The Effect of Movement Strategy and Elastic Starting Strain on Shoulder Resultant Joint Moment during Elastic Resistance Exercise

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Abstract

Purpose: The purpose of this study was to compare the shoulder resultant joint moment (RJM) during a shoulder internal rotator exercise using elastic resistance employing four different movement strategies and two different starting elastic strains.

Methods: Ten subjects aged 27.4 ± 2.6 yr (5 female and 5 male) with no previous shoulder pathology performed four sets of six repetitions of shoulder rotation though 180° using elastic resistance (Thera-Band® elastic band, blue) during two acceleration (medium and low) and two cadence (2s:2s, <1s:1s) strategies at 0% elastic starting strain. The acceleration movement strategies were also performed with starting strain of 30%. A mathematical model using Newtonian mechanics was used to compute the RJM. Elastic band recoil force was measured with a force transducer. Forearm acceleration was determined by a miniature uniaxial accelerometer secured at the wrist. Electrogoniometer data were collected to determine the range of motion (ROM) as well as the angle between the forearm and band which was used to determine elastic moment arm. Paired t-tests were used to identify joint angle specific RJM differences between conditions.

Results: Angle specific comparisons revealed that RJM in the moderate acceleration movement strategy was significantly different from RJM in the low acceleration movement strategy through 150° (83%) of range of motion ($p<0.05$). Shoulder RJM was up to 111% higher in the moderate acceleration strategy ($P < 0.01$). Angle specific comparisons revealed RJM in the <1:1 cadence strategy was significantly different from RJM in the 2:2 cadence through 108° (60%) of the range of motion ($p<0.05$). RJM was up to 47% higher in the <1:1 cadence ($p<0.01$). RJM in the low acceleration strategy was significantly greater with 30% elastic start strain relative to 0% elastic start strain through 180º of angular excursion ($p<0.001$). The pattern and magnitude of neuromuscular loading was significantly different in higher acceleration movement strategies (moderate acceleration and fast cadence). **Conclusions:** These findings indicate that differential limb acceleration as a result of movement strategy significantly affects shoulder load during elastic resistance exercise. The pattern and magnitude of load was different in each movement strategy and could result in differential neuromuscular adaptation through training. Clinicians and exercise professionals should consider movement strategy/acceleration as an important factor when prescribing elastic resistance exercise for safety and efficacy.

Key Words: Acceleration, Cadence, Elastic Resistance, Resultant Joint Moment, Cueing/Movement Control Strategy

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Introduction

Strength and endurance are critical for our ability to function and are recognized as fundamental physical traits necessary for health and enhanced quality of life (Kraemer 2002). Ability to move is based on the capacity to generate external force using muscles under the control of the nervous system. The development of the neuromuscular system in terms of strength, endurance, and coordination are important in the areas of rehabilitation, fitness, and health. Resistance exercise has been shown to be the most effective method for developing strength, and it is currently recommended as part of a overall fitness and health program by major health organizations (ACSM, CSEP, NSCA, etc) (Kraemer 2002). Resistance training, aerobic endurance, and flexibility are recognized as the three key components of a comprehensive fitness program (Kraemer 2002).

In terms of resistance training, understanding the moment (torque) generating demands placed upon the body during resistance exercise would allow researchers to further characterize the dose/response relationship and allow rehabilitation and fitness practitioners to better utilize the known aspects of the dose/response relationship in exercise prescription. This enhanced understanding would make it possible to create resistance exercise programs that are more efficient, safe, and specific to individual capabilities. Optimally, the quantity of resistance utilized should closely match individual neuromuscular capabilities resulting in a load that is sufficient but not excessive for suitable tissue adaptation.

The use of elastic resistance (bands and tubing) forms an integral component of many resistance training programs (Hughes 1999; Patterson R.M. 2001; Simoneau G. S. 2001). Low cost, ease of use, portability, reliable static loading profile, and the ability to provide resistance independent of body segment orientation relative to gravity all contribute to the frequent use of this form of resistance in rehabilitation and fitness settings and in home and facility based exercise programs (Mikesky A. E. 1994; Page, P., and T. S. Ellenbecker. 2003). Studies have shown elastic resistance to be an effective method for increasing torque generating ability in diverse populations. Strength gains of 10%-15% measured with an isovelocity dynamometer in young healthy subjects after elastic resistance exercise three times a week for four to six weeks has been reported (Anderson 1992; Fornataro S. 1994; M. 1994; Manley M. L. 1999). Geriatric subjects demonstrated strength increases of 9%-22% measured with an isovelocity dynamometer after training with elastic resistance three times a week for twelve weeks (Heislein 1994; Mikesky A. E. 1994; Krebs 1998; Jette 1999). The gains observed have not yet been attributed to the joint torque generating requirements imposed by the elastic, as there has been no study which has documented joint torque (also known as resultant joint moment) in elastic exercise.

The use of elastic resistance has wide application in the rehabilitation setting. Some authors advocate elastic resistance training of the shoulder rotator cuff musculature after shoulder injury or shoulder surgery to provide increased glenohumeral stability through strength gains and to improve (normalize) scapulohumeral rhythm. Scapulohumeral rhythm is the relationship between scapulothoracic and glenohumeral motion, a 2:1 ratio is considered ideal for optimal upper extremity function (Brewster 1993; Wilk 1993; Hintermeister R. A., G. W. Lange, J. M. Schultheis, M. J. Bey, and R. J. Hawkins 1998). Enhanced performance measured by the velocity of tennis serve was reported after elastic resistance training three times a week for four weeks (Treiber 1998). Collegiate baseball players demonstrated a 20% increase in eccentric shoulder rotator cuff muscle torque measured with an isovelocity dynamometer after training three times a week for six weeks with elastic resistance (Page 1993). Elastic resistance exercise utilized for training the quadriceps and hamstring muscles after anterior cruciate ligament reconstruction surgery has been reported (Steadman J. R. 1989; Bynum 1995; Hintermeister R. A., M. J. Bey, G. W. Lang, R. J. Steadman, and C. J. Dillman. 1998). The authors concluded that this form of resistance imparted suitable progressive resistance to these muscle groups as measured by elecytromyographic activity of eight lower extremity muscles. Other reported areas of elastic exercise application include: patellofemoral dysfunction rehabilitation (Zappala 1992), affected limb strengthening after below knee amputation (Custon 1994), strength and proprioception training of unstable ankles (Docherty 1998), as one component in pulmonary (Debigara 1999) and cardiac rehabilitation (Verrill 1992; Christopherson 1998), and in stroke rehabilitation (Duncan 1998).

Despite the high level of use in clinical practice, there is very little information known about the pattern and magnitude of resistance provided from elastics through the range of motion of exercise. As one would expect, the recoil force of elastic resistance is known to have a strong linear relationship to elastic elongation (Hughes 1999; Labbe 2000; Patterson 2001), allowing clinicians to estimate the amount of applied elastic force during exercise based on the percent change in elastic strain. However, recoil force is only one of the components that determine the load (joint torque generating requirements) placed upon the body during elastic resistance exercise. The moment of elastic force is one of the torques that must be overcome during exercise; the moment of elastic force is simply the product of elastic force and elastic moment arm. The moment arm is the perpendicular distance from the line of action of the elastic band to the joint axis of rotation. The moment arm varies with joint angle and with the points of attachment of the elastic. The principal components that contribute to the joint torque or resultant joint moment generating requirements during elastic exercise are

1) the moment of elastic force,
2) the moment of weight of the limb, and
3) the product of acceleration of the limb and the moment of inertia of the limb.

