of muscle activation where the moment generated by the external load exceeds the moment generated by the muscle, which results in the muscle lengthening.

Resultant joint moment (RJM) – The net rotational moment or moment exerted about a joint by all tissues forces spanning the joint, including muscles, ligaments and bones.

Strength – Strength is the ability to control rotation and translation of body segments and is directly related to the ability to produce a RJM. RJM requires the generation of a moment about the joint by the muscle. This involves both muscular (cross-sectional area, fibre length, pennation angle, muscle fibre type) and neural physiological properties (motor unit type), as well as the length of the moment arm (perpendicular distance between the line of force application and the axis of rotation) and contraction velocity. The inability to control a load at any velocity, for any contraction type or at any given range of motion would reflect a strength deficit.

Stabilizer - The muscle(s) primarily responsible for providing stabilization at one segment while movement is occurring about other segments.

Dynamic Joint Stability – The ability of a segment to maintain optimal orientation of the articular surfaces during loaded conditions, i.e. during periods of accelerated motion. Acceleration is an integral component when considering human motion. Motion is dictated by the Newtonian Equations of Motion where the two equations must be satisfied; 1) the sum of all forces = mass x acceleration, and 2) the sum of all moments = moment of inertia x angular acceleration. When a system is at rest (undergoing zero acceleration, a special condition exists for the Equations of Motion where the acceleration is equal to zero. This is called Static Equilibrium occurs when the balance of forces and moments within the system result in zero acceleration or when the sum of all forces or moments equals zero. True static conditions rarely exist in the human system. The ability of the human system to minimize acceleration will directly impact the moments or forces at the joint.

Fatigue - Neuromuscular fatigue is a reduction in the ability of a muscle to produce a resultant joint moment that occurs with repeated contractions and results in a decreased ability to control or produce segmental rotation or control joint translations. Fatigue reflects acute changes in both the peripheral and central components of motor performance.

Differential fatigue - The different magnitude of fatigue occurring between muscle groups. This can either be between the prime mover and the stabilizer, between prime movers, between synergists or between stabilizers about a joint.

INTRODUCTION

An increased understanding of glenohumeral rotator fatigue is essential for insight into the mechanisms of dynamic joint stability and instability. This understanding will provide guidance for the design of exercise programs targeted at preventing disorders associated with inadequate dynamic joint stability and promoting recovery from shoulder injury.

The glenohumeral joint has the greatest range of motion of any joint but is normally stable during three-dimensional motion. In order to accomplish this it relies heavily on the local tissues for its stability due to the minimal contribution of inherent constraint provided by the bony anatomy (Cure 1996, Bigliani, Kelkar, Flatow, Pollock and Mow 1996). Dynamic stabilization of the shoulder is provided by the surrounding musculature. The rotator cuff muscles (supraspinatus, infraspinatus, teres minor and subscapularis) are the primary dynamic stabilizers of the glenohumeral joint. Together these muscles are necessary to provide normal shoulder kinematics and glenohumeral stability. EMG studies have clearly demonstrated that the muscles of the rotator cuff are active throughout the entire range of arm elevation to varying degrees (Arwert, deGroot, Van Woensel and Rozing 1997, Kronberg, Nemeth and Brostrom 1990, Saha 1971, Sjogaard and Sogaard 1998). EMG activity of 10 to 20% of maximum has been demonstrated in all the rotator cuff muscles during a precision hand movement task, confirming their role in achieving proximal stability (Laursen, Jensen and Sjogaard 1998).

The net effect of all the forces spanning the shoulder joint is to minimize translations of the humeral head on the glenoid fossa in three dimensions including both superior-inferior and anterior-posterior translations (Culham and Peat 1993, Inman, Saunders and Abbott 1944, Poppen and Walker, 1976). Control of translation of the humeral head is directly dependent on the precise integration of afferent information by the central nervous system and subsequent efferent adjustments and control (Enoka 1994).

Fatigue by definition will result in decreased muscular performance (Enoka 1994) and may subsequently impact upon the ability of the dynamic stabilizers to control glenohumeral translation. The central nervous system can employ a variety of different strategies, such as reducing motor neuron firing rates, load sharing, and alternate recruitment to facilitate optimal performance and minimize the development of fatigue. There also is evidence that both the synergist and antagonist muscles are activated differently by the nervous system depending on the type, duration or intensity of the task (Pincivero, Aldworth, Dickenson, Petry and Shultz 2000, Tamaki, Kitada, Akamine, Murata, Sakou and Kurata 1998).

During their role as prime movers, the internal and external rotators operate with a defined strength ratio that is variable between individuals (Cook Gray, Savinar-Noque and Medeiros 1987, Hughes, Johnson, O'Driscoll and An 1999, Timm 1997). An appropriate balance of dynamic stabilizer muscle strength contributes to the maintenance of normal shoulder kinematics (Hughes et al. 1999). An alteration of balance between the internal and external rotator strength has been demonstrated during fatigue when they are acting as prime movers (Ellenbecker and Roetert 1999). Imbalances between the shoulder internal and external rotators may disrupt neuromuscular control and result in undesirable humeral head migration during upper extremity tasks. This could result in impingement of the rotator cuff tendons under the coracoacromial arch and subsequent degenerative changes; ultimately producing a partial or complete tear of the rotator cuff tendons. Rotator cuff disorders are the most common cause of shoulder pain and dysfunction (Cohen and Williams 1998, Matsen and Arntz 1990b, Soslowsky, Carpenter, Bucchieri and Flatow 1997a).

The findings of Pincivero et al. (2000) regarding EMG activation of the quadriceps and hamstrings during a lunging task, suggest that antagonists and agonists may fatigue differently. There have been no studies that have examined the concept of differential fatigability of the shoulder muscles during upper limb tasks, or for that matter this concept has not been thoroughly evaluated in the scientific literature for any body segment or motion. Differential fatigue would be especially significant between the internal and external rotators, because it would disturb the existing strength ratio that contributes to the dynamic stability of the glenohumeral joint.

Despite the importance of the rotator cuff on dynamic joint stability, data available regarding fatigue of the glenohumeral musculature is limited to their role as prime movers and does not address fatigue during their role as stabilizers. Research on fatigue during an upper limb task has focused primarily on the deltoid and upper trapezius muscles and has essentially ignored the important role of the internal and external rotator muscles.

The purpose of this proposed study is to describe the differential fatigability of upper extremity musculature (specifically the internal and external glenohumeral rotators and the forearm flexors) during repetitive grip tasks. This study will help to fill the literature gap and explore the possibility of glenohumeral rotator fatigue as a potential aetiology of various shoulder conditions including impingement syndrome, and repetitive stress injury (including but not exclusive to tendonitis, tendinosis etc.) of the upper extremity. Research and therapeutic interventions have primarily focused on performance as it relates to strength and not fatigue. The results of this study will have direct implications for clinical prevention and intervention in the area of shoulder rehabilitation and will provide a basis for further research into this area.

REVIEW OF LITERATURE

The following review of literature establishes the importance of research directed to understanding the aetiology of shoulder dysfunction and injury as well as insight into joint stabilization throughout the human body. A brief overview of glenohumeral anatomy and biomechanics is followed by a more thorough review of dynamic joint stabilization and fatigue. Although there are very few published articles that directly impact the proposed research, a critical evaluation of relevant studies is provided within this review.

A. Anatomy and Biomechanics of the Glenohumeral Joint

A brief review of the anatomy and biomechanics of the glenohumeral joint will provide essential background information to assist in the understanding of the dynamic stability of the glenohumeral joint. The shoulder complex is composed of both the glenohumeral and scapulothoracic articulations and movement is usually generated at all three joints concomitantly (Culham and Peat 1993). For the purposes of this review, only the glenohumeral articulation will be considered.

Anatomy

The shoulder complex is afforded the greatest mobility of all the joints in the human body and primarily functions to position the hand in space. This enormous mobility is due in part to both the incongruity of the glenohumeral components, and the additional contribution to movement allowed by the scapulothoracic articulation. The following structural components of the glenohumeral joint each contribute to the considerable overall motion of the shoulder complex.

Glenoid

The glenoid is a thickening of the scapula just inferior to the acromion. The glenoid surface is slightly concave with a total surface area 3 to 4 times smaller than the humeral head that must fit into it. The supraglenoid tubercle provides attachment for the long head of the biceps tendon, while the infraglenoid tubercle provides attachment for the long head of the triceps tendon (Curl and Warren 1996).

Glenoid Labrum

The glenoid labrum is a fibrocartiliagenous ring attached around the periphery of the glenoid rim (Culham and Peat 1993). The glenoid labrum is firmly attached inferiorly, but loosely attached superiorly and often meniscoid. It provides attachment for the glenohumeral ligaments and receives supplementary reinforcement from them. The glenoid labrum functions to increase the anteroposterior depth of the socket from 2.5 to 5 mm and deepens the socket to 9 mm in the superio-inferior direction thereby improving the congruity of the articular surfaces. It also enhances stability by increasing the surface area for contact of the humeral head. The labrum varies from individual to individual and may even be absent in some (Levine and Flatow 2000).

Glenohumeral Joint Capsule

The thin loose articular capsule has a surface area two times greater than the humeral head to allow for a large range of motion. It is attached to the glenoid labrum and the neck of the humerus. Left alone, the capsule would contribute very little to glenohumeral stability. It is reinforced by the glenohumeral ligaments and is strengthened dynamically on all surfaces except the inferior one by the tendons of the rotator cuff musculature (Speer 1995).

Humeral Head

The humeral head is spherically shaped and faces medially, superiorly and posteriorly. The humeral head is three times greater in surface area than the glenoid that enhances range of motion. There are two prominences located anteriorly for both tendon and ligament attachment: the lesser tuberosity and greater tuberosity (Culham and Peat 1993).

Coracoacromial Arch

The coracoacromial arch is composed of the acromion, coracoid process and the coracoacromial ligament. The arch forms a roof over the space where the rotator cuff tendons must pass (Kent 1971). This area is often termed the impingement region. Variations in these anatomical components may contribute to the pathogenesis of impingement syndrome.

Glenohumeral Ligaments

The glenohumeral ligaments are thickenings of the capsule that vary considerably. The superior glenohumeral ligament courses from the anterior glenoid labrum to the lesser tuberosity and can vary considerably in size. The middle glenohumeral ligament arises next to the superior glenohumeral ligament from the anterior labrum and inserts into the lesser tuberosity blending with the subscapularis tendon. It too varies considerably in size. The inferior glenohumeral complex arises from the anterior labrum and inserts into the lesser tuberosity (Bigliani et al. 1996, Speer 1995). The inferior glenohumeral ligament consists of three distinct bands: anterior, posterior and axillary pouch. The three bands together form a sling that changes function depending on the position of the humeral head (Levine and Flatow 2000). The rotator interval is the triangular-shaped area between the supraspinatus and subscapularis tendons and the base of the coracoid process (Levine and Flatow 2000).

Scapulohumeral Muscles

Due to minimal contact between the glenoid and the head of the humerus, the shoulder joint depends heavily on the muscular structures for stability. The following muscles originate on the scapula and attach directly onto the humerus, providing the necessary forces for glenohumeral motion. Due to their specialized relationship, the muscles that constitute the rotator cuff are considered separately at the end of this section.

Biceps. The long head of the biceps tendon with its synnovial sheath runs from the supraglenoid tubercle and the glenoid labrum to the bicipital tuberosity on the radius. It runs under the coracohumeral ligament between subscapularis and supraspinatus. The short head of the biceps arises from the coracoid process. The primary actions of the biceps are elbow flexion and forearm supination. Secondarily, it also provides stabilization for the glenohumeral joint (Rodofsky, Harner and Fu 1994). The coracohumeral ligament functions to stabilize the biceps tendon even more so than the transverse humeral ligament (Matsen and Arntz 1990a). It is innervated by the musculocutaneous nerve.

Deltoid. The Deltoid originates on the lateral clavicle, acromion and the scapular spine and inserts into the deltoid tuberosity of the humerus. It is the primary abductor of the humerus. The anterior fibres of the deltoid contribute to humeral flexion and the posterior fibres contribute to extension. The deltoid muscle is innervated by the axillary nerve (Kent 1971).

Teres Major. Teres Major takes its origin from the posterior surface of the inferior angle of the scapula and inserts on the medial aspect of the intertubercular groove of the humerus. The tendon fibres spiral 180 degrees as they course towards the humerus. The primary actions of the teres major are internal rotation, adduction and extension of the humerus. It is innervated by the subscapular nerve (Speer 1995).

Triceps. The long head of the triceps muscle originates on the inferior glenoid labrum and inserts with the lateral and medial heads of the triceps onto the olecranon process of the humerus. The main action of the triceps is elbow extension, but the long head also contributes to extension and adduction of the humerus. The triceps is innervated by the radial nerve (Culham and Peat 1993).

Coracobrachialis. The coracobrachialis muscle originates from the coracoid process together with the short head of the biceps muscle and inserts into the medial aspect of the central humerus. It contributes to both flexion and adduction of the humerus. It is innervated by the musculocutaneous nerve (Speer 1995).

Rotator Cuff. The rotator cuff consists of four muscles: supraspinatus, infraspinatus, teres minor and subscapularis that run from the scapula to the tuberosities of the humerus. The tendons of the rotator cuff blend and fuse with one another and the joint capsule as they cross the joint. This results in the sharing of loads between tendons (Soslowsky et al. 1997a). The tendons are penetrated by the long head of biceps between the subscapularis and supraspinatus tendons. These muscles function to stabilize the glenohumeral joint by opposing the force of the deltoid muscle and centring the humeral head against the glenoid fossa during motion of the shoulder. They prevent excessive translation and significantly contribute to rotation and elevation, especially during the first 90° of elevation (Harryman, Sidles, Clark, McQuade, Gibb and Matsen 1990, Curl and Warren 1996). In order to achieve this, these muscles must both produce a moment about the glenohumeral joint to assist arm elevation and rotation, and produce a downward force to counteract the upward force produced by the deltoid.

Supraspinatus. The supraspinatus muscle originates on the dorsal surface of the scapula in the supraspinous fossa and then forms a tendinous insertion onto the superior aspect of the greater tuberosity. The muscle is fusiform shaped and is classified as a circumpennate muscle. The primary actions of the supraspinatus muscle are initiation of elevation and participation in the stabilization force couple. The suprascapular nerve innervates the supraspinatus muscle (Culham and Peat 1993).

Infraspinatus. The infraspinatus muscle arises from the dorsal surface of the scapula in the infraspinatus fossa and on the spine of the scapula and inserts onto the middle facet of the greater tuberosity. It is a circumpennate muscle. The primary actions of the infraspinatus muscle are external rotation (60%) and joint stabilization. It is innervated by the suprascapular nerve (Culham and Peat 1993, Speer 1995).

Teres Minor. Teres minor has its origin on the inferior portion of the lateral border of the scapula and it inserts onto the inferior facet of the greater tuberosity. It is also a circumpennate muscle. The primary actions of the teres minor muscle are external rotation (45%) and joint stabilization. It is innervated by the posterior branch of the axillary nerve (Speer 1995).

Subscapularis. The subscapularis originates on the anterior surface of the scapula in the subscapular fossa and inserts onto the lesser tuberosity. It is a multicircumpennate muscle. It is the only muscular component of the anterior capsule. The fibres of its tendon fuse with the middle and inferior glenohumeral ligament affording it a passive stabilization function also. The primary actions of the subscapularis are internal rotation and stabilization of the humerus. It is innervated by the subscapular nerve (Curl and Warren 1996).

Biomechanics of the Shoulder

The glenohumeral joint is a synovial ball and socket joint with three degrees of freedom of motion: flexion/extension, abduction/adduction and internal/external rotation (Kent 1971). The range of motion of the shoulder joint is considerable due to the previously described anatomical features. Normal glenohumeral kinematics are achieved through the interaction of the deltoid, rotator cuff, biceps tendon, glenohumeral capsule and the scapular muscles. Elevation of the arm is the result of both glenohumeral joint motion together with accompanying movements of the clavicle and scapula and is described in relation to the cardinal planes of the body. Flexion/extension take place in the sagittal plane. Abduction/adduction take place in the coronal plane. Internal and external rotation is described as taking place around a longitudinal axis through the humeral shaft (Kent 1971). The amount of internal and external rotation possible varies with arm position; 180° is possible with the arm positioned in neutral, however this amount reduces to only 90° with the arm in 90° of elevation and becomes only minimal when the arm is fully elevated. Translation of the humeral head occurs both in anterior-posterior and superior-inferior directions during both passive and active arm elevation. There is a greater degree of anterior-posterior translation due to the increased curvature of the glenoid in a superior-inferior direction. Under normal conditions, the humeral head translates less than 2 mm superiorly during active elevation (Karduna, Williams, Williams and Ionotti 1996, Poppen and Walker 1976). Glenohumeral kinematics are directly affected by shoulder pathology including rotator cuff lesions, anterior instability and reduced mobility.

Codman first introduced the notion of kinematic interaction between the scapula and humerus in the 1930's when he defined the concept of scapulohumeral rhythm (McQuade and Smidt 1998). Inman first identified the scapulohumeral rythym as a 2:1 ratio in 1944 (Inman et al. 1944). This ratio still applies when the total range of motion is considered. In the early stages of elevation the movements are primarily glenohumeral but after 30 to 45° the contribution of the scapula and clavicle increases, with reported ratios ranging from 1.35:1 to 7:1. The contribution of the scapula to the overall movement changes depending on the load requirements of the arm during elevation, making the relationship of the scapula and the humerus during elevation more complex than often thought (Mandalidis, McGlone, Quigley, McInerney and O'Brien, 1999, McQuade and Smidt 1998).

There is no consensus as to what constitutes normal glenohumeral joint mechanics. Studies generally have examined either joint geometry or motion characteristics, but have not related the two to each other (Bigliani et al. 1996). Major controversies that exist include the shape of the glenoid and humeral head, the conformity of the glenohumeral articular curvatures and the kinematics of the shoulder. Development of controversy may be due to limits in technology and the differences in protocols used in the research studies. Kinematic analysis using only bone data with radiographic examination eliminates the effect of cartilagenous tissue on the conformity of the humeral head and the glenoid fossa (Bigliani et al. 1996). Description of elevation of the arm in relation to the plane of the scapula rather than the cardinal axes of the body is also used to describe glenohumeral joint motion. The plane of the scapula is perpendicular to the glenoid plane and has been defined as lying obliquely between the frontal and sagittal plane, 30-45 degrees anterior to the coronal plane (Culham and Peat 1993, Poppen and Walker 1976).

B. Stabilization of the Glenohumeral Joint

The glenohumeral joint has the greatest range of motion of any joint but is normally stable during unrestrained, three-dimensional motion. Review of the anatomical components of the glenohumeral joint highlights the importance of the muscular and ligamentous components to provide the necessary stability. The forces that arise from the contractile apparatus provide dynamic stabilization. The muscles of the glenohumeral joint and scapula serve as dynamic stabilizers and prime movers during tasks of the upper extremity (Hughes and An 1996, Kronberg et al. 1990). Passive stabilization is provided by the articular surfaces, the glenoid labrum and the capsule with its ligamentous components, passive muscle tension, along with negative intraarticular pressure and joint fluid adhesive properties (Cure 1996, Bigliani et al. 1996). The role of tissue viscosity, which is directly related to velocity, in joint stabilization has not been explored in the literature. The contribution of both the dynamic and passive stabilizers to the maintenance of stability requires integration and coordination by the neuromuscular system.

Passive Stabilizers

The principal static stabilizers of the glenohumeral joint are the capsuloligamentous restraints and the negative intraarticular pressure that creates a vacuum inside the joint (Sharkey and Marder 1995, Harryman et al. 1990). There is a complex interaction of the capsule and the labral attachments to maintain stability during activity that varies depending on the position of the shoulder (Curl and Warren 1996).

The concavity of the glenoid fossa contributes to stability by centring the head of the humerus on the glenoid. Traditional x-ray analysis of the humeral head and the glenoid surface only depicts the incongruent bony surfaces and does not take into account the conformity afforded by the cartilagenous tissue, therefore underestimating the true congruence of the joint surfaces (Bigliani et al. 1996).

Superior-Inferior Translation

The coracohumeral ligament together with the superior glenohumeral ligament is considered to limit inferior translation when the arm is adducted (Levine and Flatow 2000 Soslowsky, Malicky and Blasier 1997b). With the arm at 0°, there is some disagreement between the contribution of the coracohumeral ligament and the superior glenohumeral ligament. The inferior glenohumeral ligament is the most important restraint to inferior translation at 45° and 90°of abduction: the anterior band in neutral and internal rotation and the posterior band in external rotation (Levine and Flatow 2000). Bigliani and associates (1996) identify the inferior glenohumeral ligament as the major anterior-inferior static stabilizer, with failure occurring either in the substance of the ligament and resulting in capsular stretching, or at its attachment to the glenoid resulting in a Bankart lesion. When the superior labrum is detached, there is increased strain on the inferior glenohumeral ligament and increased multidirectional translation. The rotator interval capsule indirectly stabilizes the shoulder inferiorly by maintaining negative intraarticular pressure. The coracoacromial ligament and the acromion are the major restraints to superior translation.

No single structure is primarily responsible for stability at all positions (Bigliani et al. 1996).

Anterior-Posterior Translation

The inferior glenohumeral ligament functions to provide the majority of restraint to anterior translation (Bigliani et al. 1996). The three bands of the ligament contribute in different positions of glenohumeral joint abduction (0, 45 and 90°) in a complex fashion. Position and tightness of the anterior structures varies with the position of the arm. During internal rotation, the complex moves posteriorly and limits posterior translation. Conversely during external rotation it moves anteriorly to limits anterior translation. The middle glenohumeral ligament limits anterior translation of the humeral head with the arm in 60 to 90° of abduction (Levine and Flatow, 2000). The geometry and biomechanical properties of the glenohumeral ligament contribute to its suitability as the primary anterior stabilizer. The coracohumeral ligament and the tendons of subscapularis, infraspinatus and teres minor also limit posterior translation (Cain, Mutschler, Fu and Lee 1987, Curl and Warren 1996).

The glenohumeral ligaments function together with the muscles to restrain the humeral head and facilitate normal function. Incongruent joints have larger translations at the extremes of range. Small translations of the humeral head occur throughout the normal range of motion (Graichen, Stammberger, Bonel, Karl-Hans, Reiser and Eckstein 2000, Harryman et al. 1990, Sharkey and Marder 1995). Kinematic analysis using traditional x-rays of the humeral head significantly overestimates the actual translation of the humeral head by not taking into account the increased conformity that occurs with the addition of the cartilagenous tissue. These results should be interpreted with caution (Bigliani et al. 1996). Analysis using magnetic resonance imaging would provide more a more comprehensive assessment of both bony and soft tissue effects.

The selected cutting studies in cadavers that examine the functions of these structures are limited by the considerable inherent variability in the anatomy of the glenohumeral ligament in attachment and size, which must be kept in mind when interpreting the results of the studies (Bigliani et al. 1996, Cain et al. 1987, Levine and Flatow 2000) As well, the clinical relevance of applied loads has not been shown and varies between studies. These studies were performed while excluding the dynamic stabilizers, which eliminates the joint compression forces. They cannot provide the relative contribution of both passive and dynamic forces unless they are examined together.