Some studies have assessed the static components contribution to elastic loading (i.e. moment of elastic force), but no studies have considered all the parameters (the acceleration of the limb, the moment of inertia of the limb, and the moment of weight of the limb) that contribute to load during dynamic elastic resistance exercise. Some authors have recognized the moment of elastic force as one component of load during elastic resistance exercise. In a recent publication, Chris Hughes and colleagues (1999) examined the static moment (torque) created by elastic resistance about the shoulder. Importantly, this study recognized the significance of considering limb acceleration on the loading equation but only performed a static analysis during shoulder abduction. Simoneau et al (2001) determined the static loading about the elbow arising from elastic resistance (moment of elastic force) but also did not account for the effects of acceleration. The two previous studies only considered the moment of elastic force, thus omitting two important contributors (moment of weight and limb acceleration) to load during elastic resistance exercise. No studies to date have determined the resultant joint moments arising from elastic resistance during exercise. The relative contributions of the above parameters to resultant joint moment during different conditions of exercise are not known. A thorough understanding of the resistance training parameters that effect loading with elastic resistance will enhance training efficiency by enhancing knowledge of the dose/response relationship, and will also increase safety from a reduction in execution error.

Velocity of muscle contraction guided by the use of cadence control is one of the parameters considered and controlled for during dynamic resistance exercise training and prescription (Hay 1983; Morrissey 1995; Morrissey 1998; Kraemer 2002). The rate at which an external load is lifted affects the neuromuscular (adaptive and maladaptive) responses to resistance exercise (Kraemer 2002) and is believed to have an influence on the magnitude of strength gains induced from resistance exercise training. Typically, resistance exercise velocity is determined as the time rate change of limb or trunk angular displacement over a prescribed range of motion, usually determined in the concentric and eccentric phases of an exercise. Resistance exercise velocity therefore represents the average limb or trunk speed over the angular excursion of an exercise. As such, cadence control is a form of average velocity control. Despite frequent use of cadence control there is limited information on the pattern and magnitude of load induced by different movement velocities in normal resistance training settings (e.g. free weights, elastics, CAM based machines, etc) . Only one study to date (Hay 1983) has determined the resultant joint moment during resistance training at different cadences. In this study, subjects performed upper extremity elbow curl training using a barbell at slow velocity (three seconds to perform concentric contraction: two seconds to perform eccentric contraction), medium velocity (two seconds concentric contraction: two seconds eccentric contraction), and fast velocity (one second concentric contraction: two seconds eccentric contraction). A metronome was used to cue a cadence to obtain the desired average movement velocities. The mass on the barbell corresponded to the subject's 40%, 60%, and 80% four repetition maximum. A mathematical model and motion analyses camera were used to determine RJM. Results showed that the RJM/joint angle profile in fast velocity was significantly different from the slow and medium velocity at the ends of the exercise range of motion. Since then, no further studies have examined the effect of movement velocity on load during resistance exercise.

Elastic strain is a normalized measure of elastic elongation and is calculated by the change in elastic length relative to original elastic length multiplied by 100 (e.g. a two meter elastic stretched to five meters has undergone a 150 % strain: 3/2 x 100 = 150%). Elastic starting strains other than 0% (resting length) are discouraged during elastic resistance exercise as a form of elastic load progression (Page, P., Ellenbecker, T. 2003). It is stated that elastic start strain increases will result in a change in the kinematics and biomechanics of elastic resistance exercise resulting in less optimal elastic loading profiles relative to elastic resistance performed with 0% start strain (Page, P., Ellenbecker, T. 2003). There is limited data to support these recommendations.

There is a need for a study to examine all the aspects of loading arising during elastic resistance training (as identified above). Further, there is a need to systematically examine the variation in loading that occurs with 1) different movement strategies which include different movement velocities and with different accelerations and 2) different loads (starting stain of elastic band).

The purpose of this study was to quantify and compare the shoulder resultant joint moments during a shoulder rotation exercise using elastic resistance performed with four different movement control strategies and with two starting elastic strains. The four movement strategies were subcategorized into two themes; acceleration control and cadence control. The two acceleration strategies were 1) "dynamic" during which arm acceleration was moderately high, and 2) "controlled" during which arm acceleration was minimized. The two cadence strategies were 1) less than one second to one second (<1:1) and 2) two second to two second (2:2). The starting elastic strain was 0% for all four strategies. Thirty percent (30%) starting strain was also evaluated in the two acceleration strategies. 30% starting strain was chosen because it is utilized clinically during elastic resistance exercise, higher elastic starting strain s are not typically used. The relative contributions of the moment of elastic force and the acceleration dependant moment ($I\alpha$ component) to the shoulder resultant joint moment during the different movement strategies were computed. Shoulder rotation elastic exercise was chosen because it is commonly utilized in shoulder rehabilitation.

Literature Review

Resultant Joint Moment

All dynamic resistance exercise involves the movement of body segment(s) through a specified range of motion against some form of external resistance with the goal of improving neuromuscular strength, endurance, and coordination. Many different modes of resistance are available to provide external resistance including free weights (i.e. isotonic resistance), elastic bands and tubing, pulleys, isokinetic dynamometers, impulse inertial trainers, and cam based machines (e.g. Nautilus equipment). Resistance training programs use the aforementioned equipment in a variety of combinations and progressions as part of the overall therapy program.

In any therapeutic intervention, the dose/response relationship is important to establish for achievement of optimal results. In order to achieve the desired outcome, it is important to understand the parameters that can be varied and how they influence exercise dose. This knowledge is important to provide adequate stress to the neuromuscular system to induce suitable tissue adaptation. As important is the avoidance of overstressing tissues, which can delay recovery and result in re-injury. There are a very limited number of research studies documenting the dose (loading characteristics) of any of the resistance training equipment; in particular the least amount of information of literature exists around elastic resistance.

Load is defined as the instantaneous moment generating requirements about each joint engaged in exercise due to the motion of the weight and segments involved and any external forces other than gravity (Komi 1992; Lieber 1992).

In general for the use of elastic resistance for shoulder exercise, the moments acting on the shoulder consist of:

1) The moment of external elastic force (Melastic)

2) The resultant joint moment (RJM)

3) The acceleration dependent moment (moment of segment inertia (I) x angular acceleration of segment (α))

The Newtonian Equations of Motion governing the dynamic behavior of the model displayed below states that the sum of all the moments acting about a joint ($\sum$M) is equal to the product of the moment of inertia and angular acceleration of the segment(s) (Iα).

$$\sum \mathbf{M} = \mathbf{I}\alpha$$

$$\mathbf{RJM} + \mathbf{M}_{\mathbf{elastic}} = \mathbf{I}\alpha$$

Rearranging to solve for RJM we get

$$\mathbf{RJM} = \mathbf{I}\alpha - \mathbf{M}_{\mathbf{elastic}}$$

The RJM represents the torque generating requirements of all the tissues spanning a joint, in particular arising from agonist muscle. Knowing the RJM during every instant of a movement provides the best non-invasive depiction of load (Putnam 1991), also known as the load or loading profile. No studies to date have determined the RJM for elastic resistance in exercise. Two recent studies have begun to examine aspects of elastic loading characteristics (Hughes 1999; Simoneau G. S. 2001), by determining the moment of elastic force during elastic resistance exercise. However both of these studies did not consider all the facets that contribute to elastic resistance load.