Dynamic Stabilizers

Dynamic stabilizers of the shoulder are necessary to provide normal shoulder kinematics and glenohumeral stability. The ligamentous structures function only at the extreme positions of rotation to prevent excessive translation: in the mid-range they are relatively lax (Soslowsky, An, DeBano and Carpenter 1996, Bigliani et al. 1996). In the mid-range most joint stability is through the co-ordinated and synergistic action of the glenohumeral musculature (Bigliani et al., 1996, Kronberg et al. 1990). Equilibrium is established by the interaction of three forces: the weight of the upper extremity, the abduction force produced by the deltoid and supraspinatus, and the resultant forces of the intraarticular pressure and downward pull of the rotator cuff (Inman et al. 1944). This is achieved with coordinated muscle contraction through appropriate neural activation strategies. Joint stability is enhanced through muscular contraction by increasing compression of the articular surfaces, inducing motion that results in tightening of ligamentous structures, forming a muscular barrier to translation, and centring the force of the humeral head on the glenoid surface through coordinated muscle activity (Halder, Itoi and An 2000, Levine and Flatow, 2000). The rotator cuff muscles (supraspinatus, infraspinatus, teres minor and subscapularis) are the primary dynamic stabilizers of the glenohumeral joint. The deltoid, long head of biceps and other muscular structures about the shoulder provide a certain degree of stabilization as well (Levine and Flatow 2000, Matsen and Arntz 1900b). During elevation of the arm through abduction, the dynamic force requirements reach their maximum at 60°, and the force falls dramatically after 90°(Inman et al. 1944). The dynamic stabilizers of the shoulder complex include not only the glenohumeral muscles, but also the scapulothoracic and thoracohumeral muscles of the shoulder girdle. Acting together the scapulothoracic and thoracohumeral muscles form a force couple necessary for upward scapular rotation. The magnitude of the forces required from these muscles to rotate the scapula is much smaller than those required for arm elevation (Inman et al. 1944). For the purposes of this review only the muscles involved with the glenohumeral joint will be examined.

The Role of the Rotator Cuff.

The rotator cuff functions both as a stabilizer of the glenohumeral joint to optimise positioning of the humeral head, and as a prime mover for rotation and elevation of the humerus (Matsen and Arntz 1990b). The subscapularis is the primary posterior stabilizer when the arm is in 90° of flexion. For anterior stabilization there is equal contribution from the subscapularis, teres minor, supraspinatus, and infraspinatus muscles. Acting as a group, these muscles form a force couple with the deltoid in the frontal plane (Culham and Peat 1993). This is a simplistic concept developed to understand the role of forces in controlling the relative translation of the humerus on the glenoid, rather than a comprehensive mathematical model. A force couple is a moment created by two equal but oppositely directed forces (Mantone, Burkhead, and Noonan 2000). For the frontal plane force couple, the deltoid and supraspinatus are the elevators, and the infraspinatus and teres minor the depressors that helps to maintain the relationship between the humeral head and the glenoid fossa and prevent the upward displacement of the humeral head on the glenoid, which would otherwise occur with unopposed action of the deltoid and supraspinatus (Culham and Peat 1993, Inman et al. 1944, Poppen and Walker, 1976).

Kuechle, Newman, Itoi, Niebur, Morrey and An (1997) used moment arm analysis and designated the supraspinatus as the most efficient elevator and teres minor as the most efficient depressor during abduction of the glenohumeral joint. As the angle of abduction increases, infraspinatus changes from an elevator to a depressor, and the reverse occurs with subscapularis. The subscapularis and teres minor form a force couple in the horizontal plane (Speer 1995). When the rotator cuff muscle are loaded together, stabilization of the humerus occurs in all four directions (Halder, et al. 2000). Kronberg et al. (1990) demonstrated the activation of the rotator cuff to limit translation of the humeral head using EMG analysis.

In a study by Sharkey and Marder (1995), 5 intact shoulder cadaver specimens from both males and females (mean age76 years) were examined, along with 5 healthy male shoulders (mean age 33) using simulated muscle action (with nylon webbing, cables and computer-driven motors) for the deltoid, supraspinatus, infraspinatus-teres minor, and subscapularis muscles. Forces for the long head of the biceps muscle were not simulated despite possible function of the biceps to prevent upward migration when activated (Kumar, 1989). For each specimen the arm was abducted in the plane of the scapula from 0° to 120° with the humerus in neutral rotation. Glenoid height, humeral head diameter, and glenohumeral positioning were measured at 30°, 60°, 90° and 120° using radiographs and computer software. Their results showed that the vertical position of the humeral head in relation to the glenoid varied with position and muscle activity. In the cadaveric specimens, when the deltoid muscle was activated alone, there was a significant shift in the humeral head position superiorly at all 4 positions by up to 1.3mm (at 120°). When the entire rotator cuff was activated at a constant force throughout the range, the position of the humeral head varied from 0.0mm at 30° abduction to 1.5mm at 120°. Measurement taken in the healthy shoulders also demonstrated a progressive shift upward in head position (up to 0.7mm at 120°), which is less than 50% of the values obtained for the cadaver specimens. These differences may be in part due to the absence in the cadaver models of the numerous other muscles that are active during elevation of the arm such as teres major, latissimus dorsi, pectoralis major and particularly the long head of biceps brachii. However, this study does display a significant difference in humeral head position between deltoid activity alone and deltoid and rotator cuff activity, suggesting the importance of the rotator cuff activity to humeral head positioning regardless of the contributions of other muscles.

Graichen et al. (2000) confirm the importance of muscular control for joint stability with 3D MRI evaluation of the humeral head during both passive and active elevation. Their results clearly demonstrated the centring effect of the rotator cuff on the humeral head during elevation.

Other Muscular Contributions

Deltoid. The exact contribution of the deltoid to stability of the shoulder has not been clearly demonstrated. Assumptions regarding its role have been based upon its large size and power.

Biceps. When the long head of biceps contracts it may prevent upward migration of the humeral head and therefore contribute to the stability of the glenohumeral joint (Pagnani, Deng, Warren, Torzilli and O'Brien 1996, Kumar 1989). The long head of biceps may be active during abduction, especially when the arm is in external rotation (Inman et al. 1944). EMG studies have shown that biceps activity during abduction peaks at 132° with the arm in neutral rotation. It is also active during flexion from 0° to 90° with the arm in external rotation (Matsen and Arntz 1990a). The biceps tendon may also contribute passively to stability. Decreased translation has been noted with loading of the biceps tendon especially in external rotation both for anterior-posterior and inferior translations (Rodosky, Harner and Fu 1994). There is increased strain on the biceps tendon when there has been a rotator cuff tear; consequently the majority of biceps tendonitis is seen following a tear in a rotator cuff tendon (Matsen and Arntz 1990a).

Assessment of Muscle Contribution to Glenohumeral Motion

The muscles about the shoulder must be able to create adequate moments in order to achieve elevation of the arm. Verification of glenohumeral muscle function has been investigated through electromyographic analysis, cadaver studies, biomechanical analysis and nerve block studies. EMG studies have clearly demonstrated that the muscles of the rotator cuff are active throughout the entire range of arm elevation to varying degrees (Arwert et al. 1997, Bradley and Tibone 1991, Inman et al. 1944, Kronberg et al. 1990, Perry, Barnes and Merson 1989, Saha 1971, Sjogaard and Sogaard 1998) both concentrically and eccentrically (Kronberg and Brostrom 1995). EMG activity is an indicator of muscular activity, but not a direct indicator of muscle force or moment produced by the muscle. The contribution of the muscles when summated closely approximates the force analysis requirements for abduction (Inman et al. 1944). When examined using EMG and dynamometry, Perry et al. (1989) calculated that the supraspinatus muscle contracts with a constant force at approximately 40% of maximum during abduction, the infraspinatus contracts at 20% of maximum, and subscapularis contracts with a similar pattern but less force. This is in partial agreement with the EMG results of Kronberg et al. (1990) which identified supraspinatus activity as averaging between 30-40% of maximal activation through abduction, with infraspinatus following a similar pattern of activation but averaging an even higher percentage. These muscles show a more consistent level of activity throughout the range of motion than the prime movers (deltoid) for both flexion and abduction. During external rotation, infraspinatus was activated to almost 100% at mid-range with the arm in the neutral position, but for internal rotation subscapularis and pectoralis major did not exceed 30% activation. From these results they concluded that the rotator cuff muscles act as stabilizers during movements of the glenohumeral joint.

Kronberg, Brostrom and Nemeth (1991,1995) examined the shoulder muscles both concentrically and eccentrically in normal and unstable shoulders. Highlights of their findings include: firstly, several rather than single muscles are active for each movement, secondly, EMG activity is increased in those shoulders with instability, particularly in supraspinatus, subscapularis and infraspinatus muscles which are identified as stabilizers during movements of the shoulder, and thirdly, the activity in the stabilizer muscles is increased in unstable shoulders. Increased muscle activity patterns were seen in supraspinatus, subscapularis and infraspinatus in shoulders with joint laxity. This is likely due to the increased need for stability from the musculature to increase anterior stability and compress the humeral head into the glenoid fossa (Kronberg and Brostrom 1995).

Results from biomechanical models (Hughes and An 1996, Poppen and Walker 1976) for infraspinatus, supraspinatus and subscapularis also indicate concurrent activation of these muscles and agree with EMG study results. The synergistic activation of infraspinatus and supraspinatus during abduction functions to produce stabilization by the negation of the internal and external moments generated by the two muscles. These studies have also demonstrated that the role of the supraspinatus is to act synergistically with the deltoid to produce elevation of the arm and assist in stabilization of the glenohumeral joint (Hughes and An 1996, Inman et al. 1944, Kronberg et al. 1990 Kronberg and Brostrom 1995, Soslowsky et al. 1997a, Poppen and Walker 1976). More recently, Gagey and Hue (2000) suggest that the deltoid muscle compresses the humeral head against the glenoid and prevents upward displacement of the humerus. This theory contrasts with the original model by Inman (1944), which describes only an upward force exerted on the humerus during the onset of elevation. This discrepancy likely evolved from the realization that the deltoid functions as multiple units and can be controlled with varying strategies by the central nervous system (Wickham and Brown 1998).

Kuechle et al. (1997) categorized the rotator cuff muscles using moment arm calculations as an indication of muscle efficiency. They predicted that teres minor, followed by infraspinatus would be the most efficient external rotators with the arm in both neutral and 90° of abduction. Subscapularis followed by pectoralis major, latissimus dorsi and teres major were predicted to be the most efficient internal rotators with the arm in both neutral and 90° of abduction. These predictions are supported by EMG study results (Arwert et al. 1997, Kronberg et al. 1990, Perry et al. 1989). Clinically, shoulder moments are measured using an isokinetic dynamometer, but this cannot provide information on the function of each muscle individually.

Studies examining both EMG activity patterns and biomechanical analysis during shoulder motion have provided compelling evidence for the role of the rotator cuff muscles as dynamic stabilizers. However, the role of the rotator cuff in the aetiology of shoulder dysfunction has not been clearly elucidated.

C. Neuromuscular Stabilization

The activation of muscles for the purpose of joint stabilization is designed to enhance stability of the entire linked system. Joint stability requires an induced force from the system that is opposite to the displacement force and is related both to the muscle force produced and the moment arm or joint position. The dynamic stabilizers ensure that the articular surfaces and body segments that are not involved in the movement maintain their optimal orientation, as well as maintain the position of the system relative to the base of support. This requires involuntary activity from both postural and segmental stabilizer muscles.

There is ample evidence in the literature indicating that muscle activity occurs distant from the actual body segment in motion (Bouisset, Richardson and Zattara 2000, Laursen et al. 1998, Zedka and Prochazka 1997), and that this associated muscle activity is correlated to acceleration (Bouisset et al 2000). Zedka and Prochazka (1997) suggest that peripheral rather than central feedback mechanisms may be responsible for this orchestration of motor activity for proximal joint (trunk) stabilization during rapid voluntary, distal segment motion. However, both peripheral and central mechanisms should be considered until further evidence is provided. The achievement of joint stability is dependent not only on the afferent information provided to the system, but also on the subsequent neural regulation of muscle force.

Afferent Feedback

Neural feedback from sensory receptors provide information to the system on its state and environment in order to mediate motoneuron activity through both direct synapse to the motoneuron and by supraspinal pathways. These specialized receptors transform mechanical deformation related to changes in position, force, velocity and pressure into electrical signals. The system is able to organize and generate an appropriate rapid change in force or position in response to any disturbance, thereby contributing to stabilization and enhancing subsequent movement. This is known as feedback and is provided by both proprioceptors and exteroceptors. Proprioceptors detect information generated within the system itself, and exteroceptors discern external stimuli. These same receptors also provide important information on joint angle and orientation for more rapid control known as feed-forward control. This processing system is used for anticipatory commands: for example to predict the path of a ball and place the hand in the appropriate position to catch the ball (Pearson and Gordon 2000).

Proprioception is the ability to sense the position and movement of the limbs without the use of visual feedback. Proprioception requires the imperceptible ability of the feedback mechanisms to provide rapid information to the central nervous system. It encompasses both the sensation of joint motion (kinaesthesia) and joint position (joint position sense). Three types of mechanoreceptors located in both the muscle and the joint provide this information: (1) muscle spindles (2) golgi tendon organs and (3) joint receptors (Enoka 1994).

Muscle spindles are a collection of mini muscle fibres enclosed in a capsule (intrafusal) that lie parallel to the muscle fibres and are innervated by group Ia and II myelinated afferents. Group Ia afferents are able to detect small changes in muscle fibre length and respond to the velocity as well as the overall amount of change in fibre length. In contrast, group II afferents respond only to the change in length. As a result, information regarding not only joint and segment position, but length and rate of change of the muscle length is provided by the spindles (Enoka 1994). The muscle spindle can be regulated by the central nervous system through the gamma motoneurons, which innervate the intrafusal fibres within the spindle, thereby "presetting" the spindles with appropriate tension and allowing spindles to function at a variety of muscle lengths. (Pearson and Gordon 2000).

Golgi tendon organs are sensitive to muscle force and relay information by Ib myelinated afferent fibres. Group III afferents are thinly myelinated and conduct impulses at a slower rate than group Ia or II fibres, and group IV afferents are unmyelinated. Many of these endings are associated with blood vessels in the tendon.

Joint receptors, located primarily in the joint capsule, provide information regarding joint position, acceleration, joint displacement and noxious stimuli (Pearson and Gordon 2000). This is achieved through the use of both rapidly and slowly adapting mechanoreceptors and nociceptors. The rapidly adapting mechanoreceptors are pacinian corpuscles, which respond to mechanical stimuli. The slowly adapting joint receptors are Ruffini's endings, which sense stretch and respond to movements of the joint at the extremes of range. Joint mechanoreceptors are innervated by type II afferent fibres. Joint nociceptors respond primarily to movements beyond normal range, but may be sensitised to respond to innocuous stimuli during inflammatory states. Joint nociceptors are innervated by group III or group IV afferent fibres.

Proprioceptors enable the central nervous system to: precisely control the length and rate of change of muscle length, appropriately sequence muscle activity, and coordinate activity in multiple segments in order to stabilize and enhance subsequent movement (Enoka 1994). The central nervous system is required to control the force output of the muscle(s) involved in order to achieve this.

Force Regulation

The motor unit is the smallest element of the neuromuscular system where regulation of force occurs through both the recruitment of additional motor units and discharge rate modulation of active motor units (Tamaki et al. 1998). There are three distinct types of motor units within the muscle: slow twitch, fast twitch fatigue resistant, and fast twitch fatigable. The motor units in a given muscle are rarely recruited synchronously (Tamaki et al. 1998). Altering the amount of motor unit activity and the properties or characteristics of the muscle or motoneuron controls force (Loeb and Ghez 2000). Neuromotor control strategies can differ within segments of the same muscle, enabling individual segments to perform a variety of functions (Tamaki et al. 1998, Wickham and Brown 1998). The properties of the muscle that contribute to force will change depending on the length of the muscle fibre, the velocity of the contraction, and the architecture of the muscle (angle of pennation). The length of the muscle fibre and the velocity of the fibre contraction directly affect the number of cross-bridges that can attach by the amount of overlap between the myosin and actin filaments and the time allowed for cross-bridge formation.

The contribution of muscles to movement is dependent on their ability to produce moment or moment about the joint. The moment produced is a product of the force generated by the muscle and the length of the moment arm. There are five system components in the development of joint moment:

1. The rigid bone, tendon and ligament link (moment arm length, number of joints involved, other muscles involved)
2. The synnovial joint,
3. The characteristics of the muscle (cross-sectional area, fibre length and architecture),
4. The neural afferents and efferents, and
5. The sensory receptors.

The resultant muscle force is the net effect of the muscles that cross the joint, not the force from an individual muscle (Enoka 1994). The control of movement requires a minimum of one pair of opposing muscles.

The ideation, programming, execution and modification of motor commands are coordinated suprasegmentally in the cerebral cortex as well as the limbic system, cerebellum and basal ganglia. The descending motor system thus exerts effects onto a segmental motor system that is itself modifiable. Due to the complexity of both the suprasegmental centers and the musculoskeletal system, many different activation patterns and muscle groups can be organized to perform a desired movement. For example, the excitability of motoneurons, of interneurons interposed in reflex pathways, and transmission from primary afferent terminals are all subject to regulation. Consequently, synergists do not act independently, but are coordinated to regulate the properties of the joint through interneuronal circuitry (Krakauer and Ghez 2000). A muscle is able to perform more than one action due to the variation in its attachment points. The contribution of the muscle is directly dependent on both the length of the moment arm and the force produced by the muscle fibres, which would both vary throughout the range of motion.

Joint stability is directly influenced by the neural activation of the antagonist and agonists. Normally the antagonist contracts eccentrically while at the same time the agonist is contracting concentrically as a prime mover. The exact nature of this relationship during stabilization is not yet clear. Co-activation of the antagonist and agonist enhances joint stability by making perturbation more difficult. Appropriate muscle activation would also be required for continuous orientation of the humerus on the glenoid. The appropriate muscle activation is achieved through the divergent interneuronal connections of the spindle afferents and the reciprocal innervation of the descending axon (Loeb and Ghez, 2000).

The shoulder muscles are an excellent example of the use of dynamic stabilization to maintain desired orientation of the upper arm during movement of the hand. There is a need for increased proximal stability in order to maintain stability distally at the hand (McIntyre, Mussa-Ivaldi and Bizzi 1996). Conversely, manipulation of the distal segment (hand) will have direct impact on the proximal joint stabilization requirements. EMG activity has been demonstrated in the rotator cuff muscles during a precision hand movement task, confirming their role in achieving proximal stability (Laursen et al. 1998).

The success of joint stabilization is dependent on the complex integration of afferent and efferent information both at the spinal and supraspinal levels in order to control the balance of muscle forces required at the shoulder joint. The challenge then becomes how to appropriately measure shoulder joint stabilization. Without an accurate measurement of sufficient joint stability, a discussion of insufficient joint stability or instability cannot be entertained. Numerous investigators have measured the strength ratios of the joint stabilizers at the shoulder, to discern a possible relationship between strength balance and effective dynamic joint stability.

Dynamic Strength Ratios

Appropriate balances of the dynamic stabilizer muscle forces are necessary to maintain normal shoulder kinematics. It is possible that an imbalance of these muscles may contribute to injury especially in high velocity movements (Hughes et al. 1999). An estimate of these forces is provided through the measurement of shoulder muscle strength. Factors such as age, gender, hand dominance and specific training are all known to significantly influence shoulder muscle strength (Hughes et al. 1999, Timm 1997, Wilk, Andrews and Arrigo 1995). Several investigators have examined the strength of the dynamic stabilizers at the shoulder, specifically, the internal and external rotators (Cook, et al. 1999, Timm 1997, Wilke et al. 1995, Wilk, Andrews, Arrigo, Keirns and Erber 1993). The "normal" external (ER)/internal (IR) rotation strength ratio has been reported as 65% by Wilk et al. (1993) and Cook et al. (1987) in healthy subjects

Hughes et al. (1999) looked at normative values of isometric agonist and antagonist internal and external rotator shoulder strength ratios in 120 male and female subjects aged 20 to 78 years old. External/internal isometric strength ratios increased from 0.60 with the arm in neutral rotation and 15° abduction to 0.73 with the arm in 30° external rotation and 90° abduction. Change in strength ratios with arm position change is expected based on the length tension properties of the muscle that affect force production. External/internal ratios also increased with age with the arm in 90° abduction from 0.66 for ages 20-29, to 0.79 for subjects over 60. Shoulder strength is generally found to decrease with age (Kuhlman, Iannotti, Kelly, Riegler, Gevaert and Ergin 1992). An increase in the ER/IR ratio with age may be indicative of a greater decrease in internal rotator strength due to changing upper limb activity patterns with age. There was no significant relationship between dominance or gender and strength ratios. Their reported results were very similar to dynamometer data on young athletes reported by Cook et al. (1987), Wilk et al. (1995), and Hinton (1988).

Kramer and Ng (1996) examined ER/IR ratios and peak moments in 40 subjects aged 45-75 years old. The isometric external/internal ratio in the mid position of rotation was 0.80. Ratios for concentric and eccentric contractions ranged from 0.82 to 0.88, with a lower ratio obtained during higher speed (120 °/s) contractions. Warner et al. (1990) previously reported similar results for concentric IR/ER ratios in normal subjects of 1.2-1.5.

Scoville, Arciero, Taylor and Stoneman (1997) reported on the ratio of end range eccentric antagonist/concentric agonist strength ratios in active young males ages 17-21 years old. The end range is the range where the antagonist is firing eccentrically to decelerate the concentric motion of the agonist. Isovelocity testing was performed at 90°/second at 90° of abduction. The eccentric antagonist/concentric agonist ratio found at end range medial rotation was 1.08:1 (lateral rotators/medial rotators) and at the end range lateral rotation was 2.39:1 (medial rotators/lateral rotators). A greater strength of the muscle responsible for deceleration was expected in order to overcome and decelerate the segment in motion. The large difference found at the end range of lateral rotation suggests that a deceleration force greater than double that of the concentric muscle force may be needed to maintain dynamic stability of the glenohumeral joint. The values at the end range of medial rotation suggest that the lateral rotators need to be as strong as the medial rotators to maintain shoulder stability. This ratio may increase if the testing speed increased: concentric strength values would decrease with increasing speed, but eccentric values would remain the same. The generalization of these results is limited to young males who are active in upper limb activities.

Several studies have found strength imbalances in certain athletic populations (baseball, tennis and swimming) and in subjects with impingement syndrome (Cook et al 1987, Ellenbecker 1991, Hinton 1988 and Wilk et al 1993 and 1995). De Pauw and

Kriellaars (1996) examined dynamic control ratios for eccentric external rotation and concentric internal rotation in both healthy and impingement subjects. Dynamic control ratio is calculated using the peak eccentric external rotation moment and the peak concentric internal rotation moment. Figure 1 illustrates the strength discrepancy of the internal and external rotators between control and impingement subjects at several angles and velocities of testing. Note that there was a significant difference between the two groups for the external rotators, but not the internal rotators, resulting in a decrease in the ER/IR ratio for the impingement group.

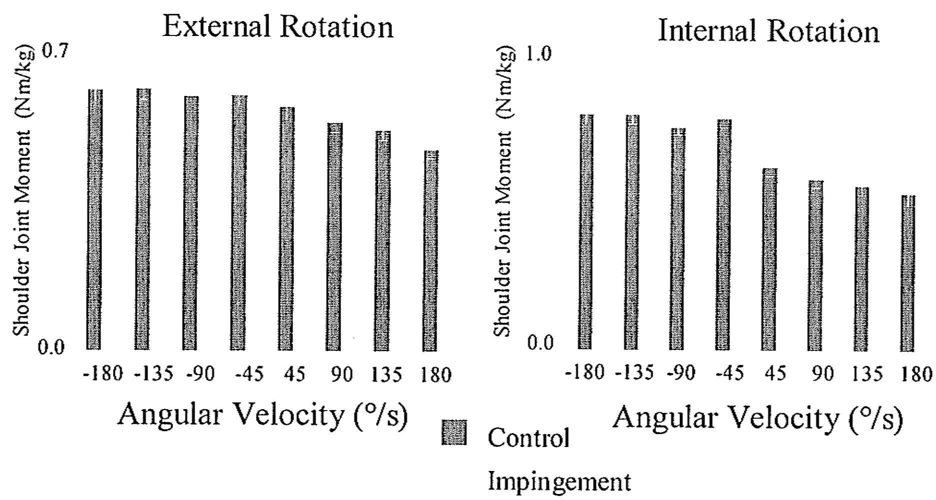

Figure 1: Comparison of internal and external rotator strength. Comparison of peak shoulder joint moment (Nm/kg) between those in a control group (dark) and an impingement group (light) for both internal and external rotation at angular velocities between –180°/sec and 180°/sec. Adapted from DePauw and Kriellaars (1996).

Figure 2 demonstrates these differences in the DCR between the impingement and control groups. The impingement group displays a significantly lower ER/IR ratio (0.83) than the control group (1.00). This results directly from the significant external rotation strength deficits seen in the impingement group.

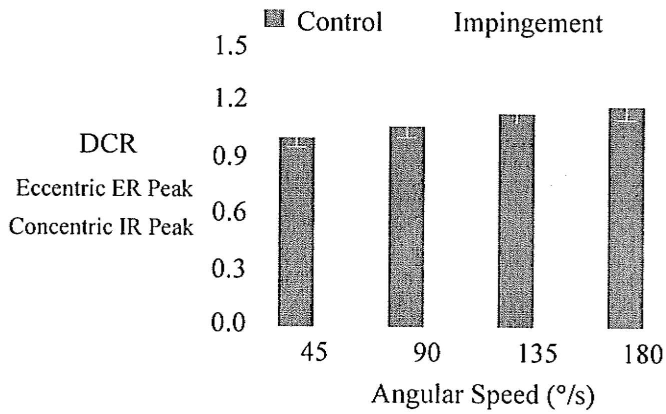

Figure 2: Glenohumeral Dynamic Control Ratios (DCR) for control and impingement subjects. The ratio of eccentric external rotation peak moment is divided by the concentric internal rotation peak moment to derive the DCR. The DCR of the control group (dark) is compared to the DCR of the impingement group (light). Adapted from DePauw and Kriellaars (1996).

The effect of fatigue on the strength of the internal and external rotators has been measured by only a few authors (Beach, Whitney and Hoffman 1992, Chandler, Kebler, Stracener and Ziegler 1992, Ellenbecker and Roetert 1999, and Falkel and Murphy 1988) and only in athletic populations such as swimmers and tennis players. The most recent study by Ellenbecker and Roetert (1999) examined external and internal rotator muscle strength fatigue ratios in 72 elite junior tennis players aged 12-18. Concentric internal and external rotation were strength tested using a dynamometer in 90° of glenohumeral abduction in the coronal plane in order to approximate the angle used in tennis overhead movement patterns. The relative fatigue ratio (work in the last 10 repetitions divided by the work in the first 10 repetitions) was calculated for the dominant and non-dominant arm. The internal rotators were more fatigue resistant than the external rotators according to their results. The mean relative fatigue ratio for internal rotation was 83.42 but for external rotation was 70.13. Surprisingly, there was no significant difference in fatigue ratios between the dominant and non-dominant arm. The authors concluded that greater fatigue resistance of the internal rotators would produce an imbalance of the ER/IR ratio. This could affect the stabilizing influence of the external rotators on the anterior glenohumeral ligaments and capsule to prevent excessive anterior translation, and on the caudal force vector produced by the internal and external rotators to prevent excess superior translation. Although the concept of differential fatigability is alluded to in this article, it does not examine the rotators in their role of stabilizers, only as prime movers.

Results from these studies indicate that strength ratios are highly individualized, making the development of normative data difficult. Secondly, these ratios can be affected by athletic performance, shoulder dysfunction and fatigue. Finally, alteration of this ratio by means of deconditioning or fatigue could certainly affect the degree of dynamic control afforded by the internal and external rotators.

D. Neuromuscular Fatigue

The processes that underlie fatigue begin at the onset of any task even if a decrease in force is not yet detectable (Gandeiva 2000). Enoka (1994) defines fatigue as simply "a class of acute effects that impairs motor performance". Fatigue begins rapidly with maximum voluntary concentric contractions, and gradually with sub-maximal force. Several variables contribute to the production of fatigue including motivation, neural strategy (pattern of muscle activation), intensity and duration of the task, speed of contraction required, and the sustaining nature of the activity (Enoka 1995).

Peripheral Factors

Peripheral fatigue occurs distal to the motor axon, primarily in the muscle. Peripheral fatigue is a failure in the transmission of the neural signal or a failure of the muscle to respond to neural excitation (Sieck and Prakash 1995), resulting in a decrease in force production. Components that are potentially implicated include inadequate neurotransmitter release, failure of action potentials both along the axon (pre-synaptic) and the sarcolemma (post-synaptic), and failure of excitation-contraction coupling. The exact mechanism for failure of signal transmission is not known. Ischaemia from increased intramuscular pressure is known to exacerbate this impairment (Bigland-Ritchie, Rice, Garland and Walsh 1995). The tendency of a muscle to develop high intramuscular pressure during an increase in muscle force varies with the architecture of the muscle, the type of contraction and the intensity of the developed force (Bigland-Ritchie et al. 1999, Gandevia, Enoka, McComas Stuart & Thomas 1995, Sjogaard and Sogaard 1998). Fatigue is highly specific to the muscle fibre type due to mechanical, biochemical and metabolic properties of the fibres that determine its ability to generate and sustain force .

Failure of excitation-contraction coupling involves a sequence of events that includes propagation of the action potential along the sarcolemma and into the T tubules, the release of $Ca^{2+}$from the sarcoplasmic reticulum, Ca^{2+} binding and activation of the contractile apparatus. Reduced maximal muscle force is caused initially by a reduced capacity of the myofibrils to produce force even in the presence of Ca^{2+} saturation. This is most likely due to an increase in phosphate concentration, not to an increase in H+ concentration (acidosis) as previously thought (Sahlin, Tonkonogi and Soderland 1998, Westerblad, Allen, Bruton, Andrade and Lannergen 1998). This is followed by a decline in force due to lowered Ca^{2+} release from the sarcoplasmic reticulum, which reduces the number of activated cross bridges (Westerblad et al. 1998). The net effect of a change in these events with intense muscle fibre activity is a reduction maximal muscle force and power (Taylor, Butler and Gandevia 2000). The relative contributions of these components to muscle fatigue remains controversial.

The firing rates of active motoneurons decrease during a sustained isometric MVC in response to a slowing of relaxation rates (Sawczuk, Powers and Binder 1995). Additional motor units are recruited to compensate for this. Decreased motoneuron firing rates are likely due to a variety of factors including: the intrinsic properties of the motoneuron, reflex inhibition and disfacilitation of the Ia afferents, Renshaw cell inhibition and insufficient drive from supraspinal centres (Gandeiva 1998). Decreased motoneuron firing rate optimises force production and avoids premature peripheral fatigue during maximal isometric contractions (Binder-Macleod 1995, Sawczuk et al. 1995). This concept is known as muscle wisdom (Binder-Macleod 1995). However, muscle wisdom has not yet been confirmed in sub-maximal or dynamic contractions (Bigland-Ritchie et al. 1995). Optimisation of force production may also occur with motoneuron cycling. Kriellaars and Massey (1996) examined motoneuron recruitment during repetitive concentric contractions and found that individual motor units fired for 5 repetitions, were silent for 5 repetitions and then fired again. This was termed the load-sharing hypothesis.

Central Factors

Central fatigue is defined as a progressive exercise-induced reduction in voluntary activation of the muscle (Gandevia et al. 1995). Central fatigue occurs proximal to the motor axon and results in an increased sense of effort and a decreased neural drive to the muscle. The sense of effort is a major factor influencing motivation (Bigland-Ritchie et al. 1995). Other factors such as the levels of circulating hormones that control motivation and pain tolerance, the descending drive to motoneurons and interneurons, recruitment of initially uninvolved motor units, and the resistance of the muscle to maximal activation also play a role in the development of central fatigue (Gandevia 1998, Enoka 1995). Central fatigue occurs with both maximal and sub-maximal exercise. Central fatigue can be generated by both spinal, motoneuron and supraspinal factors involving impaired firing rate modulation or reduced motor unit recruitment (Miller, Kent-Braun, Sharma and Weiner 1995). Studies have established a reduction in corticospinal discharge or drive with fatigue that results in a reduction in the rate of motoneuron firing, however, very little is actually known about the reason for the insufficient motor cortical drive (Gandevia 1998, Gandevia et al. 1995). Different types of intensity, duration, environment and task affect the influence of the variables on the development of fatigue.

The role of afferent feedback

Sensory feedback contributes to the optimisation of force output during muscle fatigue through the modification of spinal circuits. The control of motoneuron firing and excitability during fatigue depends not only on descending modulation and alteration of motoneuron excitability, but on the complex influence of afferent feedback and recurrent inhibition both directly on the alpha and gamma motoneurons and on various interneurons as well (Gandeiva 2000, Windhorst and Boorman 1995). Mechanisms by which this feedback system may be influenced during fatigue include adaptation with repetitive firing, activation of the fusimotor system, presynaptic inhibition through interneurons, and changes in the characteristics of the muscles such as temperature and metabolite concentration.

Feedback from group Ia afferents is known to influence motoneuron-firing rates in non-fatigued muscles. The precise effect of afferent feedback on motoneuron discharge during fatigue is not yet known. The discharge rate of muscle spindle afferents decreases during sustained isometric contractions, which could result in disfaciliation of the motoneuron (Pearson and Gordon 2000). However, when afferent feedback conduction is blocked during fatigue, decreased motoneuron firing rates were not observed (Gandevia 1998). Disfacilitation of the motoneurons from afferent feedback could be compensated by spinal or supraspinal modulation.

Firing rates of non-spindle group II, group III and IV afferents responding to chemical or mechanical stimuli increases during sustained sub-maximal isometric contractions (Pearson and Gordon 2000). These nerve endings are quite numerous, consequently a small change in their discharge results in a massive increased input to the central nervous system (Pearson and Gordon 2000). However, the precise effects of fatigue on afferent feedback during concentric and eccentric contractions and the method by which they alter motor control during fatigue is still unknown (Gandeiva 2000, Pearson and Gordon 2000) and is likely quite different during non-isometric contractions. Afferent feedback will affect not only the homonymous and synergist muscles, but also the antagonist muscles through reciprocal inhibition (Windhorst and Boorman 1995).

Several studies have examined the effects of fatigue on joint position sense and movement sense at the shoulder and the knee and all report diminished proprioception occurring with fatigue (Bjorklund, Crenshaw, Djupsjobacka and Johansson 2000, Carpenter, Blasier and Pelizzon 1998, Lattanzio, Petrella, Aproule and Fowler 1997, Pederson, Lonn, Hellstrom, Djupsjobacka, and Johansson 1999, Sterner, Pincivero and Lephart 1998). All these studies required the conscious awareness of the subject for detection of movement, which is not exactly the same as proprioception. Nevertheless, they still provide evidence for disturbed afferent feedback with fatigue that may impair the neuromuscular coordination of joint stability.

Neural Strategies to Delay Fatigue

The onset of fatigue can be delayed by the employment of several neural strategies including the variation in recruitment of motor units. The complex interaction of neural connections can modify the excitability of groups of motoneurons during conditions such as fatigue (Enoka and Stuart 1992, Windhorst and Boorman 1995). These modifications can occur both within the homonymous muscle and between synergist and antagonist muscle groups. The contribution of synergist and antagonist muscles to sub-maximal muscle force can be varied without affecting the net output of force from the muscle. During prolonged contractions, several studies have reported alternating motor unit activity in the homonymous muscle (Fallentin, Jorgenson and Simonson 1993, Jensen, Pilegaard and Sjogaard 2000, Sjogaard, Kiens, Jorgensen and Saltin 1986, Tamaki et al. 1998). Others claim that certain slow twitch motor units may maintain continuous activity during low-level prolonged contraction that may overload the metabolic capacity of the fibres (Schnoz, Laubli and Kreuger 2000).

Tamaki et al. (1998) clearly demonstrated alternating recruitment strategies during prolonged sub-maximal contractions in a synergistic group of muscles; the triceps surae (lateral gastrocnemius, medial gastrocnemius, and soleus). Surface EMG recordings were used to assess static, concentric and eccentric contractions at a workload that corresponded to 10% of maximal voluntary contraction (MVC). EMG analysis indicated alternating activity among all three synergistic muscles over the 210-minute time course of both static and dynamic tasks. This alternating activity was seen to increase significantly in the second half of both the static and dynamic tasks. The contribution of motor unit activity varied with the joint angle, confirming the importance of muscle fibre length and moment arm on muscle force output. Motor unit activity was rotated in a complementary pattern in order to maintain a constant force output from the muscle. Examination of homonymous muscle EMG, demonstrates a significant increase in EMG up to the middle period of the task only. This suggests that the strategy of alternating activity in the synergist muscles may attenuate the level of recruitment necessary in the homonymous muscle. Jensen et al. (2000) supports this indirectly with their examination of prolonged low-level static contraction of supraspinatus. Their results confirmed variable recruitment of motor units, but a lower than expected degree of motor unit rotation with fatigue. This may be due to alternating motor unit activitation with a synergistic muscle (deltoid), which was not examined.

Pincivero et al. (2000) examined the effects of fatigue on quadriceps-hamstring co-activation during a forward lunge. EMG analysis of the vastus lateralis, vastus medialis, biceps femoris and semitendinosis was performed on ten subjects during a repetitive standardized lunge task to volitional fatigue. The results of the study indicate a significant increase in EMG activation of all recorded muscles as the task progressed, except for semitendinosis. More specifically, during the down phase of the lunge, there was a significant difference between the increase in activation of the quadriceps and biceps femoris muscles. Interpretation of these results must be done with caution. Firstly, the quadriceps muscles are functioning eccentrically and the hamstrings concentrically during the downward lunge. The physiological requirements of these contractions are different. Secondly, the hamstrings muscles are performing as biarticular muscles during the downward phase, but the vastus medialis and vastus lateralis are only monoarticular muscles. The biomechanical requirements of these muscle groups are quite different.

Although these studies were both performed on the weight-bearing lower limb, extrapolations can be made to include the shoulder musculature during static and dynamic contractions. The precise type of contraction (eccentric/concentric) that the internal and external rotators of the shoulder are undergoing during stabilization is not known. However, it is likely that both types of contractions occur for each muscle at various times depending on the position of the humeral head. Also, the internal and external rotators are both monoarticular muscles.

Two important concepts emerge from a review of the effects of fatigue on muscle control. Firstly, a reduction in force output of the muscle will eventually occur with sustained sub-maximal contractions. This force reduction will vary depending on the characteristics of the muscle involved and the type of contraction it is involved in (concentric or eccentric). This may result in an alteration of dynamic joint stabilization by disturbing the optimal dynamic control ratio. Secondly, the central nervous system employs various strategies to delay the onset of muscle fatigue and maintain optimal muscular performance. Specific training of the involved muscles may optimise the neural strategies and delay or eliminate any alteration in dynamic stabilization at that joint or limb segment.

Consideration of the effects of fatigue on dynamic joint stabilization during a multi-segmental task must take into account all segments involved. During a gripping task, consideration must be given to the distal segment and forearm muscles as prime movers, the elbow joint as the intermediary buffering component, and the shoulder muscles as the stabilizers. A variety of different strategies can be employed for each of these components to facilitate optimal performance and minimize the development of fatigue. The demands ultimately placed on the shoulder stabilizers during the task may then depend on the optimal employment of strategies by the central nervous system for the each of the segments.

E. Shoulder Instability

The role of the glenohumeral musculature in dynamic joint stabilization has been confirmed in the literature. The contribution of dynamic stabilization to glenohumeral pathology is not as clear. Superior migration of the humeral head is one mechanism by which inadequate stabilization could trigger rotator cuff disorders and impingement syndrome. A brief review of rotator cuff disease, impingement syndrome and research on fatigue during upper limb tasks is included to complete the review of literature.

Migration of the Humeral Head

The humeral head translates superiorly in relation to the glenoid fossa during abduction due to contraction of the deltoid muscle (Inman et al. 1944, Poppen and Walker 1976) and the effects of the joint capsule (Harryman et al. 1990). The force couple created by the downward force of the rotator cuff muscles opposes the tendency towards superior translation. Without a functioning rotator cuff, either from fatigue or pathology, increased superior translation of the humeral head during abduction may occur (Sharkey and Marder 1995).

The width of the subacromial space may play an important role in the pathogenesis of impingement syndrome. It is hypothesized that a reduction in this width can cause impingement of the subacromial structures. One way to measure this width is to measure the acromio-humeral distance. The distance can range from 6 – 14 mm on radiographs in healthy subjects in the neutral position. Studies have shown that approximately 50% of those with rotator cuff disease will have a reduced acromio-humeral distance.

Graichen, Bonel, Stammberger, Haubner, Rohrer, et al. (1999) used 3D open MR imaging techniques to study this distance in 10 healthy subjects and 10 with clinically confirmed impingement syndrome. The subjects were examined at 60°, 90°, and 120° of abduction both during muscle relaxation and during electromyographic stimulation of the middle and anterior portions of the deltoid muscle (at 90°only). The mean acromio-humeral distance decreased at each position tested without muscle activity, but only decreased significantly from 90 to 120° of abduction (decreasing distance was partially caused by the greater tuberosity approaching the acromion). When the deltoid muscle was activated, the only position for which the distance decreased significantly was at 60° of abduction (mean, -32%), and the distance remained almost constant for all three positions tested with the deltoid activated (4.7mm @60°, 4.1mm @ 90°, and 4.8mm @120°). At 120° of abduction, the distance was actually increased. When compared to the subjects with impingement syndrome (n=6), there was no significant difference for any position with the muscles relaxed, but when the deltoid was activated, the distance was significantly smaller in the impingement group (1.4mm versus 4.4mm). Patients with full thickness rotator cuff tears (n=3) had a reduction in the subacromial width at 30° and 90° with and without muscle activity. Yamaguchi, Sher, Anderson, Garretson, Uribe, Hechtman, and Neviaser (2000) reported increased superior humeral head migration during elevation in subjects with rotator cuff tears compared to normal subjects using scapular x-ray films.

Chen, Simonian, Wickiewicz, Otis and Warren (1999) examined the effects of neuromuscular fatigue on glenohumeral joint kinematics using radiographic analysis. Twelve male subjects with no subjective shoulder problems were radio graphed at four angles of abduction between 0 and 135 degrees. To produce fatigue, subjects were instructed to abduct the non-dominant arm in the prone position to 100° in the plane of the scapula repeatedly until unable to reach a 45° angle of abduction 3 times in a row (this correlated to a 30% decrease in deltoid strength on the isokinetic dynamometer). A dumbbell weight equivalent to 5% of their body weight was used. The authors chose the non-dominant arm on the assumption that fatigue would be easier to produce. This limits generalizations to dominant arm function, which would be involved in the majority of upper limb tasks.

Anteroposterior radiographs were taken in the plane of the scapula before and immediately after the shoulder was fatigued with the subject holding a 1 kg. mass in their hand and maintaining the shoulder in a neutral position. Radiographic analysis was performed by measuring the humeral head position in relation to the glenoid equator (bisection of a line connecting the superior and inferior edges of the glenoid rim) with a metric calliper along the horizontal axis on each radiograph.

The data was analysed using two-tailed paired t tests with an alpha level of 0.05 and the results were presented graphically as illustrated in Figure 3. Lines connecting the four angles leave the impression that the humeral head travelled that exact course between those angles, assuming linear interpolation. However, rather than following a linear path from 1 to 45°, the humeral head may have immediately migrated up to its next position. By choosing a concentric/eccentric exercise task to fatigue the upper limb, but a static condition to assess humeral head position, a discrepancy was created. Since the humeral head was only measured at four angles, it is incorrect to assume the path it followed between those angles. Cine-x-ray analysis would have provided dynamic assessment and provided further information about the exact path the humeral head travelled.

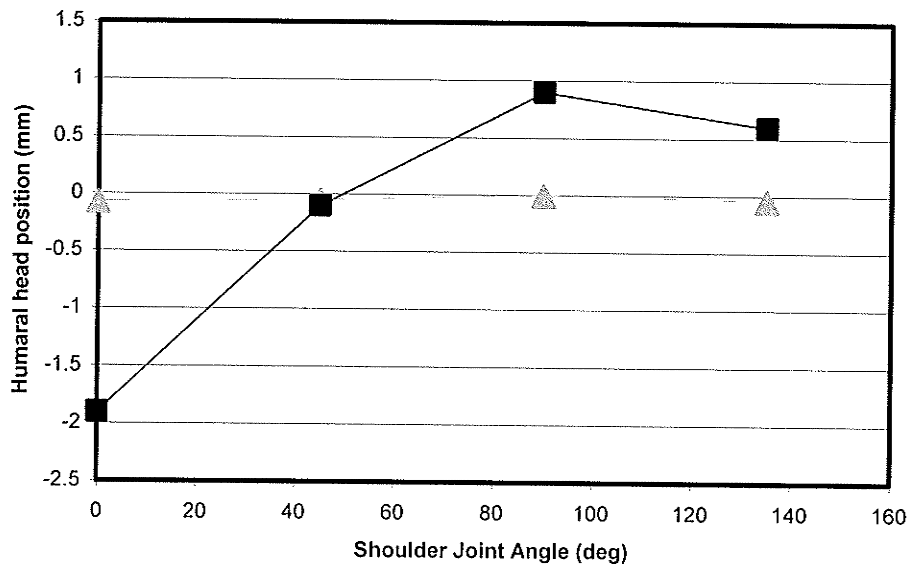

Figure 3: Migration of the humeral head (mm) before (triangles) and after (squares) fatigue. Adapted from Chen et al. 1999.

The results showed a significant difference in the migration of the humeral head at 45, 90 and 135° of abduction after fatigue. Before fatigue the position of the humeral head was below the centre of the glenoid for all angles of abduction. The position remained fairly constant through all positions, (see Figure 3) but did migrate by up to 0.3 mm (± 0.4 mm). After fatigue, the humeral head migrated superiorly by an average of 2.5 mm at all four intervals tested. The greatest migration above the horizontal occurred at 90° of abduction. This may be directly related to the increasing activity of the deltoid and supraspinatus muscles between 90° and 180°, which exert an upward pull when contracting.

Interestingly, when fatigued, the humeral head was actually situated inferior to the horizontal axis of the glenoid by a statistically significant average of 1.2 mm at 0 degrees of abduction. Requiring the subjects to hold a 1 kg. mass in their hand during the radiograph may have introduced a confounding variable. One possible explanation for the inferior migration of the humerus at 0 degrees is a decrease in drive from the central nervous system to the deltoid muscle after fatigue. This strategy could allow quicker recovery of the transmission pathways. If this was the case, then upon activation of the deltoid muscle the humeral head position should be immediately restored. Knowledge of the precise positions of the humeral head from 0 to 45 degrees would confirm this hypothesis. Without that information, explanations are only speculative. Further understanding of these finding may be important in certain pathologies of the shoulder such as inferior instability or subluxations.