Resistance Exercise Movement Velocity

The parameters most often considered and controlled in resistance exercise prescription include:

- Magnitude of external load. Typically expressed as a repetition maximum (RM), and determined as the maximum number of repetitions a particular load can be lifted safely with proper technique through the prescribed range of motion. Load is most often specified in an over-simplified manner as the mass of a weight or thickness of elastic band or tubing
- Type of muscle contraction: concentric, eccentric, or isometric
- Range of motion: Angular excursion of the limb(s) or trunk expressed in degrees or radians
- Velocity of limb or trunk movement: often expressed as cadence
- Number of Repetitions: Exercise volume = repetitions x resistance
- Number of Sets
- Rest period between sets, rest between exercises, rest between exercise sessions
- Frequency: Number of resistance exercise sessions per week

Movement velocity and muscle contraction velocity are both used in resistance exercise literature to describe the time rate change of angular displacement of the limb(s) or trunk over the range of motion of the exercise. Movement velocity represents average speed over the angular excursion of the exercise. Control of movement velocity in resistance exercise is typically guided by the use of a cadence strategy whereby the length of time to complete the concentric muscle contraction phase and the eccentric muscle contraction phase are directed by either a metronome of a self count method (e.g. two seconds concentric phase, two seconds eccentric phase) (Kraemer 2002). The influence of movement velocity on strength development has been studied. Strength gains resulting from constant speed (isokinetic) training have been shown to be primarily velocity specific (i.e. the training induced enhancements primarily occur at velocities at or below the training velocity (Lesmes 1978; Costill 1979; Behm 1991)).

The Overload Principle states that optimal tissue adaptation occurs when the torque produced from external resistance most closely matches individual torque generating capabilities through the range of motion of the exercise resulting in adequate neuromuscular stimulation. Results of studies examining non constant speed resistance exercise are equivocal in terms of the optimal movement velocity for inducing strength (torque generation) and functional performance gains (Morrissey 1998). Palmieri (1987) examined the effects of lower extremity training using Nautilus equipment in which two groups of subjects participated in a ten week training program performing lower extremity exercises with the same weight at durations of 0.75s or 2s during the concentric phase of the exercise. No differences in the 1RM squat or vertical jump distance were shown between the groups at the end of the training program. Young & Bilby (1993) failed to show a difference in 1RM squat, vertical jump distance, maximum isometric force, and thigh muscle girth between two groups of subjects who trained in barbell squat at “slow” vs. “fast” concentric movement strategy. Actual training velocities in the slow and fast conditions were not reported.

Conversely in a study by Jones et al (1999) university football players participated in a series of upper extremity free weight barbell exercises three times a week for fourteen weeks. All subjects performed the same number of repetitions and sets at 50%, 75% and 90% of their 10RM 5RM and 3RM. Subjects in the experimental group used a maximum acceleration movement strategy during the concentric phase whereas subjects in the control group used “conventional concentric velocity” during the concentric phase. Although the actual upper limb movement velocities were not reported, the average difference in movement velocity between the two groups was between 15%-17%. The group that trained at maximum acceleration demonstrated greater increases in upper extremity power as measured by a medicine ball throw distance and 1RM bench press compared to the group that trained at conventional concentric velocity. The authors concluded that training at “fast” velocities is more effective for improved performance in high level athletes than traditionally slower velocities. In another study (Morrissey 1998), subjects participated in squat barbell training three times a week for seven weeks performing three sets of eight repetitions with an 8RM load using either a 2s: 2s (50º/s average velocity) or 1s: 1s (100º/s average velocity) cadence. The group that trained at the 1s:1s cadence improved more on long jump and average hip torque and power as measured with video motion device compared to the group that trained at the slower cadence. No difference was reported between the groups in 1RM squat and maximum knee extensor isometric torque. No differences in performance in vertical jump were reported. There were velocity specific changes in isokinetic testing (i.e. 100º/s training group improved isokinetic strength most at 100º/s testing). LaChance et al. (1994) studied 75 healthy college aged males. Three groups of subjects performed the maximal number of repetitions of push ups and pull ups at either "fast self paced", 2s:2s cadence, or 2s:4s cadence. The greatest amount of repetitions and work were performed at the self paced speed followed by the 2:2 cadence. Lowest scores were achieved at the slowest cadence.

Only one study to data (Hay 1983) has determined the RJM during non constant speed resistance training using different cadences. In this study, three subjects performed upper extremity barbell curl trials at slow (3s:2s), medium (2s:2s), and fast (1s:2s) movement velocities at 40%, 60% and 80% of their 4RM. A metronome was used to control the cadences. A mathematical model and motion analyses camera were used to determine the RJM at the elbow joints. Results showed that RJM in the fast velocity condition was up to 31% higher (80 Nm vs. 105 Nm) than RJM in the medium and slow velocity conditions during the initial phase of the exercise (from 160º to 90º of elbow flexion), and was subsequently up to 70% lower (0 Nm vs. 70 Nm) from 90º to 30º of elbow flexion when the arm was decelerating as it came to a stop at repetition mid-point. No significant RJM differences were reported between the velocity conditions in the eccentric phase (all conditions were performed with a two second cadence in the eccentric phase). The implications of the different RJM/joint angle profiles between the velocity conditions were not discussed. No recommendations regarding constant speed resistance exercise training velocities were made either.

The American College of Sports Medicine (ACSM) recommends that untrained healthy adults use slow (2s:4s) or moderate (1-2s:1-2s) cadences in free weight resistance training, and that intermediate trained individuals use moderate velocity, and for advanced training the inclusion of slow, moderate, and fast (<1s:1s) cadences be used to maximize strength (Kraemer 2002). These guidelines are based on very limited and indirect evidence. The effect of movement velocity on RJM during elastic resistance exercise is unknown despite common use in clinical practice. There is a need for a study to determine the load induced by different resistance training velocities so that resistance exercise guidelines and recommendations can be improved.

Material Properties of Elastic Resistance

Elasticity refers to the property of a material that resists and recovers from a deformation. It is defined as the ability of material to return to an original resting position when stretched (Purvis 1997). The resting length of elastic is the position where the slack is taken up but no tension or stretch exists (Simoneau G. S. 2001). The tendency of an elastic material to return to its resting length when elongated is defined by Hooke's Law (below).

F=K x X

- F: Recoil force of band.
- X (strain): A normalized measure of elongation calculated by the change in tubing length/resting tubing length x 100.
- K (proportionality constant): The stiffness of the material, which is determined by the volume of latex in the elastic band.

The force/deformation relationship of elastics used for exercise has been shown to be strongly linear. Hughes et al (1999) calculated strong linear relationships (r^2 values 0.94 to 0.98) with Thera-Band® elastics, (**Figure 1**). Labbe et al (2000) calculated r^2 values of 0.95 to 0.99 (**Figure 1**). The slope of the force /deformation relationship for each color of elastic band is different. For each change in elastic band color there is a non-uniform change in recoil force. At the same strain levels, the range of the change in recoil force is 20%-30% between successive elastic band colors.