The authors conclude that fatigue of the rotator cuff alone, in the absence of any pathology, causes abnormal glenohumeral kinematics. Fatigue of the rotator cuff during this study can only be inferred. Only the prime mover (deltoid) was actually measured. Normally the rotator cuff muscles counteract the upward pull of the deltoid muscle and together centre the humeral head on the glenoid. This is confirmed by the findings of this study showing no significant migration of the humeral head before deltoid fatigue. The ability to maintain the humeral head in a centred position was compromised when the deltoid was fatigued; the humeral head migrated superiorly, exposing the rotator cuff tendons to compression in the subacromial space. Clinically this could translate to tendonitis, and partial or complete tears of these tendons.

A second mechanism for damage that is not mentioned by the authors is by impingement of the greater tuberosity and its attached rotator cuff against the superior glenoid labrum. (Jobe 1996) Normally during abduction the axis of the humeral head angulates posteriorly and superiorly and the head rotates superiorly. The primary passive stabilizer restricting this motion is the inferior glenohumeral ligament. The subscapularis muscle is acting as a dynamic stabilizer. Therefore, as subscapularis becomes fatigued hyperangulation of the humeral head could occur, increasing contact of the greater tuberosity on the glenoid labrum. Increased time and frequency of this contact could lead to pathology. Further cine-x-ray analysis should assess angulation and superior migration of the humeral head after fatigue for a complete picture of pathological kinematics. Further research should also include examination of the dominant arm in both males and females, and recovery time.

The findings from this study have direct implications for our proposed research. We hypothesize that differential fatigue of the rotator cuff muscles (infraspinatus, teres minor, subscapularis and supraspinatus) will occur with a task of the upper limb where they are required to act as stabilizers and not prime movers. Firstly, the results allude to differential fatigue. The rotator cuff muscles may be fatiguing at a different rate than the deltoid during this task; disrupting the force couple, and disturbing the ability of the dynamic stabilizers to centre the humerus on the glenoid. Secondly, from these results we can assume that as fatigue of the rotator cuff muscles occur, the humeral head may be migrating superiorly. It is possible that it may also be increasing its contact with the superior glenoid labrum. This offers an explanation for injury to the supraspinatus tendon with repetitive motion tasks of the upper limb, particularly with the limb positioned at 90 degrees of abduction. Although we will be studying only one specific upper limb task, the same principle can be applied to numerous other tasks and may implicate many common industrial activities.

Rotator Cuff Disorder

Rotator cuff disorders are the most common cause of shoulder pain and dysfunction (Cohen and Williams 1998, Matsen and Arntz 1990b, Soslowsky et al. 1997a). They are pervasive in sports (especially those involving repetitive overhead motions), occupations, and in the elderly population (Matsen and Arntz 1990b). The incidence of rotator cuff pathology increases with age (Blevins, Djurasovic, Flatow and Vogel 1997, Cohen and Williams 1998, Neer 1983) particularly beyond age 40, with 39% of individuals older than age 60 developing the disorder (Bigliani, Ticker, Flatow, Soslowsky, and Mow 1991). Only 8% of tears are acute (Matsen and Arntz 1990b). According to Neer, in individuals under the age of 40 it is more common with significant trauma to have a fracture or an injury to the glenohumeral ligament before a tear of the rotator cuff occurs. In his study of 233 cadavers with rotator cuff tears, only 8 specimens were under 40 (Neer 1983). The rotator cuff's function can fail before it develops a full-substance tear (Uhthoff and Sano 1997). Partial thickness tears occur approximately twice as often as full thickness tears and most commonly occur on the articular surface within 1 centimetre of their insertion (Soslowsky et al. 1997a). The precise aetiology of rotator cuff injury is unknown. The possible underlying causes of rotator cuff disorder can be classified as either extrinsic or intrinsic.

Extrinsic mechanisms of tendon failure include macro trauma, repetitive micro trauma, kinematic abnormalities and impingement of bony or soft tissue structures. Acute macro trauma is an uncommon mechanism of rotator cuff tear, although the incidence does increase in individuals over 40 years of age, which may be related to degenerative changes (Blevins et al. 1997). Repetitive micro trauma may lead to degenerative changes or microscopic failure within the tendon, which can progress, to partial or full-thickness tears. Although there is potential for some repair of the tendons, the extent of healing is unclear and decreases in individuals over age 50 (Blevins et al. 1997). Compression or impingement of the tendons can occur as they travel under the coroacoacromial arch (Curl and Warren 1996, Matsen and Arntz 1990b, Uhthoff and Sano 1997). According to Neer (1983), 95% of all rotator cuff tears are related to impingement. An increased incidence of rotator cuff disease in subjects with curved or hooked acromions (secondary to enthesopathy from calcification along the coracoacromial ligament), distal osteophytes on the clavicle, or increased tension or cross-sectional area in the coroacoacromial ligament provides evidence for this (Curl and Warren 1996, Neer 1983, Soslowsky et al. 1997a, Uhthoff and Sano 1997).

Intrinsic rotator cuff disorders are those that arise from forces within the rotator cuff muscle–tendon complex (Fu, Harner and Klein 1991). The aging of the rotator cuff is a progressive process characterized by disruption of the integrity of the tendon attachment and a decrease in its cellularity and vascularity, most often in the supraspinatus tendon (Nirschl 1989). According to Uhthoff and Sano (1997), degenerative processes constitute the most frequent cause of rotator cuff disorder. Degeneration of the rotator cuff begins with changes in cell arrangement, granularity, disruption and thinning of fascicles, and calcifications that decrease the tensile strength and elasticity of the tendon (Uhthoff and Sano 1997). This may lead to acute episodes of shoulder pain from fibre failure and tendonitis with repeated episodes of fibre failure resulting in weakness of the tendon and increased vulnerability. This could progress to a partial or full thickness tear of the rotator cuff tendon.

Soslowsky et al. (1997a) examined 17 surgical patients with confirmed supraspinatus tendon tears (5 also had infraspinatus tears) aged 32-66 and discovered the same basic pattern of aging in all tendons. There was a characteristic loss of the wavy pattern of collagen fibres, hyalinization, microtearing of the tendon fibres, and poor vascularity in the region of all the tears with no evidence of any repair. Contrary to previous observations, they did not find any spurs on the acromioclavicular joint or on the acromion on any of the patients.

The pathogenesis of rotator cuff disorder is multifactorial including both extrinsic and intrinsic factors. Weakening of the tendon likely begins before tendon failure occurs. However, little information is available regarding the relative significance of each component or the interplay between them. Abnormal glenohumeral kinematics during upper limb movement, leading to rotator cuff impingement, represents one possible mechanism for the development of rotator cuff tendon pathology.

Impingement Syndrome

Impingement is defined as the encroachment of the acromion, coracoacromial ligament, coracoid process, and the acromioclavicular joint upon the supraspinatus, infraspinatus and long head of biceps tendons as they pass beneath these structures during glenohumeral joint is movement. Compression can also occur against the posterior superior glenoid rim and the convexity of the humeral head in certain arm positions (Blevins et al. 1997, Jobe 1996). Non-pathologic contact between these structures likely occurs regularly, however increased contact time or force could change this to pathologic (Jobe 1996).

Shoulder impingement syndrome however is not as clearly defined and can indicate a variety of disorders that manifest as anterior shoulder pain, particularly during overhead activities (Fu et al. 1991, Jobe 1996). The incidence of impingement syndrome increases with age (Cohen and Williams 1998). In the position of 90° abduction and maximal external rotation, the posterior supraspinatus tendon and the anterior infraspinatus tendon can be pinched between the humeral head and the posterior superior glenoid rim (Jobe 1996). The pressure in the subacromial space increases over five times the resting value in elevation through flexion of the glenohumeral joint and is accentuated further with a 1 kilogram mass held in the hand (Fu et al. 1991).

The width of the subacromial space may play an important role in the pathogenesis of impingement syndrome. It is hypothesized that a reduction in this width can cause pathological impingement of the subacromial structures (Fu et al. 1991, Matsen and Arntz 1990a).

There are numerous factors that may increase the likelihood of impingement occurring. Structurally, the acromioclavicular joint, acromion or coracoid process may have congenital abnormalities, spurs may develop on their under surface, or the greater tuberosity of the humerus may be prominent (Fu et al. 1991). The subacromial bursa may become inflamed or thickened from previous injury or inflammation. The rotator cuff tendons can become thickened from calcium deposits or scarring (Burns and Whipple 1993). Functionally, the scapula may develop an abnormal position from kyphotic posture or acromioclavcular joint separations, or it may develop dys-rhythmic motion of the acromion and the humerus from joint stiffness, muscle weakness or fatigue. Tightness of the posterior capsule, anterior capsular laxity, rotator cuff weakness or fatigue and tearing or rupture of the biceps tendon may increase humeral head translation in a superior direction (Cohen and Williams 1998, Matsen and Arntz 1990a). Neer attributed 95% of all rotator cuff tears to impingement of the tendons against the acromion (Neer 1983).

Extrinsic components such as repetitive motion may also contribute to impingement (Jobe 1996). This can include occupational and exercise related activities that incorporate factors such as arm position, number of repetitions, and weight-lifting requirements. The precise role of these factors has not yet been clearly defined in the literature. The literature suggests that fatigue of the rotator cuff muscles may increase impingement either through increased humeral head migration (Chen et al. 1999) or increased contact of the greater tuberosity against the glenoid (Jobe 1996). Humeral head migration can become problematic when it is inappropriate and occurs either periodically leading to cumulative trauma and subsequent tissue failure, or as an isolated event leading to failure. A better understanding of the magnitude of fatigue of the dynamic stabilizers during an upper limb task will contribute to a better understanding of the factors involved in both impingement and disease of the rotator cuff tendons.

The effects of upper limb tasks on the shoulder.

The incidence of musculoskeletal pain in the shoulders and upper limbs related to repetitive tasks is second only low back pain (Wiker, Chaffin and Langolf 1989).

Prevalence of shoulder impairment is reported as 8.8% and shoulder pain at 80%, (Makela, Heliovaara, Sainio, Knekt, Impivaara and Aromaa 1999). The demands upon the shoulder with hand movement vary with the posture of the upper limb, the external forces placed upon the arm and the anatomical differences between individuals (Inman et al. 1944, Wiker et al. 1989).

Numerous studies have been done to examine the effects of repetitive upper limb tasks on upper quadrant muscle fatigue (McQuade, Dawson and Smidt 1998, Wiker et al. 1989, Schnoz et al. 2000, Sjogaard and Sogaard 1998). The majority of these used EMG recordings from the scapulothoracic or deltoid muscles as indicators of shoulder muscle fatigue, but these studies have generally had large variability and the majority use small sample sizes.

Sjogaard and Sogaard (1998) confirmed two important factors in repetitive motion using kinematic analysis: (1) acceleration of the limb, especially with added weight increases the force demands of the task, and (2) varying the velocity of movements may change recruitment patterns and delay the onset of fatigue. These findings have direct implications to many repetitive tasks.

Intramuscular pressure levels high enough to disturb blood flow can occur during sustained low level intensity contractions in certain muscles including the supraspinatus and infraspinatus (Palmerud, Fosrsman, Sporrong, Herberts and Kadefors 2000, Sjogaard and Sogaard 1998, Wiker et al. 1989). Intramuscular pressure in the supraspinatus and infraspinatus is related both to elevation of the arm (maximum at 90°) and the external load (Palmerud et al. 2000).

Wiker et al. (1989) studied the effects of arm position, load and work to rest ratios on fatigue of the upper trapezius and deltoid muscles in four subjects. EMG recordings, tremor analysis and discomfort were used to assess fatigue in these muscles. Their findings indicate that an increase in fatigue is related to an increase in the mass carried in the hand, a higher work to rest interval, and an increase in moment about the shoulder.

McQuade et al (1998) examined the effects of fatigue on scapulohumeral rhythm. Three scapulothoracic muscles and the middle deltoid were assessed using EMG analysis. The task involved repetitive elevation of the arm through abduction using a Cybex dynamometer pulley system to provide resistance lasting approximately 1-2 minutes. All four muscles showed similar EMG signs of fatigue. Fatigue was associated with an increase in scapular rotation in the mid to end-range of elevation. The muscles that were analysed however were acting primarily as prime movers of either the humerus or the scapula. Increased rotation of the scapula may represent compensatory motion to prevent subacromial impingement of the humerus as fatigue of the stabilizers occurred.

Although research has provided relevant information concerning important biomechanical considerations during upper limb tasks, the literature is clearly deficient in information regarding upper limb tasks and fatigue of the dynamic stabilizers of the glenohumeral joint. Research has been focused on EMG accessible muscles such as the deltoid and the upper trapezius, and on muscles acting in their role as prime movers only.

F. Summary

The anatomy and biomechanics of the shoulder complex is designed to provide maximum mobility of the upper limb. Although both passive and active structures contribute to glenohumeral stability, the primary dynamic constraint of humeral motion arises from the rotator cuff musculature. The dynamic stabilization of the shoulder joint is important to prevent undesired translation of the humeral head, which may result in damage to tendons and passive structures. Dynamic instability of the glenohumeral joint is speculated to play a role in the development of rotator cuff disorder and impingement syndrome. During activities of daily living involving the upper limb, the ability of the rotator cuff to provide stabilization of the glenohumeral joint may be adversely affected by selective fatigue of these muscles. It is important to determine the relative magnitude of this fatigue during upper limb tasks using a well-controlled study.

There has been no data published in the literature regarding the fatigue of the glenohumeral musculature acting as stabilizers during an upper limb task. Upper extremity activities that involve grip tasks are common to activities of daily living, industry, and sports. Injuries to the upper extremity during these activities are also common. Imbalances between the shoulder internal rotators (IR) and external rotators (ER) created by this differential fatigue may lead to altered kinematics of the glenohumeral joint resulting in injury and disability. This knowledge is critical to a better understanding of the patho-mechanics of shoulder dysfunction.

The results from this study will have immediate clinical implications to the fields of rehabilitation, sports, and industrial medicine by identifying a potential new aetiology for upper extremity repetitive stress injuries. This will allow for more appropriate evaluation, rehabilitation, and training of individuals with or at risk for upper extremity injuries by reinforcing the importance of shoulder strength, endurance, muscular balance, and joint angle in providing and maintaining appropriate glenohumeral joint stability during grip activities.

PURPOSE AND OBJECTIVES

A. Purpose

The proposed study will provide valuable objective information regarding fatigue of the glenohumeral rotators during a repetitive upper limb task. This will be the first study to examine the concept of differential fatigue of shoulder stabilizers during an upper limb task. This information will contribute to the further understanding of rotator cuff pathology and assist the health care provider in planning prevention and treatment programs for shoulder dysfunction in individuals of all ages.

B. Objectives

1. To determine and compare the fatigability of the internal and external rotators acting as glenohumeral stabilizers during a repetitive gripping task in a group of healthy subjects.
2. To determine if the magnitude of fatigability is dependent upon the shoulder joint position during the strength evaluation.
3. To determine the effect of upper limb position during the repetitive task on this fatigability.

C. General Hypotheses

1. Fatigue of the internal and external rotators will occur at a different rate than the prime movers during an upper limb repetitive grip task.

2. The magnitude of fatigue will occur at a different rate between the stabilizer muscles and will result in a change in the IR/ER ratio.
3. Fatigability of the glenohumeral rotators will be dependant on the position of the upper limb during the fatigue task protocol.
4. Both the initial strength and the rate of fatigue of the internal and external rotators will be angle dependant.

D. Limitations

Information garnered from assessing a relatively static upper limb task will not necessarily apply to more dynamic upper limb tasks. Evaluation of fatigue using only isometric testing will not necessarily provide information regarding fatigue during concentric and eccentric motion.

E. Delimitations

Subject selection was delimited to include only males aged 18 to 50 years of age with no history of upper limb injury or upper body weight training. Dynamometry testing was delimited to include only isometric evaluations in the scapular plane with shoulder joint positions of 30° and 60°.

F. Assumptions

1. The subjects selected from age 18 to 50 years of age are representative of the general population.
2. The isometric moment measure on the dynamometer is an accurate quantification of shoulder joint moment. This assumes that several criteria were met including alignment of the axis of rotation of the shoulder with the dynamometer, adequate stabilization of the subject.
3. The subject is exerting maximal effort during the strength measurements and the maximal gripping test. Note: the person is not exerting maximally for the repetitive grip task until the final repetition.
4. The acquired strength data follows a normal distribution curve.

METHODOLOGY

The following section on methodology will provide a detailed account of study design, subject recruitment, instrumentation, study protocol, data reduction and analysis and statistical analysis.

A. Study Design

This project involved a test-retest design with a short-term intervention where the subjects were required to perform a strength test both before and after a standardized fatiguing task performed in two distinct positions. The primary comparisons will be within person, corresponding to strength evaluations before and after the fatiguing task.

B. Subjects

The total number of subjects to be recruited in the study was determined by power analysis with the power index set at 3.28 (alpha=0.05, Beta =0.2). The power analysis equation uses this power index, combined with the expected mean difference and variance of the data to estimate the number of subjects required to provide sufficient statistical power. Isometric strength of the shoulder internal and external rotators is the primary parameter measured in this study. Based on preliminary data, the strength deficits after repetitive grip task was 16%. The variance used in the equation is based on estimates of the variance of isometric strength measures from published literature on internal/external rotation testing, as well as standard deviation calculations from preliminary data. Examples of the power equation using each of these variables can be found in Appendix A. The total number of subjects per group required to show the expected differences is a minimum of 8 using preliminary data calculations, and a maximum of 11 using published literature calculations. The total number of subjects used in this study is 15. This number should allow for a small level of subject attrition and missing or bad data.

Inclusion Criteria

Male subjects were included in this study. Only healthy subjects with asymptomatic shoulder function were included. Healthy was defined as not having had any type of medical problems within the last 12-month period for which they had to seek medical attention. Subjects had no known history of ergogenic or performance enhancing drug use.

Exclusion Criteria

Subjects who did not meet the above-specified criteria were excluded from this study. They were excluded if they had any history of upper extremity trauma, surgery or arthritis, or any medical or orthopaedic condition that would have prevented them from completing the protocol. Subjects were also excluded if they participated in upper extremity weight training, competitive sports or manual labour. Subjects with known neuromuscular disorder or restricted upper extremity range of motion were also excluded from the study. Left-side dominant subjects were excluded from this study for ease of testing.

General Recruitment Procedure

Subjects were recruited both through word of mouth and by posted advertisements at the Bannatyne campus and Fort Garry campus at the University of Manitoba.

Selection

After being recruited for the study, the appropriateness of the subject's inclusion into the study was determined by the one of the investigators to ensure conformity with the inclusion/exclusion criteria of the study. Each subject was required to fill out a questionnaire, which assisted with the identification of inclusion and exclusion criteria as well as provided information regarding occupational and recreational activity. (Appendix D).

Informed Consent and Ethical Approval

All subjects were required to complete an informed consent prior to participation in study. The study was approved by the University of Manitoba Faculty of Medicine Committee on the Use of Human Subjects in Research (Application #H2000:093)

C. Instrumentation

Dynamometers

1. Isovelocity Dynamometer

An isovelocity dynamometer (Kin-Com, 500H, Chattecx Corporation, Hixson, Tennessee, USA) was used in this study to measure the isometric strength of the internal and external rotators of the shoulder. The dynamometer is a microcomputer- feedback controlled instrument that measures force while maintaining angular velocity over a specified range of motion. For isometric contractions, the angular velocity is set to zero. The subject applies force against the actuator arm of the dynamometer, which houses a strain-gauge transducer calibrated in force units (N). The moment (Nm) produced by the subject is equal to the product of the force (N) and the moment arm (m). The moment arm corresponds to the perpendicular distance (m) from the line of action of the force application (normal to the actuator arm of the dynamometer) to the axis of rotation of the joint (which is aligned with the axis of rotation of the dynamometer). When the moment is compensated for the effects of the moment produced by the weight of the limbs attached, then the moment is termed the resultant joint moment about the shoulder. Isovelocity dynamometers provide objective and valid estimates of shoulder muscle strength or the resultant joint moment (Nm) produced by the shoulder (Greenfield, Donatelli, Wooden and Wilkes 1990, Hellwig and Perrin 1991, Kuhlman et al. 1992). Kuhlman et al. (1992) conclude that isometric testing of the shoulder is as reliable and reproducible as "isokinetic"(isovelocity) testing. The Kin-Com 500H dynamometer used in this study has excellent mechanical reliability, with an r value ranging from 0.948 to 0.999 (Farrell and Richards 1986).

2. Hand grip dynamometer

The Jaymar handgrip dynamometer was used to assess the subject's grip strength. The validity of and reliability of the Jaymar dynamometer (Sammons Preston Ltd., Bolingbrook, Ill.) is reported as excellent in the literature (Bellace, Healy, Besser, Byron and Hohman 2000, Nitschke, McMeeken, Buny and Matyas 1999). The Jaymar handgrip dynamometer has an adjusTable handgrip in order to fit it to the subject's hand. It is able to measure tension generated within the grip up to a maximum of 100 kg (100 kg = 981 N) in 1 kg (9.8 N) increments.

3. Visual Analogue Scale

The visual analogue scale will be used to assess pain intensity/discomfort in the upper limb (Appendix E). The visual analogue scale consists of a line 10 centimetres long, with descriptive phrases related to pain intensity located at the ends of the line. The VAS has been validated for the measurement of pain intensity (Jensen and Karoly 1992).

D. Procedure and Protocol

Subjects were assessed for both their present activity levels (both upper and lower extremities) and musculoskeletal history to ensure that the subjects met the inclusion and exclusion criteria and were suitable for the study (Appendix D). All subjects accepted into the study were given uniform instructions regarding the procedures involved in the study and a general overview/familiarization of the equipment that was utilized in the measurement of upper extremity strength and fatigue. Measurement of body mass (kg) was obtained by weigh scale and estimated to the nearest 1/100th of a kg..

The VAS was administered before and after the initial strength test, following the fatigue task protocol and again after the final strength test to assess upper limb pain/discomfort in 5 subjects. The results from the VAS were used to analyse the possible relationship between strength changes and pain/discomfort in the upper limb. All testing was performed by one of the investigators.

Strength Testing

1. Internal and External Rotation

Positioning and Alignment: The subjects were seated beside the dynamometer with their shoulder joint positioned in the plane of the scapula (45° abduction and 30° forward flexion). The plane of the scapula is defined as 30° anterior to the frontal plane (Poppen and Walker 1976, Saha 1971) and has been recommended by several authors as the ideal testing position for evaluation of shoulder internal and external rotation strength (Dvir, 1995, Hellwig and Perrin 1991, Kuhlman et al. 1992), although no justification beyond the recommendation was provided. The seated position is the recommended position for testing shoulder strength, particularly for the internal and external rotators (Dvir 1995). Shoulder placement was accomplished by rotating the chair 15 degrees relative to the dynamometer arm subject's shoulder adjusting the distance of the chair laterally from the dynamometer arm to place the shoulder in the plane of the scapula. A standard hand held goniometer was used to assess shoulder position. Tape was placed on the floor to visually mark the position of the chair in order to ensure repeatable placement for each subject.

The elbow was flexed to 90° and the olecranon process was fitted into the v-shaped pad centred over the rotational axis of the dynamometer. The subject's forearm and wrists were in neutral positions and oriented parallel to the resistance arm of the dynamometer. Subjects did not grip the input shaft handle, but rather the ventral aspect of the distal forearm was approximated against the padded shaft. The longitudinal axis of the humeral shaft was visually aligned with the axis of rotation of the dynamometer (visually estimating the alignment of the axes of the dynamometer to that of the glenohumeral joint), ensuring that the recorded moment arm would correspond to the moment arm about the shoulder joint. This aids in reducing systematic errors in estimating the resultant joint moment. This required the head of the dynamometer to be rotated approximately 45°. The height of the dynamometer head was adjusted by hydraulic control system integrated into the dynamometer head to maintain the shoulders level. To minimize trunk rotation and extraneous movements, subjects were stabilized using a diagonal chest strap around the contralateral hip and ipsilateral shoulder region and a horizontal strap around the waist and chair. The subject's contralateral arm rested in their lap and their feet were unsupported.

Figure 4 illustrates the subject's basic upper limb positioning on the dynamometer.

Figure 4: Subject positioning for strength testing on the isovelocity dynamometer.

The angle of rotation and height of the dynamometer head and length of the moment arm was recorded for each subject.

Testing Protocol: Following a verbal explanation and demonstration, each subject completed a warm-up session of 2 sub-maximal isometric contractions for both internal and external rotation followed by a 5-minute rest. Five minutes was permitted to elapse prior to beginning the maximal internal and external rotation strength tests to minimize any motor learning effects and post-tetanic potentiation associated with the warm-up manoeuvres. This served as a general warm-up and familiarization with the equipment and the procedure.

Maximal voluntary contraction (MVC) is defined as a voluntary contraction that the subject believes to be maximal that is performed with continuous feedback and encouragement (Gandevia et al. 1995). During maximal effort (MVC) a standard set of verbal prompting was given to each subject using the words "push, push, push" beginning after 1 second and continuing until the end of the contraction. No other verbal prompting was given. Testing was restricted to 30° and 60° of external glenohumeral rotation with the forearm horizontal position referenced as zero. In Figure 4 the subject's arm is positioned in 30° of external rotation. The subject performed 2 isometric MVCs of both IR and ER at both angles. A balanced design was used to determine the order of testing. Half of the subject's started with two internal rotation contractions and the other half began with two external rotation contractions. The same order was used for the initial and subsequent strength test for each subject. The order of testing was recorded for each subject. Hughes et al. (1999) and Kuhlman et al. (1992) both reported that randomised order of testing did not change strength data.

The subject was instructed to gradually increase the force of contraction over a 1 second period with their forearm against the padded handgrip (the subject's hand is not involved in the contraction) until they reached their maximum and to hold it at maximal effort until the end of the 5-second contraction period. They were instructed relax for 5 seconds. The dynamometer visually and audibly counts out both the contraction and relaxation time for each repetition. A VAS was administered both prior to and following this test as a functional provocation test. The identical procedure was repeated immediately following the fatigue task protocol.

The resultant joint moment is derived by compensating the moment recorded by the dynamometer with the moment of the weight of the segments. This is commonly referred to as *"gravity correction"*. Moment values were not corrected for the moment of the weight of the forearm in this study. The mean moment of the weight of the forearm for all 13 subjects in this study was 0.99 Nm at 30° and 0.56 Nm at 60° (range of 0.38 to 1.82). These moment values would result in an error in derived peak moment of only 1.4% at 60° to 2.6% at 30° for both the internal and external rotators. This together with the use of a within-subjects design eliminates the need to compensate for these values. These calculations are in agreement with DePauw and Kriellaars (1996), who used the same testing protocol and determined that the variation in moment of the weight of the forearm and hand between subjects would not be greater than 1 Nm.

2. Hand Grip

Positioning and Alignment: The subject was standing with their arms at their sides (0° shoulder flexion), elbows extended and the forearm positioned in neutral. The handgrip dynamometer was fitted comfortably for each subject prior to testing. The dynamometer was held parallel to the subject's side with the dial facing away from their body. Positioning and procedures were in accordance with recommendation from the American Society of Hand Therapists and have been supported in the literature (Bellace et al. 2000, DeSmet, Tivez and Stappaerts 1998, Oxford 2000).

Testing Protocol: The subject was asked to squeeze the handle of the handgrip dynamometer as hard as they could without moving their arm. This was repeated two more times, for a total of three trials for each subject. The best score was recorded as the subject's maximum isometric grip strength. Sixty percent of this value was used to determine the tension required for successive contractions during the repetitive gripping task.

Fatigue Task

Positioning and Alignment: The subject was seated in the dynamometer with the upper arm positioned vertically and the forearm positioned horizontally (elbow flexed to 90°). This will be referred to as position A. The subject was instructed to move their trunk forward in order to remove any trunk stabilization afforded by the back of the chair. The subject's feet were supported on a footstool for the fatigue task. The dial of the handgrip dynamometer faced away from the subject during use. A mirror was positioned in front of the subject to provide the necessary visual feedback for the marked target value on the handgrip dynamometer. The subjects returned to the study one week later (minimum 5 days maximum 10 days later) to repeat the identical protocol using position B. Position B was defined as the subject seated with the upper arm and forearm positioned horizontally. Upper limb position was determined visually for each individual subject. The use of a mirror that the subject could observe their position, as well as verbal feedback ensured the maintenance of humeral position during the task.

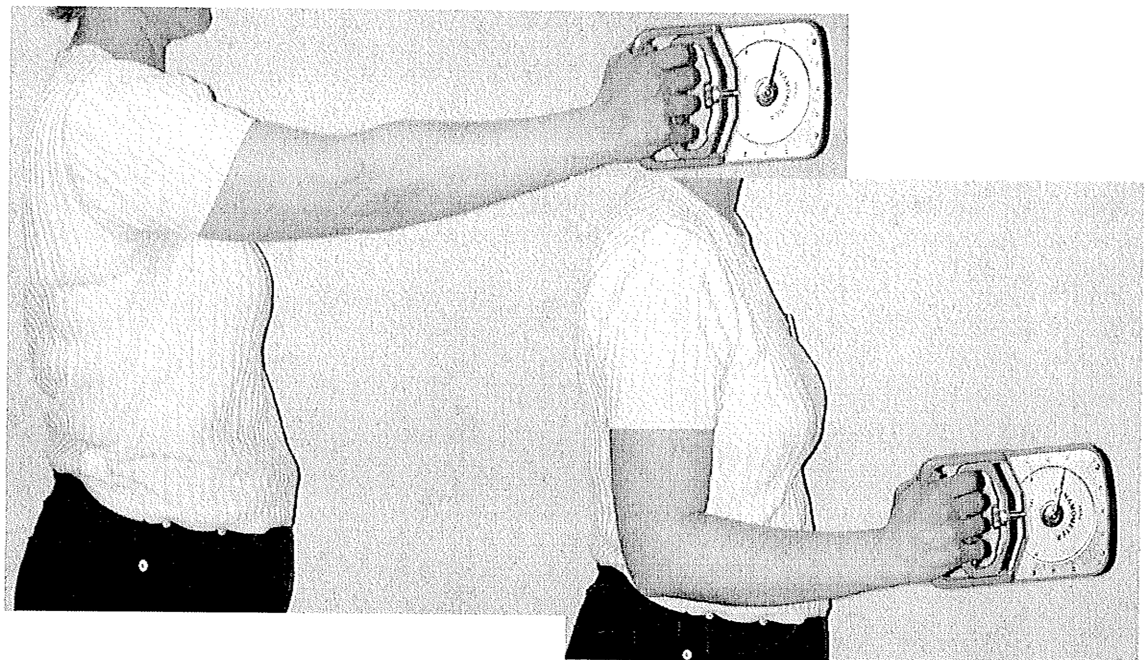

Figure 5: Upper limb position for fatigue task protocol.

Position A: upper arm vertical, forearm horizontal is shown on the right. Position B: upper arm horizontal is shown on the left.

Testing Protocol: Each subject performed repeated gripping of the handgrip dynamometer to 60% of the subject's recorded maximum value. The subject was required to hold each contraction for 5 seconds and rest for 3 seconds each time generating 60% of the maximum value, as determined by a mark placed on the dial of the handgrip dynamometer, viewable by each subject in the mirror. Contractions continued until the subject could no longer generate 50% of the original maximum value despite verbal prompting. This was defined as 50% fatigue of the prime movers. An electronic metronome was used to time the work and rest periods. The number of repetitions necessary to produce a 50% fatigue in the forearm flexors was also recorded. Consistent verbal prompting was provided to each subject to encourage consistent effort. The subjects repeated the IR and ER strength assessment using the identical protocol immediately following handgrip fatigue task to measure the glenohumeral internal and external rotator fatigue.

E. Data Collection and Reduction

The isovelocity dynamometer is equipped with an on-board data acquisition system that samples the generated signals (force, angle and velocity) at a sampling frequency of 100 Hz. The raw data collected was exported from the on-board computer and then downloaded into a conversion program (Isomap, Isodyne Inc., Winnipeg, Manitoba) that compiled the report files and converted them to text files for import into spreadsheets. This program provided summary data regarding the strength parameters including peak moment, angle of peak moment, average moment and duration for each contraction. These parameters were exported to an excel spreadsheet for further analysis. The internal and external rotation strength data for each subject was assessed for:

1. Peak shoulder joint moment during contraction
2. Average shoulder joint moment over 4.5 seconds. This was chosen based on the minimum contraction duration for all subjects of 4.5 seconds.
3. Internal/External rotation isometric strength ratios: determined for each subject at each test angle. The strength ratio was be calculated by dividing the peak/average internal rotation moment by the peak/average external rotation moment for 30 and 60 degree positions.

F. Statistical Analysis

Statistical analysis was performed using Microsoft Excel software program for Windows. The level of significance for all analysis was assessed at an alpha level of 0.05.

Paired one-tailed t-tests were used to examine differences in the isometric strength parameters of peak moment, average moment, and IR/ER ratios after the repetitive grip task, as well as to examine the differences in fatigue between position A and B. SEM and P values were reported with two significant decimal places.

Post-hoc analysis on intra-tester reliability of the strength testing protocol was determined using the intraclass correlation coefficient. The Pearson Product correlation coefficient was calculated along with scattergrams to examine possible relationships between absolute grip strength, age, body mass, activity level, initial strength and strength deficit.

RESULTS

A. Subject Demographics

A total of 15 subjects participated in this study. Two subjects were lost to attrition; one due to a low back injury that occurred between testing sessions and the second for failing to return for repeated testing in position B. Table 1 summarizes the demographics of the remaining 13 subjects included in the study. The range of activity was large with values up to 7 times per week and as low as no activity per week.

Table 1: Subject demographics

	Range	Mean ± SD
Age (yrs)	22 - 54	38.9 ± 9.9
Body Mass (kg)	74.30 - 94	80.43 ± 9.59
Average frequency of activity per week	0 - 7	3.15 ± 2.23

B. Strength Evaluation

Internal and External Rotator Strength Evaluation

Figure 6 demonstrates a typical resultant joint moment graph from the isometric strength evaluation for internal and external rotators both before and after completing the fatigue task protocol in position A. By referencing the mid position of internal and external rotation to horizontal, the internal rotator resultant joint moment is depicted by a negative value and the external rotator resultant joint moment by a positive value. For this subject the average moment deficits were 11% at 30° and 2% at 60° for the external rotators, and 25% at 30° and 13% at 60° for the internal rotators.

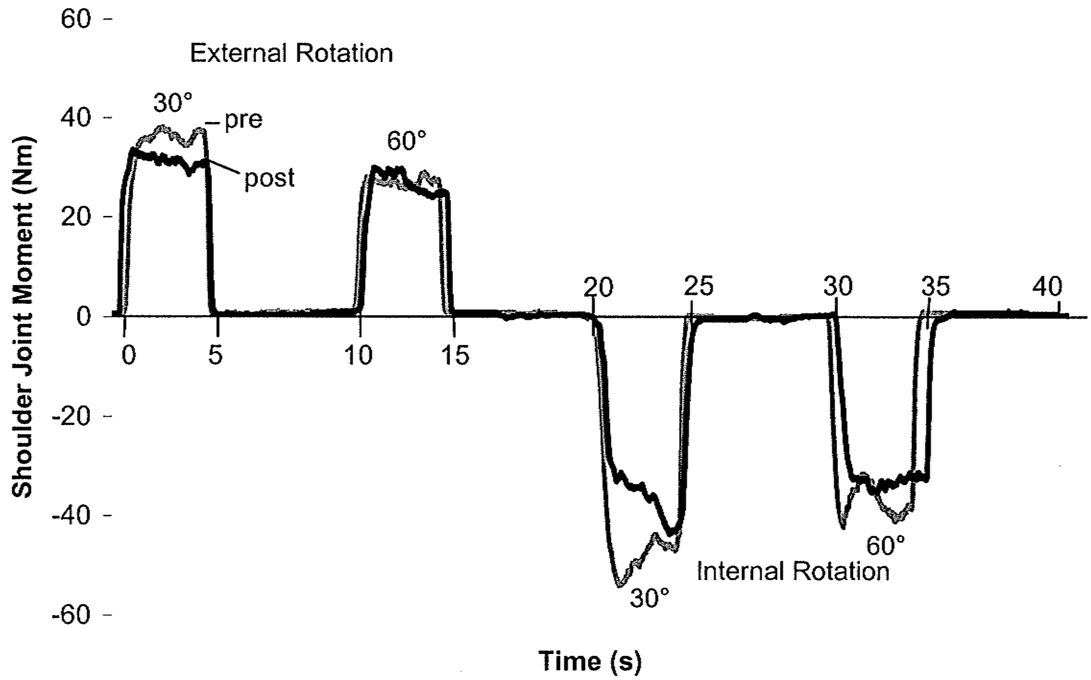

Figure 6: External and internal rotator resultant joint moment data from an individual subject at 30° followed by 60° both before (light) and after (dark) completing the repetitive grip task in position A.

Intra-rater test-retest reliability for the isometric strength test protocol used in this study was determined using post-hoc analysis of two repeated measures. Calculations were based on all 13 subjects' peak and average moment values. Comparisons were made between the initial before-task strength test and the before-task strength test one-week later. The intra-class correlation coefficient (ICC) was calculated for both the internal and external rotators at each testing angle using a 95% confidence interval (Table 2). The minimum value obtained was for external rotation peak moment ($r=0.88$) and the maximum value obtained was for the external rotation average moment ($r = 0.92$).

Table 2: Test-retest reliability of isometric internal and external rotation for peak moment and average moment using the intraclass correlation coefficient.

	Peak Moment	Average Moment
IR30°	0.91	0.86
IR60°	0.85	0.90
ER30°	0.88	0.92
ER60°	0.88	0.92

Mean absolute values for peak and average moment are reported in Table 3 for position A and Table 5 for position B. Correlation analysis of peak and average moment values revealed a statistically significant positive relationship between peak and average moment for all four strength tests ($r = 0.71$, $p <. 0.005$). Therefore, only the absolute values for average moment are graphically illustrated in Figure 7. Figure 7 shows that the resultant joint moment produced by the internal rotators and external rotators was significantly greater at the 30° test angle (dark bars) than the 60° test angle (light bars) ($p<0.0001$). Comparison between the internal and external rotators at the same angle of testing revealed a significantly greater moment of the internal rotators than the external rotators at both 30° and 60° ($p<0.01$).

A positive correlation was observed between internal rotator strength and body mass ($r = 0.73$, $p <0.005$). No significant correlations were found between strength and body mass, age or activity frequency. The mean duration of each contraction was 5.62 seconds with individual contractions ranging from 4.5 to 6.4 seconds in duration.

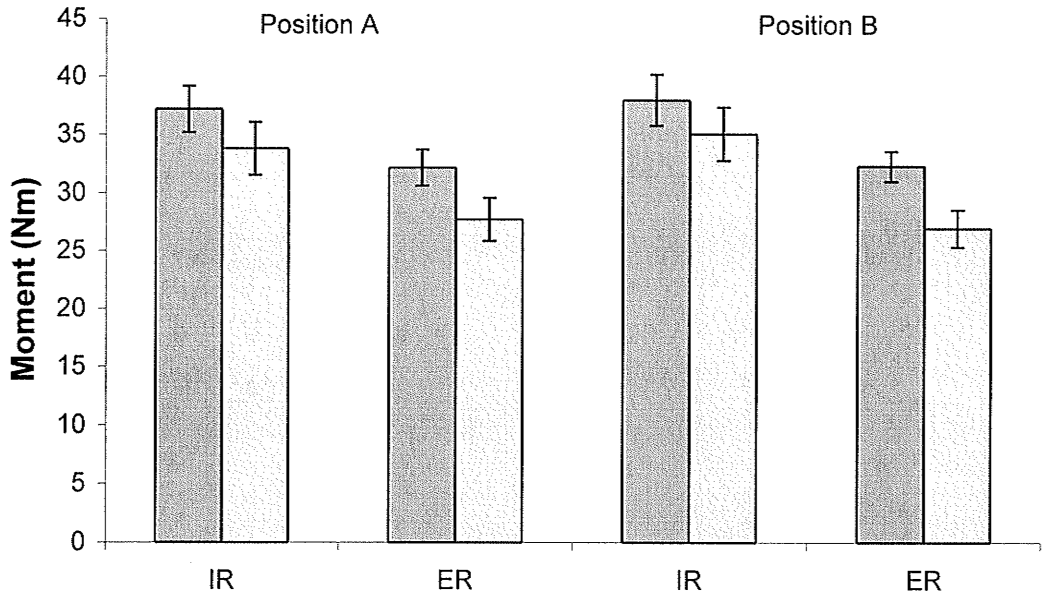

Figure 7: Comparison of initial average moment (Nm) internal (IR) and external (ER) rotator absolute values (standard error ±) between the 30° (dark bars) and 60° (light bars) test angles. Significant differences ($p<0.001$) were found between these angles for all four comparisons. Significant differences were also found ($p<0.01$) between the internal and external rotators at the same angle of testing.

Grip Strength Evaluation

The maximum grip strength and the number of grip task repetitions for each subject is shown in Table 3. Results are given for both position A and position B. There were a statistically significant ($p<0.05$) greater number of repetitions completed in position B than in position A. The average duration of the fatigue task was 3 minutes and 44 seconds for position A and 4 minutes and 32 seconds for position B which corresponds to a 47 second longer total fatigue task duration for position B. There was no statistically significant difference between grip strength values between trial 1 and trial 2 ($p=1.99$). Correlation analysis revealed a significant inverse relationship ($r = 0.82$, $p<0.001$)) between grip strength and the number of repetitions completed. This relationship is illustrated in Figure 8. A positive relationship was observed between grip strength and peak moment of the internal and external rotators ($r = 0.59$ and 0.62, $p<0.05$). There were no significant correlations found between grip strength and age, body mass or activity level or between the number of repetitions completed and age or body mass.

Table 3: Summary of maximum grip strength and number of handgrip dynamometer repetitions completed for the fatigue task protocol both for position (trial) A and position (trial) B. The mean and standard deviations are included. A significant difference was found between repetitions for position A and position B (shaded).

Subject	**Grip Strength (kg)**		**Repetitions**	
	Trial A	*Trial B*	*Trial A*	*Trial B*
1	56	61	24	21
2	44	46	30	34
3	50	46	26	45
4	48	44	21	27
5	38	42	48	29
6	42	40	39	51
7	52	53	25	28
8	40	38	38	50
9	47	46	32	38
10	56	56	25	33
11	48	42	22	38
12	51	50	19	22
13	54	52	21	31
Mean	48.15	47.38	28.46	34.38
St. dev.	5.84	6.69	8.62	9.70
P value	0.19		0.02	

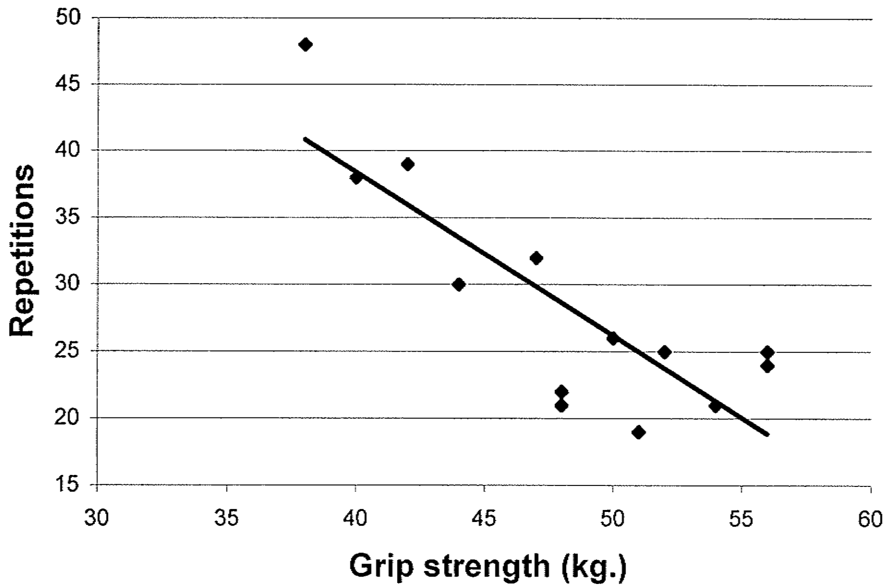

Figure 8: Scatter plot depicting maximum grip strength versus number of repetitions completed in the fatigue task protocol in position A. r = 0.82, slope = 1.22, intercept = 87.22

C. Shoulder Strength Deficit in Position A

Figure 9 graphically illustrates the average and peak moment deficits (in percent) of the internal and external rotators following the fatigue task protocol in position A. As illustrated, there was a statistically significant difference ($p<0.05$) in peak moment (dark bars) for all tests except external rotation at 60°. The mean peak moment difference ranged from 3% for the external rotators at 60° to 9% for the internal rotators at 30°. There was a significantly greater peak moment deficit of the external rotators at 30° compared to 60° ($p<0.05$), but there was no significant difference for the internal rotators ($p=0.09$). When comparing the deficit between the internal and external rotators, the internal rotators had a statistically significant greater deficit than the external rotators at 60° ($p<0.05$). Individual peak moment deficit values ranged for each strength test. Internal rotator strength changes ranged from a 9% increase (3.6 Nm) at 30° to a 28% decrease (7.9 Nm) at 30° and a 9% increase (2.8 Nm) to a 16% decrease (6.8 Nm) at 60°. External rotator strength changes ranged from a 14% increase (5.3 Nm) to a 19% decrease (7.1 Nm) at 30° and a 6% increase (2.8 Nm) to a 10% decrease (4.4 Nm) at 60°. The number of subjects who had an increase in strength ranged from 1 for IR60 and ER30 to 3 for IR30 and 5 for ER60.The smallest range in deficits was observed in the external rotators at 60° and the widest range in the internal rotators at 30°. The pattern of deficit across all four strength tests was inconsistent between subjects. For example, the same subject that had the greatest deficit at ER30 (19%) at the same time had the least deficit at IR60 (-9%).