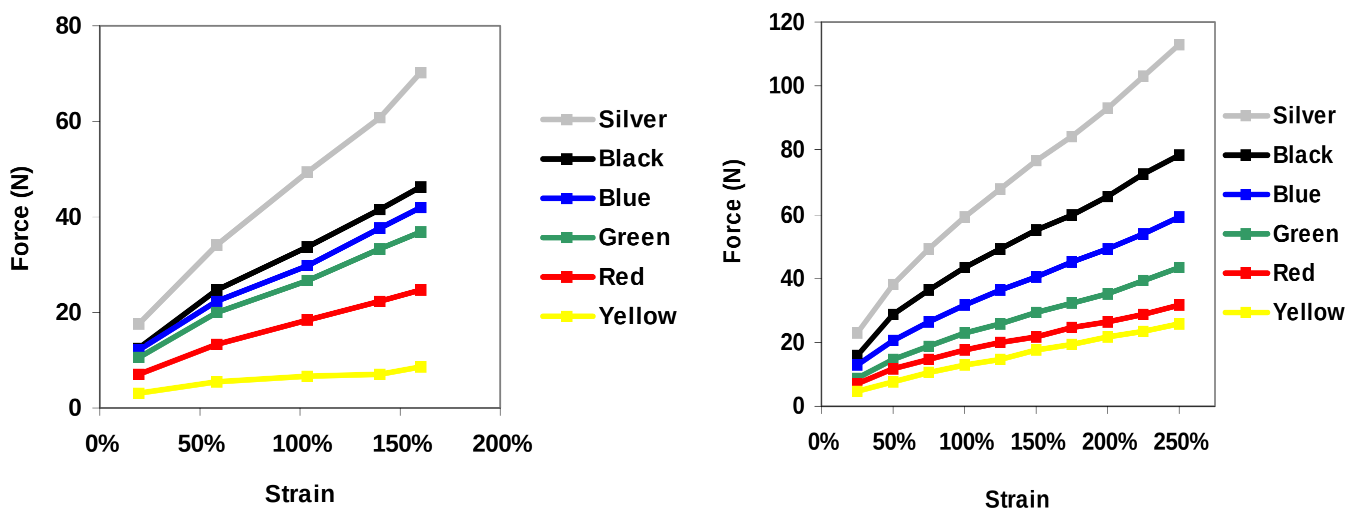


Figure 1: Elastic band recoil force in relation to percent length change of band for 6 band colors (yellow, red, green, blue, black, and silver). Data adapted from Hughes et al (1999) -right graph, and Labbe et al (2000). The thickness of band increases with corresponding higher force curves progressively from yellow to silver in legends above.

The rate of strain of elastic does not affect the recoil force of the material (Patterson 2001). The time dependent change in properties of elastics was determined by Patterson et al (2001). They calculated the decrease in recoil force to be less than 3% (1.1N) after 6,000 elongation cycles of between 100% to 200% strain of green elastic. They concluded that elastics commonly used in resistance exercise are highly fatigue resistant. They also concluded that the decrease in recoil force occurred during the initial 20 to 30 cycles of elongation and that clinically new elastic should be pre stretched 20 times to 100% strain prior to use to minimize any further force attenuation during exercise.

Elastic Loading

Hughes et al (1999) calculated the moment of elastic force produced by six different colors of elastic bands (yellow, red, green, blue, black, and silver) during shoulder abduction exercise. The moment of elastic force was calculated as the product of band recoil force (measured with a commercial strain gauge) and the moment arm of elastic band (the perpendicular distance from the line of action of the band to the shoulder joint axis of rotation). The moment arm of the elastic band was determined by a motion analysis system. Moment of elastic force were calculated statically at 30°, 60°, 90°, 120°, and 150° of shoulder abduction (**Figure 2**). The calculated resistance torque curves followed what the authors describe as an ascending-descending pattern (i.e. largest magnitude at 90º, lowest at 30º & 150º of shoulder abduction). They concluded that even though the elastic band recoil force increases progressively throughout the range of motion, the change in the magnitude of the elastic moment arm offsets the linearity and varies the moment of elastic force in an ascending–descending pattern. They also stated that the moment of elastic force profiles produced by elastics were similar to the most common strength curves produced by major muscle groups in the body identified by Kulig et al. (1984). Strength curves are the joint angle specific moment generating abilities of a muscle group. To allow comparison of the moment of elastic force with free weight (constant mass) resistance, the authors used the same moment arm values to compute static moment of force at the same joint angles with 5 lb and 10 lb weights. They demonstrated that the moment of elastic force and the moment produced from the 5 lb and 10 lb dumbbells were similar, as both displayed a general ascending– descending pattern. The authors admit to not using a comprehensive model for calculating resistive torque at the shoulder by omitting two terms 1) the angular acceleration of the arm about the shoulder joint and 2) the moment of weight of the limb about the shoulder. This limits the validity of the results because both of these factors would have an affect on shoulder RJM during this exercise. Based on Newtonian mechanics, the moment of the weight of the arm would contribute to resistive torque at the shoulder during static and dynamic conditions and the angular acceleration of the arm would contribute to shoulder load during all non-static conditions during shoulder abduction. The authors did recommend further investigation of RJM with elastic resistance exercise with accelerated movements.

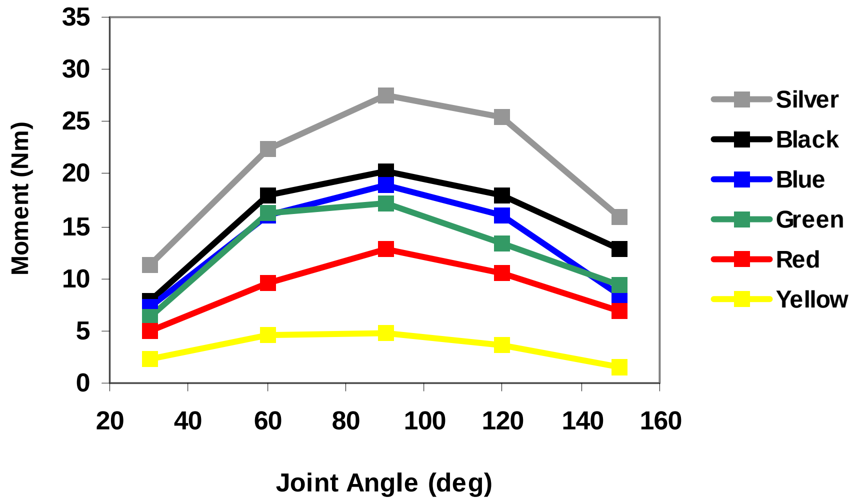


Figure 2: Moment of elastic force for six colors of elastic bands determined at five joint angles during shoulder abduction. A static model was used to determine the resistive torque. Data adapted from Hughes et al. (1999).

Simoneau et al (2001) used a static free body diagram method to calculate the moment of force at the elbow during a standing elbow flexion exercise using green elastic and a 1.81 kg (17.75 N) dumbbell. Moments of elastic and dumbbell force were calculated statically from 20° to 140° in 20º increments. Elastic force was determined by force/strain information provided by the manufacturer (The Hygenic Corporation, Akron, Ohio). The moment of the weight of the forearm and hand about the elbow joint was stated to be included in the moment calculations but it is unclear how the moment arms of these weight force vectors were measured. It is also unclear how the moment arms of the elastic and the dumbbell were measured. Forearm and hand masses were calculated using anthropometric measures of a single 667 N (68 kg) subject. The calculated moment of force values were similar for the elastic and dumbbell with both displaying an ascending-descending pattern with the largest moment magnitude at 100º of elbow flexion. The authors stated that the dumbbell provides a more progressive resistance to movement while the elastic material provides lower resistance in the initial range of elbow flexion movement and greater at the end range. The green tubing provided less average resistance through the range of motion than the 1.81 kg dumbbell. They made no specific recommendations based on these results. In this study, a comprehensive model for calculating resistive torque at the exercising joint was not used. The authors omitted the contribution from angular acceleration and moment of inertia of the forearm about the elbow by only using static calculations. Forearm acceleration would contribute to RJM requirements during non-static performance of this exercise. The moment of inertia (I) of the arm would also contribute to RJM about the elbow during this exercise. The moment of inertia of a body segment is the resistance of the segment to an angular change in motion (the angular equivalent of inertia). It is calculated by the product of the mass of the segment and the square of the distance from the centre of mass to the axis of rotation ($I=mr^2$). When the dumbbell is used, the moment of inertia of the arm segment would be higher because the centre of mass of the arm and dumbbell would shift distally in relation to the elbow and would increase the RJM required about the elbow. Apparent in their results below (**Figure 3**) is at 140°, the magnitude of the moment of elastic and dumbbell exceed the maximum RJM of the elbow flexors which would indicate the exercise could not be completed under these static loading conditions.