Figure 9 demonstrates a statistically significant difference in average moment (light bars) for all test positions ($p < 0.05$). The mean difference ranged from 5% for the external rotators at 60° of external rotation to 7% for the internal rotators at 30° and 60°. The deficit in average moment for the internal and external rotators did not differ significantly between testing angles. Although there was no statistically significant difference between the internal and external rotator average moment deficit at 30° ($p=0.31$), there was a greater deficit of the internal rotators at 60° that approached statistical significance ($p=0.06$). Individual average moment strength changes ranged from a 23% increase (7.3 Nm) at ER30 up to a 31% decrease (7.57 Nm) at IR30. Table 5 summarizes the t-test values for position A.

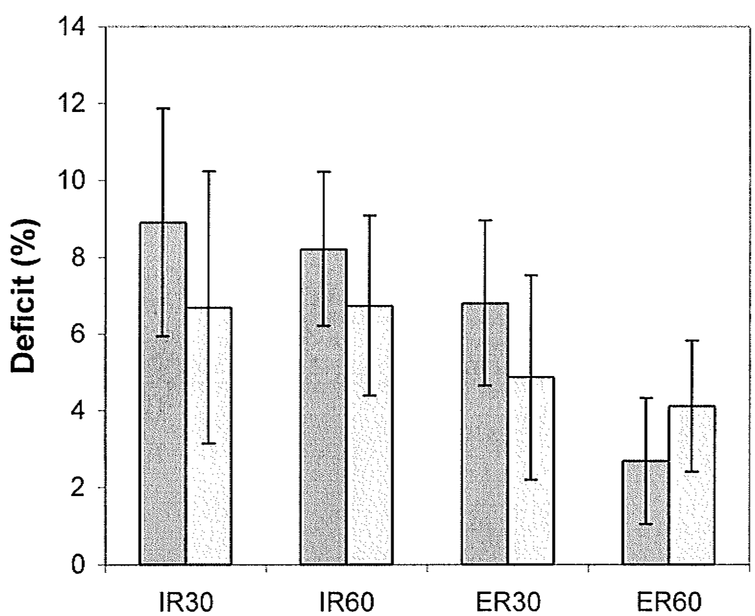

Figure 9: Percent deficit (standard error) for peak (dark bars) and average (light bars) moment for position A. All four tests are displayed. Significant deficits ($p<0.05$) were found for all tests, except peak moment ER60.

Table 4 lists the average absolute values (Nm) for both peak and average moment for position A. The difference in average moment ranged from 1.1 Nm for the external rotators at 60° to 2.6 Nm for the internal rotators at 60°. The difference in peak moment ranged from 1.7 Nm for the external rotators at 60° to 3.9 Nm for the internal rotators at 30°. Both the internal and external rotators generated a significantly greater moment at 30° compared to 60° both before and after the fatiguing task ($p<0.001$).

Table 4: Mean value (±standard error) for peak moment (Nm) and average moment (Nm) of internal and external rotator strength both before and after completing the fatigue task protocol in position A.

Strength Test	Peak Moment		Average Moment	
	Pre	*Post*	*Pre*	*Post*
IR30°	44.26 ± 2.22	40.32 ± 2.24	37.18 ± 1.98	34.71 ± 2.11
IR60°	40.39 ± 2.47	36.86 ± 2.08	33.8 ± 2.25	31.23 ± 1.82
ER30°	39.40 ± 1.70	36.55 ± 1.66	32.15 ± 1.55	30.46 ± 1.49
ER60°	35.33 ± 2.30	33.63 ± 2.31	27.71 ± 1.86	26.62 ± 1.86

Table 5: Summary of statistical analysis for internal and external rotator peak and average moment. Level of significance $p<0.05$. Significant values are shaded..

Strength Test	Peak Moment		Average Moment	
	Position A	*Position B*	*Position A*	*Position B*
IR30°	.004	.008	.037	.267
IR60°	.0005	.039	.001	.046
ER30°	.004	.00003	.040	.001
ER60°	.096	.015	.026	.002

D. Shoulder Strength Deficit in Position B

The same calculation methods used for position A were employed for determining the deficits seen in both peak and average moments for position B. Figure 10 displays the peak and average moment deficits (in percent). The peak moment (dark bar) deficits were statistically significant for all four strength tests ($p<0.05$). The mean peak moment difference ranged from 2% for the internal rotators at 60° to 9% for the external rotators at 30°. Although there was no statistically significant difference in the peak moment deficit detected between the 30° and 60° angles, there was a trend towards a greater deficit of the internal ($p=0.09$) and external rotators ($p=0.08$) at 30°. Although not exceeding the alpha level set for this study, the p values were both approaching significance. There were no significant differences measured between the peak moment deficit of the internal and external rotator muscles at either 30° or 60°.

Individual internal rotator strength changes ranged from a 4% increase (1.5 Nm) to an 18% decrease (9.9 Nm) at 30° and a 5% increase (1.7 Nm) to a 9% decrease (3.7 Nm) at 60°. External rotator strength changes ranged from a 2% (0.8 Nm) to 18% decrease (6.7 Nm) at 30° and a 7% increase (1.7 Nm) to a 31% decrease (10.6 Nm) at 60°. The number of subjects who experienced an increase in peak moment ranged from 0 at ER30 to 3 at IR30 and ER60 to 4 at IR60. In contrast to position A, the greatest range in deficits was observed in the external rotators at 60°, and the smallest range in the internal rotators at 60°. The pattern of deficit between strength tests was variable between subjects for instance; the same subject that had the greatest deficit at ER30 (19% decrease) at the same time had the least deficit at IR60 (9% increase).

The results for average moment are also displayed in Figure 10. There was a statistically significant difference in average moment (light bars) for all test positions except internal rotation at 30° ($p<0.05$). The mean difference ranged from 2% for the internal rotators at both 30° and 60° of external rotation to 8% for the external rotators at both angles. The deficit in average moment did not differ significantly between the 30° and 60° testing position. The external rotators had a significantly greater deficit (%) at both the 30° and 60° angle of testing than the internal rotators ($p<0.05$). Individual average moment deficit values ranged from a 10% increase (4.3 Nm) at IR30 up to a 28% decrease (7.4 Nm) at ER30. Table 5 summarizes the t-test values for position B.

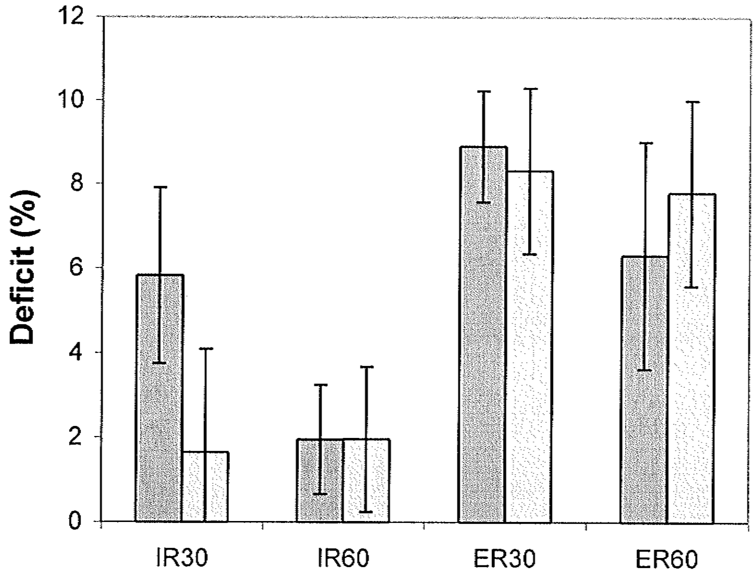

Figure 10: Percent deficit (standard error) for peak (dark bars) and average (light bars) moment of the internal and external rotators for position B. Significant deficits ($p < 0.05$) in peak and average moment were found for all tests except average moment IR30

Table 6 lists the averaged absolute values (Nm) and the standard error for both peak and average moment for position B. The difference in average moment ranged from 0.6 Nm for the internal rotators at 30° to 2.7 Nm for the external rotators at 30°. The difference in peak moment ranged from 1.0 Nm for the internal rotators at 60° to 3.4 Nm for the external rotators at 30°. These differences are in agreement with the derived percentage values. Significantly greater peak and average moments were generated by the internal and external rotator muscles at the 30° angle following fatigue ($p<0.01$).

Table 6: Mean value (±standard error) for peak moment (Nm) and average moment (Nm) of internal and external rotator strength for position B both before and after completing the fatigue task protocol.

Strength test	Peak Moment		Average Moment	
	Pre	*Post*	*Pre*	*Post*
IR30°	45.06 ± 2.33	42.52 ± 2.43	37.97 ± 2.2	37.40 ± 2.33
IR60°	40.60 ± 2.37	39.58 ± 2.06	35.07 ± 2.29	34.02 ± 1.86
ER30°	37.31 ± 1.51	33.96 ± 2.06	32.31 ± 1.28	29.61 ± 1.34
ER60°	31.65 ± 2.29	29.46 ± 1.86	26.97 ± 1.6	24.83 ± 1.55

E. IR/ER Ratio Changes for Position A and B

The IR/ER ratio for each subject was determined by dividing the mean average moment for internal rotation by the mean average moment for external rotation. IR/ER ratios were derived at each test angle for both position A and position B. The same calculations were computed using peak moment values. Figure 11 graphically illustrates the IR/ER ratios at the 30° test angle for each subject before and after completing the repetitive grip task in position B. At this angle, 7 subjects experienced an increase in their IR/ER ratio, 2 subjects a decrease in their IR/ER ratio and 4 subjects showed no change at all. Variable changes in the IR/ER ratios occurred at 60° and for both angles after completing the task in position A.

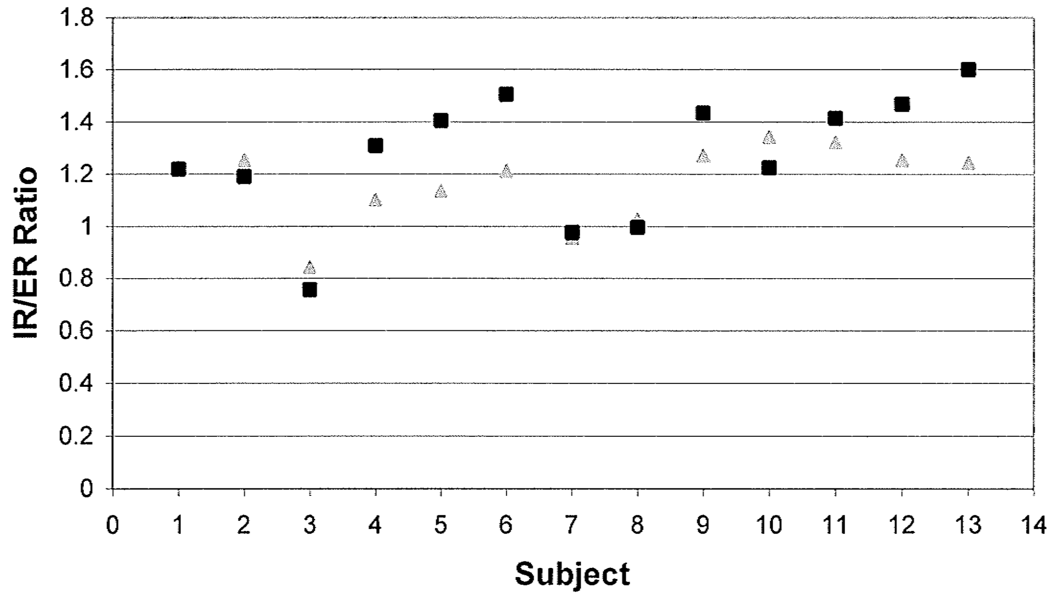

Figure 11: Scatter plot of internal/external strength ratios at 30° for all 13 subjects both before (triangles) and after (squares) completing the fatigue task protocol in position B.

The mean IR/ER ratios for position A (based on peak moment values) were: 1.18 at 30° and 1.30 at 60°. These ratios both decreased after fatigue to 1.16 at 30° and 1.23 at 60° The mean IR/ER ratios for position B (based on peak moment values) were 1.20 at 30° and 1.31 at 60°. There was a statistically significant difference in the IR/ER ratio at the 60° test angle ($p<0.05$). In contrast with position A, these ratios both significantly increased after fatigue to 1.25 at 30° and 1.37 at 60° ($p<0.05$). Individual IR/ER ratio values ranged from 0.81 to 1.78 before the task and 0.76 to 1.94 after the task. A summary of the t-tests for IR/ER ratios is presented in Table 7.

Table 7: Summary of statistical analysis for peak and average moment IR/ER ratios. Level of significance $p<0.05$. Significant values are shaded grey.

	Position A		Position B	
Test Angle	*Peak*	*Average*	*Peak*	*Average*
30°	0.31	0.35	0.09	0.02
60°	0.03	0.17	0.14	0.04

The percent change in peak and average moment IR/ER ratios that occurred following the fatigue task protocol in both position A and position B are detailed in Table 8. Calculations were made for the mean value based on both actual and absolute (in parenthesis) values. The mean IR/ER ratio increased by 1 to 5% for both test angles in position A, but decreased by 4 to 8% for both test angles in position B. The change in individual values varied from a decreased IR/ER ratio by 50% in position B to and an increased IR/ER ratio of 28% in position A.

Table 8: The percent change in IR/ER ratio values after completing the fatigue task protocol in both position A and position B. Both the mean change (absolute change) and the range of change are given in percent value. A negative value indicates an overall decrease in the IR/ER ratio and a positive value indicates an overall increase in the IR/ER ratio.

Percent Change in IR/ER ratios				
	Peak Moment		**Average Moment**	
	Mean	*Range*	*Mean*	*Range*
Position A				
30°	2 (12)	-32 to +24	1 (13)	-22 to +28
60°	5 (9)	-8 to +20	3 (8)	-6 to +20
Position B				
30°	-4 (9)	-26 to +9	-8 (12)	-28 to +11
60°	-6 (10)	-50 to +10	-7 (10)	-33 to +7

Figure 12 provides an illustrative comparison between the internal (dark bars) and external rotators (light bars) strength deficit for position A and position B and visually confirms not only the greater change in the IR/ER ratio in position B, but a dependency of IR/ER ratio pattern changes on upper limb position.

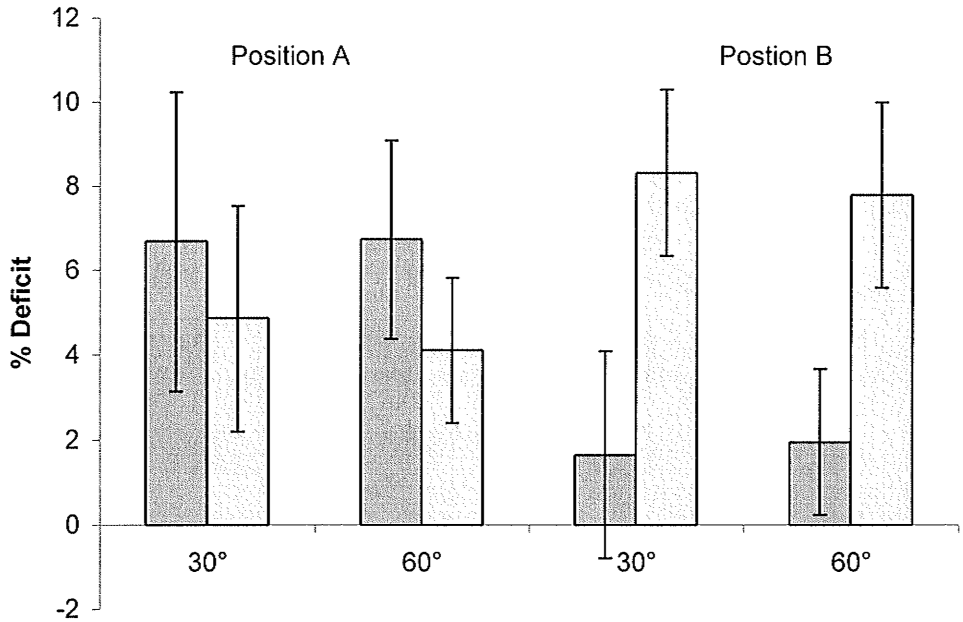

Figure 12: Comparison of the internal (dark) and external (light) rotator strength deficit for position A and position B at 30° and 60° testing angles. Significant differences were found in all the IR/ER ratios after completion of the task, except at the 30° testing angle.

F. Effect of Upper Limb Position.

Figure 13 visually displays the difference in average moment deficit between position A (dark bars) and position B (light bars). There was a statistically significant greater difference (5%) in the internal rotator deficit for position A at 60° ($p < 0.02$) and a 5% deficit at 30° that was not statistically significant ($p=0.08$). There was a non statistically significant difference of 3 to 4% between the external rotator deficit in position A compared to position B at both 30° and 60° testing respectively ($p=0.17$ and 0.12). Understandably, the mean maximum deficit of the internal rotators occurred with position A (7%), and the mean maximum deficit of the external rotators occurred with position B (8%). These differences resulted in a significantly greater change in the IR/ER ratio for position B than for position A ($p<0.05$) at both 30° and 60°, which can be visualized in Figure 12.

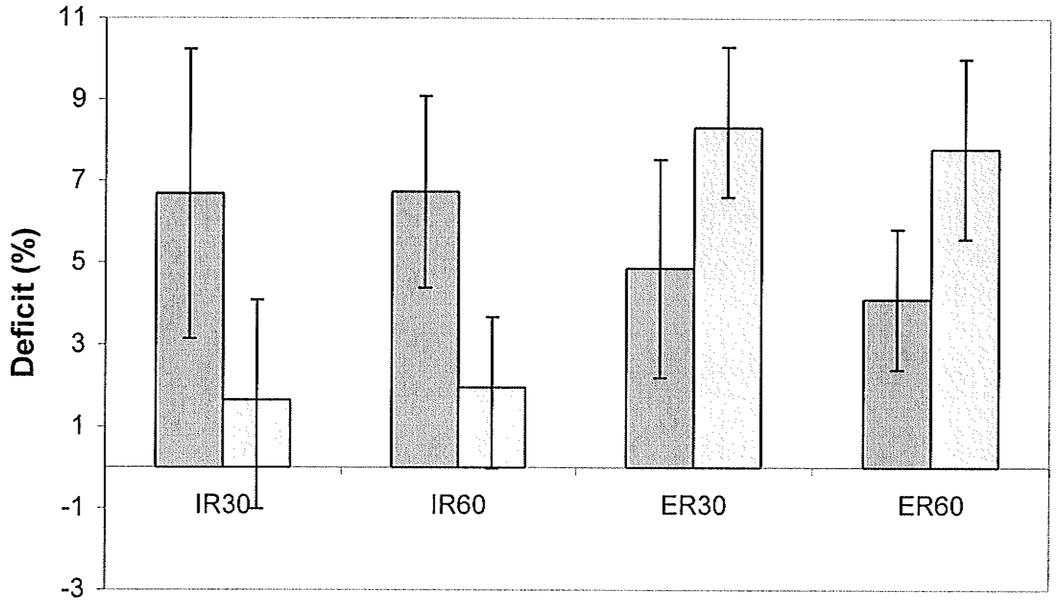

Figure 13: Percent strength deficit in position A (dark bars) and position B (light bars) for average moment. A significant difference was found between position A and position B for the internal rotators at 60° ($p<0.02$). Comparison between IR/ER ratio changes for position A and position B were statistically significant at 30° and 60°.

G. Other Results

No correlations were found between strength deficit and age, body mass, activity frequency or grip strength in either position A or position B. A positive correlation was found between the external rotator peak moment deficit (60°) and the number of repetitions completed in position B ($r = 0.60$, $p <0.05$).

H. Visual Analogue Scale Scores

VAS scores were tabulated on the first five subjects for both position A and position B. No significant difference ($p = 0.14$) was found in shoulder discomfort from prior to testing until after the final strength test. A significant increase ($p < 0.01$) was seen in forearm/hand discomfort following the repetitive grip task. After considering these findings, the VAS was discontinued to expedite the protocol and facilitate the primary objectives of this study. It was felt that the time taken to complete the VAS, particularly following the fatigue task protocol might have potentially reduced the strength changes being measured.

DISCUSSION

This study examined shoulder internal and external rotator isometric strength changes following an upper limb task in a group of healthy male subjects. The results indicate that while acting as dynamic glenohumeral joint stabilizers during the performance of a standardized repetitive gripping task, the internal and external rotators experience significant fatigue. More importantly, the position controlled grip task used in this study induced a differential rate of fatigue between the internal and external rotators resulting in a significant shift in the IR/ER ratio, which is indicative of a neuromuscular imbalance about the shoulder. Fatigue was measured in this study by calculating the deficit in moment generation (Nm) of the internal and external rotator muscles after completing the standardized fatiguing grip task in two distinct positions. The overall objective of this study was to quantify and compare the fatigue of the internal and external rotators, as well as determine the effect of upper limb position on that fatigue. The results indicate the following: firstly, performance of the task in either position resulted in significant fatigue of both the internal and external rotator muscle groups, secondly, the magnitude of fatigue was dependant on the muscle group tested, the shoulder joint angle for testing and the specified task position, and finally, a wide range of individual strength changes were observed, with increased strength observed in some subjects, while others experienced over 30% deficits in peak or average moment after completion of the task in both positions.

Significant fatigue occurred in both the internal and external rotator muscles, but they did not fatigue to the same magnitude. The external rotators fatigued to a greater magnitude relative to the internal rotators with the arm positioned horizontally (position B), while the internal rotators fatigued at a greater rate relative to the external rotators with the arm positioned by the side (position A). These differences in the magnitude of fatigue resulted in significant changes in the mean IR/ER ratios for both positions. Individual changes in ratios had a wide range in variation that resulted in either an increase or a decrease in this ratio that is not necessarily represented by the mean changes. Nevertheless, a shift in neural control of the musculature in either direction could impair dynamic glenohumeral joint stabilization leading to undesirable humeral head migration, which is highly dependent on these muscle groups operating within a well defined, but individualized ratio. Consistent with our hypothesis, the internal and external rotators also demonstrated differential fatigue when compared to the prime movers. The forearm flexors were fatigued to 50% of their initial maximum value for each subject. In contrast, the mean maximum value of fatigue for the internal and external rotators, acting as stabilizers of the glenohumeral joint was 9%.

We have defined the difference in magnitude of fatigue between muscle groups as differential fatigue. The results of this study clearly demonstrate the occurrence of differential fatigue and highlight its potential contribution to the development of neuromuscular imbalance between the internal and external rotators leading to reduced stability at the shoulder. Differential fatigue can be explained by both anatomical and neural factors. Anatomical factors include; 1) the variation in muscle fibre type composition between the internal and external rotators which will determine the muscle's ability to sustain force. This is not the most likely explanation since both the internal and external rotators both operate within similar functional capacities as glenohumeral stabilizers and would therefore both likely require similar muscle fibres, 2) the differences in the development of intramuscular pressure within each muscle which can impair signal transmission (Bigland-Ritchie et al. 1999), and 3) variation of individual load requirements to achieve humeral head stabilization that will determine the intensity developed by each muscle. The neural factors relate to the ability of the nervous system to employ strategies to delay the onset of fatigue. These include optimising motor neuron firing rates, load-sharing or cycling of motor units, and alternating the recruitment of motor units between synergist muscles (e.g. between infraspinatus and teres minor). Previous activity patterns of the muscles will determine the ability of the nervous system to employ these strategies. Consequently, these neural strategies may adapt with training and reduce the occurrence of differential fatigue, as well as limit the impact on humeral head migration during tasks. In addition, the individual can learn to improve the control of distal segments in order to reduce the movement-induced perturbations at the shoulder. This could be achieved by direct reduction in accelerations at the distal endpoint of the upper limb while repetitively gripping, or by employing the interposed elbow joint to act to reduce the movement-induced or "buffer" perturbations at the shoulder.