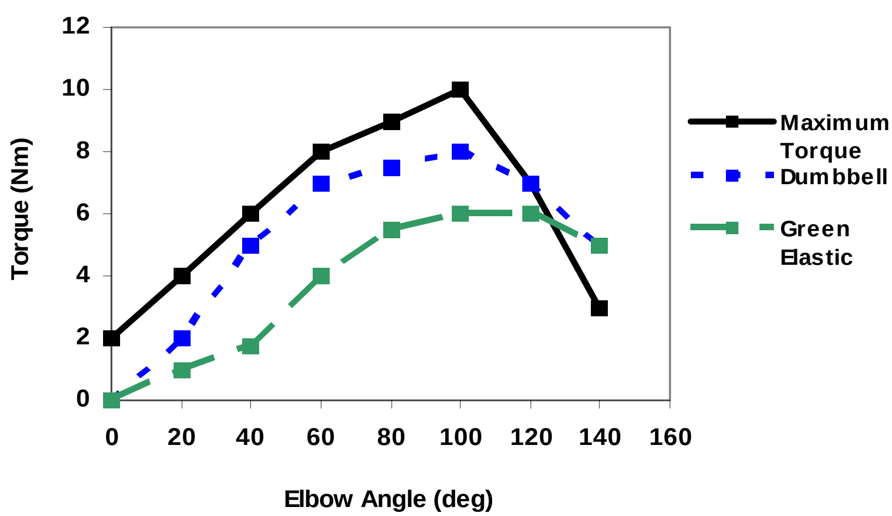


Figure 3: Moment about the elbow (y-axis) produced by green elastic tubing (dashed green line) and a 17.75-Newton dumbbell (dotted blue line) during elbow flexion exercise. Maximal voluntary torque curve of the elbow flexors is displayed in black (An et al 1989). Adapted from Simoneau et al (2001).

Electromyography during Elastic Resistance Exercise

The electromyography (EMG) of the elbow flexor muscles during an elbow flexion/extension exercise using two modes of resistance was compared by Lim and Chow (1998). An elastic tube and a dumbbell were used at two resistance intensities (10 RM and 20 RM). Two sets of five repetitions were done with each mode at the two resistance intensities (20 repetitions total with elastic and dumbbell). The EMG/time profiles recorded through the range of motion were shown to be different between the two modes. Higher elbow flexors muscle activity was observed with the dumbbell during the early ascending and late descending phases of the exercise. Higher elbow flexor muscle activity was observed with the elastic resistance during the late ascent and early descent phases of the exercise. The authors stated that the two modes of resistance induced different patterns of stress on the elbow flexor muscles based on the different EMG/time profiles. Movement control strategy during the exercise was not reported. Movement control strategy would determine forearm acceleration which contributes to elbow resultant joint moment. It is not clear whether the differences in the pattern and magnitude of EMG were due to the different modes of resistance or the potential difference in movement control strategy employed by subjects.

Acceleration and Human Movement

Accelerometry is the study of acceleration of an object or human body segment. Accelerometers have been previously used in human motion studies, and have recently been introduced into the realm of exercise analysis. Accelerometers have been used to measure head stability in the elderly (Keshner 2004) and as an indicator of lumbar spine stiffness during exercise (Colloca 2004). Many recent studies have used accelerometers to measure head and neck kinematics during motor vehicle impacts as an indicator of cervical spine trauma (McConnell 2003); (Olvey 2004); (Yang 2003); (Yoganandan 2002). Webber and Kriellaars (2004) used accelerometry to measure lumbar spine kinematics before and after lumbar stabilization instruction. Other reported areas of accelerometry in human motion analysis include: The measurement of limb tremor in neurological impaired patients (Tamas 2004); (Ushe 2004); (Golan 2004), in workplace safety to measure lumbar spine and seat vibrations of subjects in taxis, mining trucks, and helicopters (Kumar 2004);(de Oliveira 2004). Accelerometers have been used to measure activity levels with individuals suffering from low back pain (Liszka-Hackzell 2004), subjects with diabetes (Kirk 2004) and multiple sclerosis (Ng 1997), and in healthy adults (Mathews 2002). Accelerometers are also used in athletic product safety analysis. Caswell and Deivert (2002) measured the amount of impact force attenuation (peak acceleration reduction) of different lacrosse helmets during repetitive drops from 1.52 meters. Naunheim et al. (2002)

measured the impact attenuation capacity of different playing field surfaces as an indicator of football player injury risk by repetitively dropping accelerometers onto the surfaces. Naunheim et al. (2003) also measured headband effectiveness in terms of peak head acceleration attenuation during the heading of a soccer ball.

There are currently no studies that document the RJM at any joint during elastic resistance exercise that account for the moment of the weight of the moving body segment, the angular acceleration of the body segment, and the moment of elastic force. The two studies that have attempted to determine torque from elastic resistance are limited by only including the moment of elastic force in their calculations (i.e. static analysis only). Accelerometers have been shown to be highly sensitive to human motion and accelerometry provides a means by which to compute RJM during dynamic conditions.

Purpose

The purpose was to quantify and compare the shoulder resultant joint moment during an internal rotator shoulder exercise using elastic resistance employing four movement control strategies (moderate and low acceleration, and fast and slow cadence). In addition, the effect of the starting strain of the elastic on shoulder RJM was examined. This is the first study to document the instantaneous loading characteristics represented by RJM during elastic exercise. This is first study to utilize accelerometry in the calculation of load during resistance exercise.

Hypotheses

1. We predicted that the RJM during the moderate acceleration movement strategy would be higher than the RJM during low acceleration movement strategy at the initiation of the exercise due to acceleration effects. This is based on the prediction that forearm acceleration resulting from moderate acceleration cuing would be greater than forearm acceleration resulting from low acceleration cuing at the initiation of the exercise.
2. We predicted that the RJM in fast cadence movement strategy would be higher than the RJM in the slow cadence movement strategy at the initiation of the exercise due to acceleration effects. This is based on the prediction that forearm acceleration resulting from fast cadence cuing would be greater than forearm acceleration resulting from slow cadence cuing at the initiation of the exercise.
3. We predicted that the RJM during the end of the concentric phase and beginning of eccentric phase of movement would be diminished in the moderate acceleration movement strategy relative to the low acceleration movement strategy due to deceleration effects. This is based on the prediction that forearm deceleration resulting from moderate acceleration cuing would be greater than forearm deceleration resulting from low acceleration cuing during this phase of the exercise.
4. We predicted that the RJM during the end of the concentric phase and beginning of eccentric phase of movement would be diminished in the fast cadence strategy relative to the slow cadence strategy due to deceleration effects. This is based on the prediction that forearm deceleration resulting from fast cadence cuing would be greater than forearm deceleration resulting from slow cadence cuing during this phase of the exercise.

5. We predicted that the RJM in the moderate acceleration movement strategy would be higher than the RJM in the fast cadence movement strategy at the initiation of exercise due to acceleration effects. This is based on the prediction that forearm acceleration resulting from moderate acceleration cuing would be greater than forearm acceleration resulting from fast cadence cuing during this phase of the exercise.
6. We predicted that RJM at the beginning of the eccentric phase would be reduced in the moderate acceleration strategy relative to the fast cadence strategy due to deceleration effects. This is based on the prediction that forearm deceleration resulting from moderate acceleration cuing would be greater than forearm deceleration resulting from fast cadence cuing during this phase of the exercise.
7. We predicted that the RJM through the entire angular excursion of the exercise in the low acceleration strategy would be higher with the 30% elastic start strain compared to 0% elastic start strain.
8. We predicted that the RJM in the low acceleration strategy using blue elastic band with 30% start strain would be equivalent to RJM in the low acceleration strategy using a thicker band (black or silver) with 0% start strain.
9. We predicted that a change in movement strategy would not result from higher starting strain. Specifically, that the low acceleration strategy with 30% starting strain would not become a moderate acceleration movement pattern, and that moderate acceleration strategy with 30% starting strain would not become a higher acceleration pattern. This prediction is based on preliminary trials with cueing instructions using higher elastic loads (start strains).
10. We predicted that brief instruction followed by short demonstration and verbal cueing would be effective for acceleration control and would result in a clear distinction between low and moderate acceleration strategies.
11. We predicted that brief instruction followed by short demonstration and verbal cueing would be effective in achieving the desired cadence control.

Methods

Experimental Design

A group of ten subjects performed a shoulder rotation exercise using elastic resistance in six movement conditions (**Table 1**). The shoulder RJM at each joint angle was computed for each of the six movement conditions (4 movement strategies and 2 starting strains).

Table 1: The six movement conditions are defined as follows: LA – low acceleration, MA – moderate acceleration, <1:1 – less than one second to one second cadence, 2:2 – two second to two second cadence. Two elastic start strains were employed; 0% and 30%.

Condition	Strategy	Starting strain (%)
1	LA	0
2	MA	0
3	<1:1	0
4	2:2	0
5	LA	30
6	MA	30

Six repetitions of each condition were performed with a 120 second rest period between sets. Six repetitions were chosen to minimize any fatigue effects. In order to control for the possibility of an order effect, movement strategies were performed in balanced order. Five subjects performed the moderate acceleration and <1:1 movement strategies before the low acceleration and 2:2 movement strategies, five subjects performed the low acceleration and 2:2 movement strategies before the moderate acceleration and <1:1 movement strategies. Post hoc analysis revealed no order effect on the pattern and magnitude of RJM.

Subjects

A sample of convenience was recruited by word of mouth. Volunteers with a history of upper extremity musculoskeletal pathology or cardiovascular disease, or individuals currently experiencing any form of dominant arm shoulder pain were excluded from participation. Subjects were required to comfortably rotate their dominant shoulder through 90° of internal rotation from a position of 120º of external rotation (90º rotation = forearm perpendicular to the thorax). All subjects were right hand dominant as defined by the hand they would normally throw a ball. Before enrollment in the study, all subjects were provided an explanation of the purpose, procedures, and potential risks and benefits of the study. Ethical approval for this study was received by the Human Research Ethics Board of the Faculty of Medicine, University of Manitoba, Canada. All subjects provided their written informed consent before participation. All participants attended one session at the University of Manitoba Human Performance Laboratory, Winnipeg, Canada.

Each subject's body mass (kg) was measured (Tanita BWB-800 scale). Body height (m) was measured with a height measurement slider (Health O Meter scale, Continental, Chicago, ILL).

Mechanical Model and Equations

The in vivo neuromuscular demands of a particular exercise or task can be assessed non-invasively by constructing a mechanical model of the physical system in question and solving the inverse dynamics problem using Newtonian equations of motion to compute RJM (Andrews 1983); (Crowninshield 1981); (Putnam 1991).

In this study, the upper extremity was modeled as a system of two rigid links consisting of the elbow and wrist connected to a frictionless pin joint (shoulder) through the upper arm and constrained to move in a single horizontal plane. In order to simplify the computation, the internal rotation exercise was performed in a standing position with the exercising forearm parallel to the ground. The elbow and shoulder joints remained coplanar during the exercise, therefore the moment of the weight of the forearm and hand did not produce a flexion or extension moment about the shoulder and did not influence shoulder RJM. Shoulder internal and external rotation exercises are commonly prescribed in this manner.

The assumptions included in this model include:

1) The contribution of the antagonist neuromuscular activity (shoulder external rotators) was negligible and is not included in the calculation of RJM. There is no accepted method for accurately accounting for antagonist muscle activity during human motion due to the limitations of EMG analysis (Andrews 1983). It is suggested that moments produced from antagonist muscle activity will be significant during high velocity limb displacements under loaded conditions (e.g. to decelerate the limb) (Andrews 1983);(Hay 1983), and will increase as the velocity of movement increases, and decrease with increasing skill levels (Hay 1983).
2) The moments produced by bony contact and ligament forces about the shoulder was negligible and are not included in the calculation of RJM. Evidence suggests that bone contact and ligament tension only contribute significantly to RJM at the extremes or end

of exercise range of motion (Putnam 1991). Their contribution can only be estimated indirectly using some sort of modeling approach (Andrews 1983).

Resultant Joint Moment Determination

Resultant Joint Moment was determined by employing the following equation:

$$RJM = I\alpha - M_{elastic}.$$

I alpha ($I\alpha$) was determined as the product of angular acceleration (measured by accelerometer) and the moment of inertia of the upper limb segment (derived from body segment parameters). Angular acceleration (α) was derived by converting linear acceleration (m/s^2) measured by the accelerometer into angular form (rad/s^2) by dividing by the distance of the accelerometer to the elbow. The moment of elastic resistance was determined as the product of elastic recoil force (measured by the force transducer) and the moment arm of the elastic force about the shoulder joint axis of rotation. Pythagorean Theorem was employed to determine the magnitude of elastic band moment arm. The sine of the angle between the forearm and the coronal plane (angle measured by electrogoniometer) was multiplied by the elbow to fist length. Each of the measurement devices and techniques are described in detail below.

Electrogoniometer

An electrogoniometer (Model 7541B Bourns potentiometer) was employed to measure shoulder rotation range of motion and the angle between the forearm and the elastic band during the exercise. The electrogoniometer consisted of a rotational potentiometer with a scale factor of $15mV \bullet degree^{-1}$ of rotation. Two rigid plastic arms extended from the potentiometer. The potentiometer was secured over the center of an exercise handle (The Hygenic Corp., Akron, OH) which was gripped by subjects during the exercise. One end of the elastic band was secured to the exercise handle through a threaded hinge. Using adhesive tape, one arm of the electrogoniometer was secured in series with the radial surface of subject's forearm and the other electrogoniometer arm was secured to the dorsal surface of the exercise handle in series with the elastic band. Electrogoniometer signal was sampled at 200 Hz using a Data Translation Board 9800 series - 200Hz, 12 bit analog to digital converter using programmable data acquisition software (Scope, version 2.2.0.30.). Calibration of the electrogoniometer was achieved by recording the output at three known angles measured with a standard goniometer. The scale factor was derived using the known values (0º, 45º, 90º) and applied to the electrogoniometer data.

Accelerometer

A miniature linear, uniaxial accelerometer (Model EGAX-F10-/R, 15 x 13 MM, IC Sensors, Fairfield, NJ, USA) was used to measure forearm acceleration during the exercise. The accelerometer was secured to the skin over the posterior surface of the distal end of the radius with adhesive tape. This position provided a relatively flat surface which minimized accelerometer orientation changes in the frontal plane. With the forearm in anatomical neutral position, the accelerometer was sensitive to medial and lateral motion of the forearm about the frontal (twist) axis of the shoulder. The accelerometer was connected to an operational amplifier with gain and offset control. The amplified accelerometer signal was sampled at 200 Hz (Data Translation Board 9800 series, 12 bit analog to digital converter) using programmable data acquisition software (Scope, version 2.2.0.30.). The accelerometer was calibrated using the gravitational orientation method where the accelerometer was placed in three controlled orientations: vertical (known acceleration due to gravity -1 g), horizontal (0 g), and inverted horizontal (+1 g) with corresponding voltage output recorded. The scale factor and zero offset were derived using the known values (-1, 0, 1 g) recorded in this calibration file and applied using a linear equation to the digitized accelerometer data (Webber 2004).