The strength produced by the internal and external rotators was dependant on the joint angle of testing. Both the internal and external rotators initially produced a significantly greater peak and average moment at the 30° angle of testing compared to the 60° angle. As expected, the peak and average moments generated were still significantly greater at 30° for both muscles once fatigued. The effect of shoulder joint angle on moment is explainable by the moment arm length as well as the moment angle and length tension relationships. The resultant joint moment generated by each muscle is a product of each muscle's force and the respective moment arms. The force is in turn is influenced by the length of the muscle fibre, which determines the number of cross-bridges that can be activated (length/tension relationship), and the velocity of contraction. The length/tension and force/velocity relationships, as well as the moment arm length change at each angle throughout the entire range of motion, contribute to the moment or torque generation at the shoulder. The maximal moment for each muscle is produced at both the optimal muscle fibre and moment arm length and is unique for each muscle. The results indicate that for both the internal and external rotator muscles these factors are more favourable with the shoulder positioned at 30°, this does not exclude the possibility of a greater moment being produced at an angle that was not tested. The comprehensive strength assessment (4 speeds, 90° range of motion) performed by DePauw and Kriellaars (1996) of shoulder internal and external rotators indicate that the moment generation by the IR and ER muscle groups shows systematic changes over joint angles.

The greater reduction in strength that was measured in the external rotators at 30° after performance of the upper limb task indicates that there is not only angle dependant strength, but angle dependant fatigue as well. This fatigue related dependency is likely due to neural rather than biomechanical factors. The external rotators are subjected to different load requirements during the performance of the upper limb task. Consequently, its motor units may be recruited differently than the internal rotator motor units by the nervous system to achieve the desired neural control. The internal rotator muscle group consists of a much larger cross-sectional area for neural recruitment and this may explain why no angle dependence was detected for this group at the two angles tested. However, knowing that the strength of the internal rotators is angle dependant, it is likely that angle dependent fatigue would be detected at other angles of testing between 0 and 180°.

As hypothesized, the position of the upper limb exerted a significant effect on the magnitude and pattern of fatigue of the internal and external rotator muscles. Position A had a greater effect on internal rotator fatigue, while position B had a greater effect on external rotator fatigue. The dependency of internal and external rotator fatigue on upper limb position is clearly illustrated by the significant differences measured in the IR/ER ratios between position A and position B. It can also be indirectly illustrated by examining the differences in significant p values calcualated for the IR/ER ratios between position A and position B. More specifically, there were 2 significant values, with a third approaching significance ($p=0.09$) for the IR/ER ratios for position B, but only 1 significant value and no others approaching significance for position A. The discrepancy between the shoulder position during strength measurement and the task position must be taken into account. This, combined with the knowledge that fatigue is also angle dependant, could mean that a repetitive task performed in one position could and most likely would manifest in an inability of the internal and external rotators to provide adequate humeral stabilization at a different joint angle than the task was performed at. This must be considered by clinicians for assessment, prevention and treatment.

The variable effect of upper limb position on fatigue of the internal and external rotators can be explained by several factors. Firstly, elevating the arm to position B changes the length/tension relationship of these muscles as well as the moment arm length about the joint. It may increase the length of the external rotator muscle fibres beyond their optimal length for force production to a greater extent than the internal rotators. Secondly, the moment of the weight of the upper limb is greater when the humerus is flexed, thereby increasing the force requirements of both stabilizer muscles. This increased moment of the weight would be countered by the prime movers controlling humeral flexion (since the moment of the weight would tend to produce humeral extension). The humeral flexors also have secondary roles for internal and external rotation, which will impact on their contribution to humeral head stabilization. The extent of this contribution would then influence the primary internal and external rotators and their fatigue patterns. If the contribution of the humeral flexors favoured the internal rotators, less fatigue of this muscle group would be expected. Thirdly, by potentially recruiting from a larger cross-sectional area of muscle, the internal rotators may be able to strategically delay the onset of fatigue in position B through alternating recruitment strategies. Fourthly, the change in load requirements between position A and position B may increase the amount of intramuscular pressure developed and subsequently exacerbate fatigue. (Bigland-Ritchie et al. 1999). This may vary between muscles. Finally, during performance of the task with the arm elevated, increased proprioceptive feedback from the upper limb may signal the central nervous system to recruit accessory scapular and trunk muscles to assist the glenohumeral stabilizers with a more demanding task. The muscles recruited may biomechanically or anatomically favour assisting the internal rotators rather than the external rotators. Based upon these explanations, perhaps what appears to be an increased effect of position A on fatigue of the internal rotators should actually be viewed as a decreased effect of position B on these muscles. The preceding discussion may also help to explain; 1) why the magnitude of fatigue in this position was not as large as expected considering the increase in moment of the weight about the shoulder, 2) why subjects were able to complete a greater number of handgrip repetitions in this position, and 3) why there was a positive relationship between external rotator fatigue and the number of repetitions completed in position B.

The magnitude of fatigue varied widely between individuals. Some individuals experienced no fatigue or even an increase in strength. In others, strength deficits reached magnitudes of up to 33%. Each individual's rate of fatigue also varied depending on the strength test performed. Several factors may explain these findings. Firstly, the strength requirements of the internal or external rotator muscles to adequately stabilize the glenohumeral joint during performance of the grip task would be unique for each subject. Strength is dependant on the cross-sectional area of the muscle and the recruitment of motor units, while the resultant moment is a factor of both the length of the moment arm and the force produced (N). These factors would vary with each individual based anatomical variations, as well as differences in neural recruitment. Secondly, the ability of the central nervous system to delay the onset of fatigue by alternating or rotating the recruitment of motor units both within the individual muscle and between the muscle groups may also vary between subjects. Thirdly, each person's ability to control their distal segments involved in the task will ultimately affect the demands of the proximal musculature. For example, the greater the acceleration of the limb during the task, the greater the strength required to control the humeral head. In some individuals this acceleration may be buffered by an intermediate segment (elbow), but in others the effects will be translated up to the shoulder and will need to be buffered by the glenohumeral stabilizers. Finally, although careful consideration was given in choosing the subject population, the subjects were not entirely homogeneous. Consequently, various occupational and recreational factors may have provided a training stimulus or learning effect for certain subjects. Of course, this would impact on the level of deconditioning of the individual muscles, or the level of training. For these same reasons, the magnitude of differential fatigue varied between subjects resulting in a diversity of neuromuscular imbalances among the 13 subjects. This diversity in fatigability must be taken into consideration when designing preventative and therapeutic interventions.

It is important to consider that there was no correlation found between initial strength and fatigue of the internal or external rotators. This means that an individual's strength does not predict the magnitude of fatigue the muscle may experience during the execution of a task. The results of this study clearly show that although two individuals may have the same initial strength of the internal and external rotators, the fatigability of these muscles may in fact be quite different, possibly resulting in a muscle imbalance for one individual, but not the other. Consequently, a stronger individual may actually be more susceptible to muscle imbalance and ultimately humeral head migration depending upon the demands of the task that they perform. However this lack of correlation may be due to the fact that the loads were expressed as a percent of maximum for each individual rather than specified as a fixed absolute load (such as might occur with screwing in drywall screws). Specifically, if a stronger person is tasked to perform at the same relative level (percent level) as a weaker person, then the fatigability would favour the weaker. However, the fact that the stronger persons relative level would be substantially higher in absolute terms than the weaker persons, would allow the stronger person to perform a task with a fixed absolute load at a lower percentage of their maximum. This concept of dissociation between strength and fatigue is reinforced by the inverse relationship that was detected between the strength (maximum grip strength) and fatigability (number of grip repetitions) of the forearm flexors. Therefore, the ability to complete or perform tasks of varying intensity cannot be based on the individual's initial strength. Rather, an objective measurement of fatigue, such as is outlined by the protocol of this study, would provide a more complete and accurate assessment of the individual's actual capabilities. Moreover, therapeutic interventions must be targeted at the fatigue patterns of muscles and not be directed by the strength of individuals.

Isometric testing was chosen to examine shoulder internal and external rotation strength. The test-retest reliability of the strength protocol used in this study was very high. The intraclass correlation coefficients for peak moment ranged from 0.85 to 0.92. These values are in agreement with Malerba, Adam, Hans and Kreb (1993) who also found ICCs for internal and external rotator isometric testing of 0.81 to 0.93. Complete concentric and eccentric testing would have provided a more complete analysis of the strength changes in the internal and external rotator muscles. However, as an initial foray into this area, it was imperative to limit any confounding variables such as testing time in order to ensure that the primary objectives were achieved. For the same purpose, the visual analogue scores were discontinued. Although only administered on five subjects, the results from the visual analogue scale scores suggested that shoulder pain or discomfort did not confound the measurements of strength of the internal and external rotator muscles.

The isometric IR/ER ratios for peak moment ranged from 1.16 to 1.37. This is slightly lower than those outlined by Hughes et al. (1999) who examined isometric strength ratios in 60 males and 60 females aged 20 to 78 years of age. They reported IR/ER strength ratios of 1.33 to 1.71 using peak moment analysis. This may be explained by differences in shoulder joint positioning between the two studies that will change the moment/angle relationship of the muscles. Although most investigations of IR/ER ratios have all used isovelocity testing, results from these studies indicate a high degree of variability depending on the population studied and the testing positions used.

No associations between age and IR/ER strength ratios were observed. This is in agreement with Hughes et al. (1999). Nor was there a relationship detected between age and strength deficit. This lack of association indicates firstly that age cannot predict susceptibility to differential fatigue, secondly that younger individuals will not automatically be less vulnerable to neuromuscular imbalances resulting from differential fatigue and finally, that activity patterns throughout the course of life may overcome age related fatigability changes.

No other study has yet to quantify fatigue of the internal and external rotators while acting as stabilizers of the glenohumeral joint. However, the results of this study are in agreement with Chen et al. (1999) who indirectly examined fatigue of the internal and external rotator muscles during a repetitive upper limb task. A significant increase in humeral head migration occurred after 30% fatigue of the prime movers (deltoid). This decrease in humeral head control may be explained by fatigue of the glenohumeral rotators and the subsequent change in IR/ER ratios that would occur due to differential fatigue of these muscles. Consequently this change in muscle balance would lead to an increased or decreased influence of either the internal or external rotator moment acting on the humeral head. If the balance of neuromuscular forces shifted to favour superior translation of the humeral head, then an increase in superior humeral head migration could occur.

The prime movers in this present study were fatigued to 50% and resulted in significant fatigue of the dynamic stabilization unit of the glenohumeral joint. Based upon the evidence put forward by Chen et al. (1999), it is reasonable to suspect that during performance of the repetitive upper limb task in both position A and position B, significant increases in humeral head migration were occurring to a varying extent in each subject. Increased humeral head migration may result in impingement of the rotator cuff tendons as they traverse beneath the coracoacromial arch. Although this may ultimately lead to the development of impingement syndrome or rotator cuff disorder, the short-term consequences of these pathomechanics are not yet known.

This is the first study that has quantified neuromuscular fatigue in the internal and external rotator muscles while acting as stabilizers, and the first study to determine that differential fatigue occurs between these muscles groups during glenohumeral stabilization.

This differential fatigue will change the defined strength ratio that is to a great extent responsible for the dynamic stabilization of the glenohumeral joint. The susceptibility to differential fatigue will vary with each individual and cannot be predicted based on age or initial strength. Individualized strengthening programs may help to reduce the effects of differential fatigue. The results from this study also indicate that the shoulder joint angle and upper limb position influences the resultant neuromuscular imbalance between the internal and external rotator muscles. These findings may help to explain why certain individuals are more vulnerable to repetitive shoulder injuries, or to the development of rotator cuff disorder.

It is important to consider that a relatively innocuous task was chosen for this study. Both the contraction duration (5 seconds), as well as the overall duration of the task (less than 5 minutes) was relatively short. Tasks requiring contractions longer than 5 seconds, greater than 34 repetitions or durations in excess of 5 minutes would require greater endurance of the glenohumeral stabilizer muscles. In addition, a more functional task such as using a power drill would result in even greater demands on the stabilizer muscles due to an increase in the moment of the weight and greater acceleration of the upper limb during the task. It is not yet known what the short term or long term implications of repeated exposure to such conditions could be.

The results of this study clearly demonstrate the differential fatigability of the internal and external rotators during a relatively innocuous upper limb task. It is also evident that fatigue of these muscles is both angle and position dependent. Moreover differential fatigability cannot be predicted by initial strength or age. These findings have immediate implications for clinicians in the prevention and treatment of shoulder injuries. Careful consideration must be given not only to the position of the upper limb during a repetitive task, but for any tasks that follow as well. A thorough assessment must include fatigability behaviour of the glenohumeral stabilizers not just initial strength, and not only at a single angle of testing. The designing and implementation of therapeutic and preventative strengthening programs should then be based upon these findings.

CONCLUSIONS

The purpose of this study was to compare and quantify fatigue of the internal and external rotators during the performance of an upper limb task in two distinct positions. Fatigue was measured by calculating the change in isometric moment (Nm) of the internal and external rotators following the performance of the upper limb task. The specific strength testing protocol used in this study proved to be highly reliable in providing reproducible results among these subjects. The strength parameters of peak moment and average moment, as well as the calculation of the internal/external strength ratio were used for analysis.

Conclusions of this study are as follows:

1. Performance of a standardized gripping task in two distinct positions resulted in significant fatigue of the internal and external rotator muscles. The magnitude of this deficit varied widely between individuals and could not be predicted based on age or initial strength.
2. The magnitude of fatigue of the internal and external rotators was dependant upon the position of the upper limb during the task. Performance of the task with the arm elevated to 90° exerted a greater effect on the fatigue of the external rotators. Consequently, the IR/ER ratio was significantly higher in this position.
3. Strength of the internal and external rotator muscles was dependant on the angle of rotation of the shoulder joint, both before and after fatigue. Both the internal and external rotators were able to generate greater moment (Nm) when positioned at 30° rather than 60°. In addition, the external rotators displayed an angle dependant fatigue that was greater at 30°.
4. Differential fatigue occurred between the internal and external rotator muscles, resulting in a significant alteration of the internal/external rotation strength ratio in

either direction. This varied widely between individuals and was dependant upon the position of the upper limb during the fatiguing task.

5. Differential fatigue of the prime movers and stabilizers muscles resulted from performance of a repetitive upper limb task.

FUTURE RECOMMENDATIONS

The results of this study not only indicate that fatigue of the glenohumeral rotators occurs during an upper limb task, but more importantly that they fatigue to varying degrees in a rather unpredictable manner. Assessment of strength alone is not sufficient to provide this information. This underscores the need for the development of an instrument that can objectively assess an individual's unique muscle behaviour during the process of fatigue. The protocol used in this study could be refined into such an assessment tool. Once accurate assessment is available, the next step would be to develop a strengthening program to address the individual's specific deficits and functional demands.

Although significant correlations were not detected between internal and external rotator fatigue and age or activity level, the power of the study could be improved by increasing the sample size. The power of this study was calculated to achieve the primary objectives and may not have been large enough to detect a relationship between muscle fatigue and age or activity level. Increasing the sample size would allow further investigation into these areas. It is also possible that an increase in power would eliminate what were likely type II errors that occurred in this study.

Only healthy subjects were included in this study. Previous investigations have demonstrated a difference in strength between healthy and impingement subject populations (De Pauw and Kriellaars, 1996) as well as increased activity of the stabilizer muscles during glenohumeral range of motion in subjects with shoulder instability (Kronberg and Brostrom, 1995). These results indicate the need to examine the fatigability of the stabilizer muscles in this subject population. Fatigability of the internal and external rotators in subjects with shoulder pathology may be an important consideration for the planning and implementation of shoulder rehabilitation programs. Moreover, certain athletic populations (i.e. swimmer, baseball pitchers) and occupational groups display an increased prevalence of shoulder dysfunction (Wilk et al. 1993, Sjogaard and Sogaard 1998). A thorough examination of strength and endurance in these populations may help to identify muscle imbalances that could predispose these individuals to injury. Future studies could examine the effects of implementing specific exercise programs in these groups that are at a higher risk of developing rotator cuff disease.

Finally, the wide variation in fatigability of the internal and external rotator muscles seen in this study, suggests that certain subjects may possess a buffering system in the distal or intermediate segments of the upper limb that is able to reduce the load requirements of the glenohumeral stabilizers. Further investigation into the acceleration patterns of the upper limb during a repetitive task and the possible relationship of limb acceleration to stabilizer fatigue would potentially identify those individuals at higher risk for shoulder injury.

REFERENCES

Arwert, H.J., deGroot, J., Van Woensel, W.W., & Rozing, P.M. (1997). Electromyography of Shoulder Muscles in Relation to Force Direction. Journal of Shoulder and Elbow Surgery, 6(4), 360-370.

Beach, M.L., Whitney, S.L., & Hoffman S.A. (1992). Relationship of shoulder flexibility, strength, and endurance to shoulder pain in competitive swimmers. Journal of Orthopaedic and Sports Physical Therapy, 16, 262-268.

Bellace, J.V., Healy D., Besser, M.P., Byron, T., & Hohman, L. (2000) Validity of the Dexter Evaluation System's Jamar dynamomet6er attachment for assessment of hand grip strength in a normal population. Journal of Hand Therapy, 13(1), 46-51.

Bigland-Ritchie, B., Rice, C.L., Garland, S.J. & Walsh M.L. (1995). Task-dependent factors in fatigue of human voluntary contractions. Advances in Experimental Medicine and Biology, 386(27), 361-380.

Bigliani, L.U., Kelkar, R., Flatow, E.L., Pollock, G. & Mow, V.C. (1996). Glenohumeral Stability. Biomechanical Properties of Passive and Active Stabilizers. Clinical Orthopedics and Related Research, 330, 13-30.

Bigliani, L.U., Ticker, J.B., Flatow, E.L., Soslowsky, L.J. & Mow, V.C. (1991). The Relationship of Acromial Architecture to Rotator Cuff Disease. Clinics in Sports Medicine, 10(4), 823-839.

Binder-Macleod, S.A. (1995). Variable-frequency stimulation patterns for the optimisation of force during muscle fatigue. Muscle wisdom and the catch-like property. Advances in Experimental Medicine and Biology, 384, 227-240.

Bjorklund, M., Crenshaw, A.G., Djupsjobacka, M. & Johansson, H. (2000). Position sense acuity is diminished following repetitive low-intensity work to fatigue in a simulated occupational setting. European Journal of Applied Physiology and Occupational Physiology, 81(5), 361-367.

Blevins, F.T,, Djurasovic, M., Flatow, E.L., & Vogel, K.G. (1997). Biology of the Rotator Cuff Tendon. Orthopedic Clinics of North America, 28(1), 2-14.

Bouisset, S., Richardson, J., & Zattara, M. (2000). Do anticipatory postural adjustments occurring in different segments of the postural chain follow the same organisational rule for different task movement velocities, independently of the inertial load value? Experimental Brain Research, 132, 79-86.

Bradley, J.P., & Tibone, J.E. (1991). Electromyographic Analysis of Muscle Activity About the Shoulder. Clinics in Sports Medicine, 10(4), 789-805.

Burns, W.C., & Whipple, T.L. (1993). Anatomic Relationships in the Shoulder Impingement Syndrome. Clinical Orthopedics, 294, 96-102.

Cain, P.R., Mutschler, T., Fu, F., & Lee, S.K. (1987). Anterior Stability of the Glenohumeral Joint: A Dynamic Model. American Journal of Sports Medicine, 15(2), 144-148.

Carpenter, J.E., Blasier, R.B., & Pellizzon, G.G. (1998). The Effects of Muscle Fatigue on shoulder Joint Position Sense. The American Journal of Sports Medicine, 26(2), 262-265.

Chandler, T.J., Kebler, W.B., Stracener, E.C., Ziegler, A.K., & Pace, B. (1992).Shoulder strength, power, and endurance in college tennis players. American Journal of Sports Medicine, 20, 455-458.

Chen, S.K., Simonian, P.T., Wickiewicz, T.L, Otis, J.C. & Warren, R.F. (1999). Radiographic evaluation of glenohumeral kinematics: A muscle fatigue model. Journal of Shoulder and Elbow Surgery, 28(1), 49-52.

Cohen, R.B., & Williams, G.R. Jr. (1998). Impingement Syndrome and Rotator Cuff Disease as Repetitive Motion Disorders. Clinical Orthopedics and Related Research, 351, 95-101.

Cook, E.E., Gray, V.L., Savinar-Noque, E., & Medeiros, J. (1987). Shoulder antagonistic strength ratios: a comparison between college-level baseball pitchers and nonpitchers. Journal of Orthopaedic and Sports Physical Therapy, 8, 451-461, 1987.

Culham, E., & Peat, M. (1993). Functional Anatomy of the Shoulder Complex. Journal of Orthopedics and Sports Physical Therapy, 18(1), 342-350.

Curl, L.A., & Warren, R. (1996). Glenohumeral Joint Stability. Clinical Orthopaedics and Related Research, 330, 54-65.

DePauw L.A., & Kriellaars D.J. (1996). Shoulder strength and impingement syndrome (Abstract). 19th Annual Meeting of American Society of Hand Therapist. Atlanta, Georgia.

DeSmet, L., Tivez, B. & Stappaerts, K. (1998). Effect of Forearm Rotation on Grip Strength. Acta.Orthopaedica.Belgica, 64(4), 360-362.

Ellenbecker, T.S. (1991). A total arm strength isokinetic profile of highly skilled tennis players. Isokinetics and Exercise Science, 1(1), 9-21.

Ellenbecker, T.S., & Roetert, P.E. (1999). Testing Isokinetic Muscular Fatigue of Shoulder Internal and External Rotation in Elite Junior Tennis Players. Journal of Orthopedic and Sports Physical Therapy, 29(5), 275-281.

Enoka, R.M. (1995). Morphological features and activation patterns of motor units. Journal of Clinical Neurophysiology, 12(6), 538-559.

Enoka, R.M. (1994). Neuromechanical Basis of Kinesiology. Champaign, Illinois: Human Kinetics.

Enoka, R.M., & Stuart, D.G. (1992). Neurobiology of Muscle Fatigue. Journal of Applied Physiology, 72(5), 1631-1648.

Falkel, J.E., & Murphy, T.C. (1988). Case principles: Swimmers shoulder. In Malone, T.E. (Ed.) Shoulder Injuries, Sports Injury Management. Baltimore: Williams & Wilkins.

Fallentin, N., Jorgensen, K., & Simonsen, E.B. (1993). Motor unit recruitment during prolonged isometric contraction. European Journal of Applied Physiology, 67, 335-341.

Farrell, M., & Richards, J.G. (1986). Analysis of the reliability and validity of the kinetic communicator exercise device. Medicine and Science in Sports and Medicine, 18(1), 44-49.

Fu, F.H., Harner, C.D., & Klein, A.H. (1991). Shoulder Impingement Syndrome – A Critical Review. Clinical Orthopedics and Related Research, 269, 162-173.

Gagey, O., & Hue, E. (2000). Mechanics of the deltoid muscle. A new approach. Clinical Orthopedics, 375, 250-257.

Gandevia, S.C. (1998). Neural control in human muscle fatigue: changes in muscle afferents, moto neurones and moto cortical drive. Acta Physiol. Scandinavia, 162, 275-283.