Load Cell

A model 6001 S-Beam Load Cell Transducer and Strain Gauge Conditioner Module SGCM-401 (Intertechnology, Toronto, ON, Canada) was used to measure elastic band recoil force (N) during the exercise. The load cell had two threaded studs secured to either side of a central steel diaphragm that housed the transducer. The load cell was anchored to the vertical post of a standard four post universal barbell weight rack by means of a custom flexible metal wire harness looped through one of the threaded studs. One end of the elastic was attached to the other load cell threaded stud by means of a custom metal connecting clip. The load cell was able to pivot about the post and remain in series with the line of pull of elastic band during the exercise. In order to ensure that the elastic remained perpendicular to the forearm during the exercise the vertical position of the load cell on the rack was adjusted to the same height as each subject's forearm. This positioning is consistent with clinical practice (Page, P., Ellenbecker, T. 2003) and minimizes deviation of linear elastic line of pull on the load cell, that is it permits accurate determination of recoil force data based upon the cosine law force. Calibration of the load cell occurred before data collection began. Certified weights of 1 kg, 2 kg, and 5 kg were placed in series with the load cell in the gravitational plane. A derived scale factor was determined and applied to all force data. The amplified load cell signal was sampled at 200 Hz (Data Translation

Board 9800 series, 12 bit analog to digital converter) using programmable data acquisition software (Scope, version 2.2.0.30.).

Elastic Band

Eight different colors of Thera Band® elastic bands (The Hygenic Corp., Akron, OH) with increasing material diameter/thickness are available commercially (tan [extra thin], yellow [thin], red [medium], green [heavy], blue [extra heavy], black [special heavy], silver [super heavy], and gold [max]). Blue elastic band was used to provide resistance to shoulder internal rotation during the exercise. Elastic band blue was chosen because it is commonly used in clinical practice during shoulder rehabilitation and also represents an intermediate level of elastic resistance in relation to the other seven colors of elastic bands. The amount of resistance provided by the gold band would have likely been too great resulting in subject fatigue or an inability to perform the desired movement strategies with proper technique. The lowest level of resistance (tan) would likely have been too low and would not have provided sufficient load for subjects used in this study.

Each subject used a length of elastic that was equal to their elbow to fist length. This is consistent with recommended clinical practice guidelines (Page, P., Ellenbecker, T. 2003) to ensure that elastic length and the length of exercising limb ("lever arm") are equivalent thus preventing elastic strain from exceeding 200% during exercise. This length of elastic is also stated to provide an ascending–descending elastic torque profile during exercise which is desirable in order to match human strength curves (Page, P., Ellenbecker, T. 2003). In order to ensure consistent elastic force deformation relationships, all new elastics used in this study were subjected to twenty cycles of 200% strain prior to use to remove hysteresis.

Body Segment Parameters

The body segment parameters associated with the mechanical model included the moment of inertia of the upper arm, forearm, and hand about the shoulder axis of rotation. Published body segment parameter regression equations (Shan 2003) were used to determine the moment of inertia about the centre of mass for the upper arm, forearm, and hand of each subject based on gender, height and mass. The parallel axis theorem was used to establish the moment of inertia of these body segments about the shoulder joint. Moment of inertia values obtained with this technique were compared to moment of inertia values published by Chandler et al (1975). This comparison revealed a maximum difference of <5% which would not significantly alter the results in the study.

Range of Motion Determination

The law of parallel lines was applied to electrogoniometer data in order to determine the angle between the forearm and the coronal plane of the trunk (90º = forearm perpendicular to the coronal plane of trunk). Shoulder rotation range of motion and total angular excursion were determined in this manner. This surrogate measure of gleno humeral rotation angle is consistent with methods employed in the rehabilitation setting (Clarkson 1989).

Testing Protocol

Maximum Isometric Shoulder Internal Rotator Moment Test

A commonly used position for shoulder internal rotator strength testing was employed (Dvir 1995). In standing, subjects were positioned with the shoulder in the scapular plane (40° anterior to the frontal plane) with 0° of shoulder abduction and elbow flexed to 90°. The shoulder was then elevated to 45° and internally rotated to 60°. Prior to the test, a warm up consisting of a series of shoulder internal rotator contractions followed by one maximum effort practice trial were performed. A standard elastic exercise handle was secured to the load cell with a custom metal wire. The load cell was secured to the vertical pillar of a weight rack. Two test trials were then performed during which the subjects exerted a maximal effort pulling on the exercise handle against the load cell for five seconds followed by a rest period of 120 seconds. Strong verbal prompting was given to each subject during all the maximum isometric contraction trials. The internal rotator force moment arm was determined by measuring the perpendicular distance from the metal wire to the elbow joint with a standard measuring tape. Maximum isometric shoulder internal rotator moment was determined as the product of the force and moment arm. All values were reported in Newton meters.

Standard Shoulder Internal Rotator Exercise

Prior to performing the exercise protocol all subjects read a written description and viewed a demonstration of the exercises to be performed. A warm up consisting of ten minuets of dynamic bilateral circular shoulder rotations and static stretches of the shoulder internal rotator muscles was performed. A familiarization trail consisting of 15 to 20 repetitions of shoulder rotation using blue elastic band was performed incorporating all four of the movement strategies employed in the protocol. Verbal correction was provided from the investigators as necessary to ensure that the movement strategies were performed in the desired manner.

The exercise protocol consisted of four sets of six repetitions of shoulder rotation using blue elastic band through ~180º of angular excursion employing four different movement control strategies. In standing subjects were positioned with the left arm resting at the side and feet shoulder length apart. Right arm was positioned with the elbow flexed at 90º and the shoulder in 0º abduction and 120º of external rotation (90° external rotation corresponded to forearm perpendicular to the trunk). A standard manual goniometer with the axis positioned on the elbow olecranon process was used to verify shoulder rotation position prior to initiating the exercise.

Subjects then grasped the exercise handle with their right hand. Trunk distance from the barbell rack was adjusted so that the slack on the elastic band was taken up but there was not any elastic tension. A custom fabricated range of motion marker was positioned at the starting position in contact with the dorsum of the subject's right hand. Subjects were instructed to bring their hand as close as possible to the range of motion marker at the completion of each repetition. From the start position, subjects were instructed to pull on the band and internally rotate their shoulder until the forearm reached a point just prior to contacting their thorax. This position corresponded to full shoulder internal rotation through ~90° range of motion (from 120º to ~30º). Without pausing, subjects were instructed to then return the forearm back to the start position. Shoulder rotation exercise performed through this angular excursion is commonly employed in the rehabilitation setting during shoulder rehabilitation (Brewster 1993; Wilk 1993; Kisner 1996; Page, P., and T. S. Ellenbecker. 2003) particularly in the early phases of rotator cuff injury rehabilitation (Brewster 1993; Wilk 1993; Kisner 1996). Subjects were instructed to keep the medial region of their upper arm and the elbow in contact with the thorax as much as possible during the exercise. Subjects were instructed to keep their elbow flexed at 90º to ensure the forearm remained parallel to the ground. Instructions were also provided to maintain the forearm and wrist in neutral positions (i.e. minimize forearm supination and pronation, and wrist flexion and extension), and to avoid using any body compensatory movements (e.g. trunk rotation). Verbal correction was provided from the investigators if any deviations from these guidelines were observed during the trials, and trials were repeated if marked deviations were observed.