Gandevia, S.C., Enoka, R.M., McComas, A.J., Stuart, D.G. & Thomas, C.K (1995).. Neurobiology of Muscle Fatigue: Advances and Issues. Advances in Experimental Medicine and Biology, 384, 515-525.

Graichen, H., Bonel, H., Stammberger, T., Haubner, M., Rohrer, H., Englmeier, K., Reiser, M., & Eckstein, F. (1999). Three-Dimensional Analysis of the Width of the Subacromial Space in Healthy Subjects and Patients with Impingement Syndrome. American Journal of Roentgenology,172(4), 1081-1086.

Graichen, H., Stammberger, T., Bonel, H., Karl-Hans, E., Reiser, M., & Eckstein, F. (2000). Glenohumeral translation during active and passive elevation of the shoulder – a 3D open-MRI study. Journal of Biomechanics, 33(5), 609-615.

Greenfield, B.H., Donatelli, R., Wooden, M.J., & Wilkes J. (1990). Isokinetic evaluation of shoulder rotational strength between the plane of scapula and the frontal plane. The American Journal of Sports Medicine, 18(2), 124-128.

Halder A.M., Itoi, E., & An, K. (2000). Anatomy and Biomechanics of the Shoulder. Orthopedic Clinics of North America, 31(2), 159-176.

Harryman, D.T., Sidles, J.A., Clark, J.M., McQuade, K.J., Gibb ,T.D., & Matsen F.A. 3rd. (1990). Translationof the Humeral Head on the Glenoid with Passive Glenohumeral Motion. Journal of Bone and Joint Surgery, 72A, 1334-1343.

Hellwig, E.V., & Perrin, D.H. (1991). A comparison of two positions for assessing shoulder rotator peak torque: The traditional frontal plane versus the plane of the scapula. Isokinetics and Exercise Science, 1(4), 202-206.

Hinton, R.Y. (1988). Isokinetic evaluation of shoulder rotational strength in high school baseball pitchers. American Journal of Sports Medicine, 16, 274-279.

Hughes, R.E., & An, K.N. (1996). Force Analysis of Rotator Cuff Muscles. Clinical Orthopaedics and Related Research, 330, 75-83.

Hughes, R.E., Johnson, M.E., O'Driscoll, S.W., & An, K. (1999). Normative Values of Agonist-Antagonist Shoulder Strength Ratios of Adults Aged 20 to 78 Years. Archives of Physical Medicine and Rehabilitation, 80(10), 1324-1326.

Inman, V.T., Saunders, J.B., & Abbott, L.C. Observations of the Function of the Shoulder Joint. Journal of Bone and Joint Surgery, 26A, 1-31.

Jensen, M.P., & Karoly, P. (1992). Self-report scales and procedures for assessing pain in adults. In Turk and Melzack (Eds.), Handbook of Pain Assessment. (pp 135-151). Guilford Press.

Jensen, B.R., Pilegaard, M., & Sjogaard, G. (2000). Motor unit recruitment and rate coding in response to fatiguing shoulder abduction and subsequent recovery. European Journal of Applied Physiology, 83(2/3), 190-199.

Jobe, C.M. (1996). Superior Glenoid Impingement: Current Concepts. Clinical Orthopaedics and Related Research, 330, 98-107.

Karduna, A.R., Williams, G.R., Williams, J.L., & Ianotti, J.P. (1996). Kinematics of the Glenohumeral Joint: Influences of Muscle Forces, Ligamentous Constraints, and Articular Geometry. Journal of Orthopaedic Research, 14(6), 986-993.

Kent, B. (1971). Functional Anatomy of the Shoulder Joint. Physical Therapy, 51(8), 867-887.

Krakauer, J., & Ghez, C. (2000). Voluntary Movement. In In E.R. Kandel, J.H. Schwartz & T.M. Jessell (Eds.). Principles of Neural Science. (pp.756-779). McGraw Hill.

Kramer, J.F., & Ng, L.R. (1996). Static and Dynamic Strength of the Shoulder Rotators in Healthy, 45- to 75-Year-Old Men and Women. Journal of Orthopedic and Sports Physical Therapy, 24(1), 11-17.

Kriellaars, D.J. & Massey, K. (1999) Neuromuscular control of eccentric contractions. Proceedings of the Canadian Federation of Biological Sciences, Winnipeg, Manitoba.

Kronberg, M., & Brostrom, L.A. (1995). Electromyographic recordings in shoulder muscles during eccentric movements. Clinical Orthopaedics, 314, 143-151.

Kronberg, M., Brostrom, L.A., & Nemeth, G. (1991). Differences in Shoulder Muscle Activity Between Patients with Generalized Joint Laxity and Normal Controls. Clinical Orthopedics and Related Research, 269, 181-192.

Kronberg, M., Nemeth, G., & Brostrom, L.A. (1990). Muscle Activity and Coordination in the Normal Shoulder. An EMG Study. Clinical Orthopedics and Related Research, 257, 76-85.

Kuechle, D.K., Newman, S.R., Itoi, E., Niebur, G.L., Morrey, B.F., & An, K.N. (1997). Shoulder Muscle Moment Arms During Horizontal Flexion and Elevation. Journal of Shoulder and Elbow Surgery, 6(5), 429-439.

Kuhlman, J.R., Iannotti, J.P., Kelly, M.J., Riegler, F.X., Gevaert, M.L., & Ergin, T.M. (1992). Isokinetic and Isometric Measurement of Strength of External Rotation and Abduction of the Shoulder. Journal of Bone and Joint Surgery, 74(2)A, 1320-1333.

Laursen, B., Jensen, B.R., Sjogaard, (1998). G. Effect of speed and precision demands on human shoulder muscle electromyography during a repetitive task. European Journal of Applied Physiology, 78, 544-548.

Lattanzio, P.J., Petrella, R.J., Aproule, J.R., & Fowler, P.J. (1997). Effects of fatigue on knee proprioception. Clinical Journal of Sports Medicine, 7, 22-27.

Loeb, G.E., & Gez, C. (2000). The Motor Unit and Muscle Action. In E.R. Kandel, J.H. Schwartz & T.M. Jessell (Eds.). Principles of Neural Science. (pp. 674-693). McGraw Hill.

Makela, M., Heliovaara, M., Sainio, P., Knekt, P., Impivaara, O., & Aromaa, A. (1999). Shoulder joint impairment among Finns aged 30 years or over: prevalence, risk factors and co-morbidity. Rheumatology, 38(7), 656-62.

Malerba, J.L., Adam, M.L., Harris, B.A., & Krebs, D.E. Reliability of Dynamic and Isometric Testing of Shoulder External and Internal Rotators. Journal of Orthopaedic and Sports Physical Therapy, 18(4), 543-52.

Mandalidis, D.G., Mc-Glone, B.S., Quigley, R.F., McInerney, D., & O'Brien, M. (1999). Digital Fluoroscopic Assessment of the Scapulohumeral Rhythm. Surgical and Radiologic Anatomy, 21(4), 241-246.

Mantone, J.K., Burkhead, W.Z. Jr., & Noonan, J. Jr. (2000). Non-operative treatment of rotator cuff tears. Orthopedic Clinics of North America, 31(2), 295-311.

Matsen, F.A., & Arntz, C.T. (1990a). Rotator Cuff Failure. In F.A. Matsen & C.A. Rockwood (Eds.) The Shoulder (pp. 647-677). Philadelphia, Penn: W.B. Saunders.

Matsen, F.A., & Arntz, C.T. (1990b). Subacromial Impingement. In F.A. Matsen & C.A. Rockwood (Eds.) The Shoulder (pp. 623-646). Philadelphia, Penn: W.B. Saunders.

McIntyre, J., Mussa-Ivaldi, F.A., & Bizzi, E. (1996). The Control of STable Postures in the Multijoint Arm. Experimental Brain Research, 110(2), 248-264.

McQuade, K.J., Dawson, J.D., & Smidt, G.L. (1998). Scapulothoracic Muscle Fatigue Associated with Alterations in Scapulohumeral Rhythm Kinematics During Maximum Resistive Shoulder Elevation. Journal of Orthopedic and Sports Physical Therapy, 28(2), 74-80.

McQuade, K.J., & Smidt, G.L. (1998). Dynamic Scapulohumeral Rhythm: the Effects of External Resistance During Elevation of the Arm in the Scapular Plane. Journal of Orthopedic and Sports Physical Therapy, 27(2), 125-133.

Miller, R.G., Kent-Braun, J.A., Sharma, K.R., Weiner, M.W. (1995). Mechanisms of human muscle fatigue. Quantitating the contribution of metabolic factors and activation impairment. Advances in Experimental Medicine and Biology, 384, 195-210.

Neer, C.S. II. (1983). Impingement lesions. Clinical Orthopedics, 173, 70-77.

Nirschl, R.P. (1989). Rotator Cuff Tendinitis: Basic Concepts of Pathoetiology. Instructional Course Lectures, 38, 439-445.

Nitschke, J.E., McMeeken, J.M., Buny, H.C., & Matyas, T.A. (1999).When is a change a genuine change? A clinically meaningful interpretation of grip strength measurements in healthy and disabled women. Journal of Hand Therapy, 12(1), 25-30.

Oxford, K.L. (2000). Elbow Positioning for Maximum Grip Performance. Journal of Hand Therapy, 13(1), 33-36.

Pagnani, M.J., Deng, X.H., Warren, R.F., Torzilli, P.A., & O'Brien, S.J. (1996). Role of the long head of the biceps brachii in glenohumeral stability: A biomechanical study in cadavera. Journal of Shoulder and Elbow Surgery, 5(4), 255-262.

Palmerud, G., Fosrsman, M., Sporrong, H., Herberts, P., & Kadefors, R. (2000) Intramuscular pressure of the infra- and supraspinatus muscles in relation to hand and arm posture. European Journal of Applied Physiology, 83(2/3), 223-230.

Pearson, K., & Gordon, K. (2000). Spinal Reflexes. In E.R. Kandel, J.H. Schwartz & T.M. Jessell (Eds.) Prinicples of Neural Science. (pp.713-735). McGraw Hill.

Pederson, J., Lonn, J., Hellstrom, F., Djupsjobacka, M., & Johansson, H. (1999). Localized muscle fatigue decreases the acuity of the movement sense in the human shoulder. Medicine and Science in Sports and Exercise, 31(7), 1047-1052.

Perry, J., Barnes, G., & Merson, J. (1989). Normal Electromyographic Values of Six Shoulder Muscles During Free Motion. Orthopedic Transcripts, 13, 322.

Pincivero, D.M., Aldworth, C., Dickenson, T., Petry, C., & Shultz, T. (2000). Quadricpes-hamstring EMG activity during functional closed kinetic chain exercise to fatigue. European Journal of Applied Physiology and Occupational Physiology, 8(6), 504-509.

Poppen, N.K., & Walker, P.S. (1978). Forces at the Glenohumeral Joint in Abduction. Clinical Orthopedics and Related Research, 135, 165-170.

Rodosky, M.W., Harner, C.D., & Fu, F.H. (1994). The Role of the Long Head of the Biceps Muscle and Superior Glenoid Labrum in Anterior Stability of the Shoulder. American Journal of Sports Medicine, 22(1), 121-130.

Sahlin, M., Tonkonogi, M., & Soderlund, K. (1998). Energy supply and muscle fatigue in humans. Acta Physiol. Scandinavia, 162, 261-266.

Schnoz, M., Laubli, T., & Krueger, H. (2000). Co-activity of the trapezius and upper arm muscles with finger tapping at different rates and trunk postures. European Journal of Applied Physiology, 83, 207-214.

Scoville, C.R., Arciero, R.A., Taylor, D.C., & Stoneman, P.D. (1997). End Range Eccentric Antagonist/concentric Agonist Strength Ratios: a New Perspective in Shoulder Strength Assessment. Journal of Orthopedic and Sports Physical Therapy, 25(3), 203-207.

Sharkey, N.A., & Marder, R.A. (1995). The Rotator Cuff Opposes Superior Translation of the Humeral Head. American Journal of Sports Medicine, 23(3), 270-275.

Sieck, G.C., & Prakash, Y.S. (1995). Fatigue at the Neuromuscular Junction. Advances in Experimental Medicine and Biology. 384(5), 83-100.

Sjogaard, G., Kiens, K., Jorgensen, K., & Saltin, B. (1986). Intramuscular pressure, EMG and blood flow during low-level prolonged static contraction in man. Acta Physiol. Scan, 128, 475-484.

Sjogaard, G., & Sogaard, K. (1998). Muscle Injury in Repetitive Motion Disorders. Clinical Orthopaedics, 351, 21-31.

Soslowsky, L.J., Carpenter, J.E., Bucchieri, J.S., & Flatow, E.L. (1997a). Biomechanics of the rotator cuff. Orthopedic Clinics of North America, 28(1), 17-29.

Soslowsky, L.J., Malicky, D.M., & Blasier, R.B. (1997b). Active and Passive Factors in Inferior Glenohumeral Stabilization: a Biomechanical Model. Journal of Shoulder and Elbow Surgery, 6(4), 371-379.

Soslowsky, L.J., An, C., DeBano, C., Carpenter, J.E. (1996). Coracoacromial Ligament: In situ Load and Viscoelastic Properties in Rotator Cuff Disease. Clinical Orthopaedics and Related Research, 330, 40-44.

Speer, K.P. (1995). Anatomy and Pathomechanics of Shoulder Instability. Clinics in Sports Medicine, 14(4), 751-760.

Stephenson, D.G., Lamb, G.D., Stephenson, G.M.M. & Fryer, M.W. Mechanisms of excitation-contraction coupling relevant to skeletal muscle fatigue. Advances in Experimental Medicine and Biology, 384(2), 45-56.

Sterner, R.L., Pincivero, D.M., & Lephart, S.M. (1998). The effects of muscular fatigue on shoulder Proprioception. Clinical Journal of Sports Medicine, 8(2), 96-101.

Tamaki, H., Kitada, T., Akamine, F., Murata, F., Sakou, T., & Kurata, H. (1998). Alternate activity in the synergistic muscles during prolonged low-level contractions. Journal of Applied Physiology, 84(6), 1943-1951.

Taylor, J.L., Butler, J.E., & Gandevia, S.C. (2000). Changes in muscle afferents, motoneurons and motor drive during muscle fatigue. European Journal of Applied Physiology, 83, 106-115.

Timm, K.E. (1997). The Isokinetic Torque Curve of Shoulder Instability in High School Baseball Pitchers. Journal of Orthopedic and Sports Physical Therapy, 26(3), 150-154.

Uhthoff, H.K., & Sano, H. (1997). Pathology of Failure of the Rotator Cuff Tendon. Orthopedic Clinics of North America, 28(1), 31-39.

Warner, J.J., Micheli, L.J., Arslanian, L.E., Kenndey, J., & Kennedy, R. (1990). Patterns of Flexibility, Laxity, and Strength in Normal Shoulders and Shoulders with Instability and Impingement. American Journal of Sports Medicine, 18, 366-375.

Westerblad, H., Allen, D.G., Bruton, J.D., Andrade, F.H., & Lannergen, J. (1998). Mechanisms underlying the reduction of isometric force in skeletal muscle fatigue. Acta. Physiol. Scandinavia, 162, 253-260.

Wickham, J.B., & Brown, J.M. (1998). Muscles within Muscles: the Neuromotor Control of Intra-muscular Segments. European Journal of Applied Physiology and Occupational Physiology, 78(3), 219-225.

Wilk, K.E., Andrews, J.R., & Arrigo, C.A. (1995). The abductor and adductor strength characteristics of professional baseball pitchers. American Journal of Sports Medicine, 23(3), 307-311.

Wilk, K.E., Andrews, J.R., Arrigo, C.A., Keirns, M.A., & Erber, D.J. (1993). The strength characteristics of internal and external rotator muscles in professional baseball pitchers. American Journal of Sports Medicine, 21(1), 61-66.

Wiker, S.F., Chaffin, D.B., & Langolf, G.D. (1989). Shoulder posture and localized muscle fatigue and discomfort. Ergonomics, 32(2), 211-237.

Windhorst, U., & Boorman, G. (1995). Overview: potential role of segmental motor circuitry in muscle fatigue. Advances in Experimental Medicine and Biology, 384, 241-258.

Yamaguchi, K., Sher, J.S., Andersen, W.K., Garretson, R., Uribe, J.W., Hechtman, K., & Neviaser, R.J. (2000). Glenohumeral motion in patients with rotator cuff tears: a comparison of asymptomatic and symptomatic shoulders. Journal of Shoulder and Elbow Surgery, 9(1): 6-11.

APPENDIX A – POWER ANALYSIS

Power Analysis Equation

$n = (PI \times \sigma/\mu)^2$

PI=Power Index

σ = standard deviation (percent)

μ = relative effect (percent)

n = number of subjects

1. The following calculations are based on preliminary data:

$N = (3.28 \times 16/20)^2$

$N = (3.28 \times 0.8)^2$

N = 6.82

Where, 16 is equal to the maximum standard deviation (%) of internal and external rotator peak moment strength testing from the preliminary data, and 20 is the mean deficit (%) of internal and external rotator peak moment following the fatigue task protocol during preliminary testing.

2. The following calculations based on published standard deviation data (Kramer and Ng 1996, and Kuhlman et al. 1992) for isometric testing of the internal and external rotators, and preliminary relative difference data:

$N = (3.28 \times 20/20)^2$

$N = (3.28 \times 1)^2$

N = 10.75

APPENDIX B- PARAPHRASE & INFORMED CONSENT FORM

"Differential fatigability of the glenohumeral rotators during repetitive grip tasks"

Paraphrase and Informed Consent Form

University of Manitoba 2000

Human Performance Laboratory

School of Medical Rehabilitation

Contact: Dr. Dean Kriellaars

787-2289

Paraphrase

Injury to the rotator cuff musculature around the shoulder is a primary cause of upper limb pain, especially in those over the age of 40. Our understanding of the strength and fatigability of the muscles surrounding the shoulder, the arm, and the wrist, during hand and forearm activities is largely unknown due to the lack of information available in the literature. The rotator cuff muscles are important for keeping the shoulder sTable during movement of the arm. The availability of specific equipment in our lab provides us with the ability to perform reliable and accurate evaluation measurement of shoulder muscle strength and fatigability in a safe and reproducible manner. This study is aimed at understanding how these shoulder muscles respond to repetitive arm activities. This information will assist in the development of both preventative and rehabilitative strengthening programs for the shoulder.

Procedure

As a subject in this study, you will be asked to perform a test on a special device for measurement of the strength of your hand and shoulder muscles. These tests will require that you provide maximal effort during the procedure. You will also be asked to perform a task requiring you to squeeze a special device, called a hand grip dynamometer with your hand repetitively until you become fatigued. This test will require maximal effort during the procedure as well. Prior to your participation in the testing you will be familiarized with the special equipment. You will be asked to return one week later to perform the same tests except the position of your arm will be different. Each testing session will last no longer than one hour.

You will be asked to perform any of the following tasks:

1. Perform maximal grip strength test with the handheld dynamometer. This will be achieved by squeezing the handheld dynamometer as hard as you can. This will be repeated twice.

2. Perform maximal shoulder internal and external rotation using the Kin-Com dynamometer. This will be achieved by pushing your arm into the foam-padded armrest on the dynamometer. This will also be repeated twice for each motion. You will be instructed by the investigators to gradually increase the force of your contraction until you reach your maximum and hold it for 2 seconds.

3. Repeat grip strength testing using the handheld dynamometer in a repetitive manner 5seconds on, 5 seconds off until the strength measurement falls to half of your original, maximal value.

4. Repeat shoulder internal and external strength measurements as above immediately after completing the grip strength tasks to measure the fatigue of the shoulder muscles.

You will also be asked to return about 1 week (5 to 10 days) later to repeat the above procedure in exactly the same manner as above. The only difference will be the position that your arm is in while you do the repetitive grip strength testing.

Risks

There are no major risks associated with this study. After maximal exertion of the muscles surrounding the shoulder you may experience some mild discomfort in the shoulder, arm or wrist that may last up to 72 hours after the test. This is a normal consequence, and it will resolve on its own. A certain amount of discomfort may be associated with the test, however, if obvious pain arises at any instance during the test, it will be discontinued. Testing can be stopped at any time during study participation, and you can withdraw from the study at any point without prejudice. There have been no documented cases of muscle tearing with this testing, however, the possibility still exists that a tear could occur.

You will not be identified in any published report of the results of this study. You are not paid to participate in this project, however you are free to withdraw at any time without prejudice.

If you have any questions or do not understand any aspect of this form, please ask the investigators now or contact,

Dr. Dean Kriellaars
School of Medical Rehabilitation
University of Manitoba
Voice: 787-2289

"Differential fatigability of the glenohumeral rotators and Forearm flexors during repetitive grip tasks"

Paraphrase and Informed Consent Form

University of Manitoba 2000

Human Performance Lab

School of Medical Rehabilitation

Contact : Dr. Dean Kriellars

787–2289

Consent Form

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

I understand that I may withdraw from this study at any time without prejudice and that by signing this form that I do not waive any of my legal rights, and I am free to contact the Research Ethics Board at 789-3389 for further information regarding my individual rights.

I am satisfied that my name will only be used on this consent form which will be kept in a locked filing cabinet in a locked office at all times, and that all future references to me and my participation in this study will be made by coded entry.

Subject (Printed)______________________________

Date____________________________

(Signed)______________________________

Witness (Printed)______________________________

Date____________________________

(Signed)______________________________

Investigator

(Printed)___________________________Date____________________________

(Signed)___________________________

APPPENDIX C - PATIENT QUESTIONNAIRE

Differential fatigability of the glenohumeral rotators during repetitive grip tasks.
Initial Subject Evaluation Questionnaire.

Patient Name____

Date of Birth____________________ Sex: male_____female_____

Occupation____

What types of physical activity do you participate in?

____

____

How often do you participate in these activities?____________________

Do you participate in any competitive sports?

Yes________ No____________________

If yes , please specify____

Have you ever hurt your right or left shoulder, arm, wrist, or hand?

Yes________________ No__________________

If yes, please specify____

Have you ever had surgery on your right or left shoulder, arm, wrist, or hand?

Yes________________ No____________________

If yes, specify____

Do you experience any restrictions in movement of your upper limbs?____________

Yes____________ No_________________

If yes please specify.

____

____

Have you ever used performance enhancing drugs?

Yes_________________ No____________________

If yes, specify____

Have you have any medical conditions over the past 12 months that have required you to see a doctor?

Yes_________________ No____________________

If yes, specify____

Do you have any other medical problems?

Yes_________________ No____________________

If yes, specify____

Which arm do you throw a ball with? Right_________ Left____________

APPENDIX D – RECRUITMENT LETTER

RESEARCH STUDY

Participants are required for a study examining:

Mechanisms of Rotator Cuff Fatigue.

Subjects needed are:

Healthy males between the ages of 18 and 50 years old who do **not** participate regularly in competitive sports or upper body weight training.

Subjects will be required to attend for two half-hour sessions, one week apart.

Location: Human Performance Laboratory, third floor Rehabilitation Hospital

If you are willing to participate in this study please contact:

Maureen Walker

APPENDIX E – VAS SCORING SHEET

SUBJECT:______________________________

DATE:________________________________

POSITION:_____________________________

SHOULDER

No Pain   Worst Pain Imaginable

←――――――――――――――――――――――――――→

FOREARM/HAND

No Pain   Worst Pain Imaginable

←――――――――――――――――――――――――――→