Movement Strategy Instruction

Prior to performing the exercises, subjects were given a script to read that described the methods of performing the four different movement strategies. For the low acceleration movement strategy, subjects were instructed to perform the movement in a manner similar to the smooth motion of a sweep second hand of a watch or clock. Subjects were instructed to "smoothly" internally rotate their arm towards their trunk, and just prior to contacting their trunk without pausing, "smoothly" return their arm back to the start position at the range of motion marker. For the moderate acceleration movement strategy, subjects were instructed to perform the movement in a manner similar to the motion of a step second hand of a watch or clock (i.e. "tick tock"). They were instructed to pull hard on the band at the beginning of the exercise to initiate the movement and then from the position of full shoulder internal rotation release the arm back quickly without pausing to return to the start position. Subjects were discouraged from performing the moderate acceleration strategy with maximum effort. In the two second to two second cadence strategy, subjects were instructed to complete full forearm internal rotation from the start position in two seconds (concentric contraction phase), followed by a two seconds to return the forearm from full internal rotation back to the start position (eccentric contraction phase). In the less than one second to one second cadence strategy, subjects were instructed to complete full forearm internal rotation from the start position in less than one second followed by a one second duration to return the forearm back to the start position. In order to simulate typical rehabilitation and home exercise conditions, subjects used a self count method to perform the cadence strategy rather than a metronome. The time interval between each repetition in all the movement strategies was not permitted to exceed five seconds. The rest interval between sets was set at 120 seconds consistent with clinical practice and recommended guidelines for inter set recovery in resistance exercise training (Kraemer 2002).

Data Analysis

Electrogoniometer, force, and acceleration data were acquired in Scope Software® (version 2.2.0.30) and exported to spreadsheet (Microsoft Excel®) for further analysis. A template was created in the spreadsheet for automated calculation of shoulder resultant joint moment.

Joint Angle Normalization

Range of motion markers substantially reduces the variability in range of motion between repetitions and subjects; however, small angular changes were evident between repetitions. These differences in angle covaried with repetition durations. Consecutive repetitions with durations of 0.95 s to 1.10 s (0.1 second is a 11% difference) would result in a twenty data point differential in number of angles recorded at a sampling rate of 200 Hz. Straight point by point averaging would result in skewed data, where the mean value would not represent the repetitions especially at the end points.

In order to control for this effect, the data was joint angle normalized, whereby all data was converted from a time series to joint angle series as outlined below:

I. All joint angle data was averaged to the nearest integer value which effectively reduces sampling rate by 3-4 times.

II. At each joint angle, Melastic, Iα and RJM values were averaged.

III. To account for epochs of high acceleration were data was not captured at integer joint angles a three point moving average was applied (this occurs periodically at the start and end of the data set in the moderate acceleration and <1:1 cadence strategies).

IV. At each joint angle data from six repetitions were averaged to a mean repetition value.

This resulted in repetition averaged data for each subject in each of the six movement conditions for

1. RJM,
2. Melastic, and
3. Iα.

The data reduction and processing sequence is shown in **Figures 4, 5, and 6**.

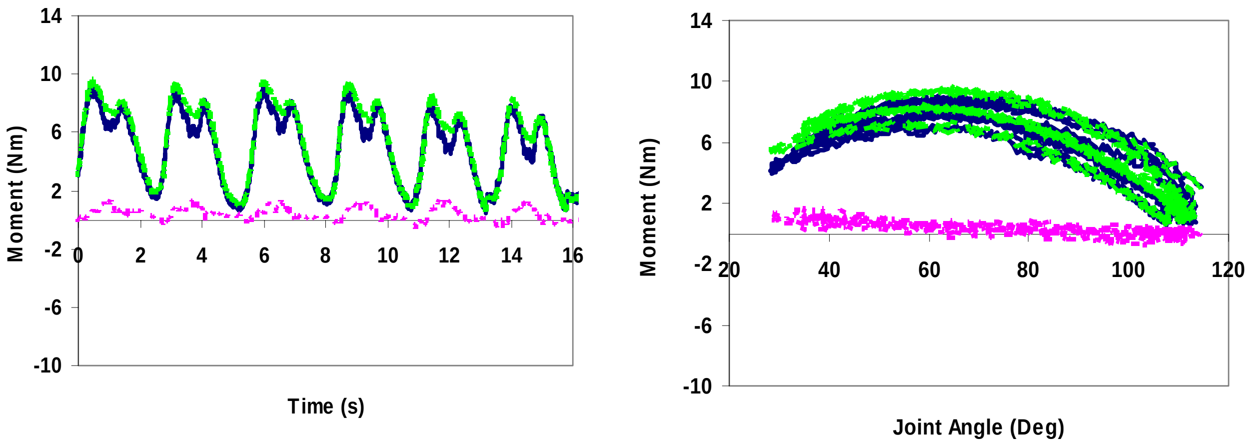


Figure 4: Moment data (one subject) for six consecutive repetitions of elastic resistance exercise employing a low acceleration movement strategy (0% elastic strain) represented in time series (left graph) and joint angle series (right graph). The data plotted versus joint angle allow all repetitions to be overlaid. Shown are the moment of elastic (dashed light green line), Iα (dotted light pink line), and the resultant joint moment (solid dark blue line).

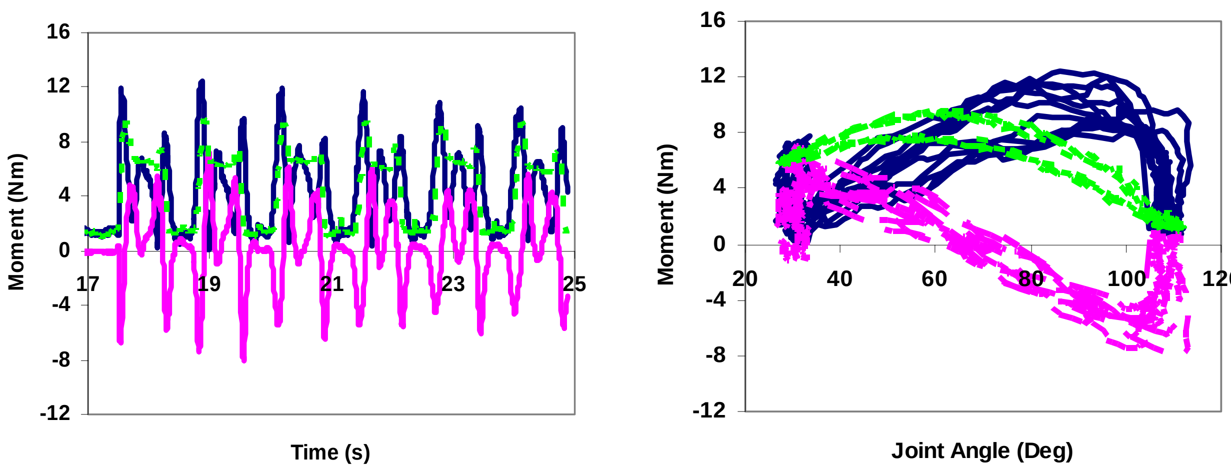


Figure 5: **Moment data from six consecutive repetitions of elastic resistance exercise from one subject employing a moderate acceleration movement strategy (0% elastic strain) represented in time series (left graph) and joint angle series (right graph). Shown are the moment of elastic (dashed light green line), Iα (dotted light pink line), and the resultant joint moment (solid dark blue line).**

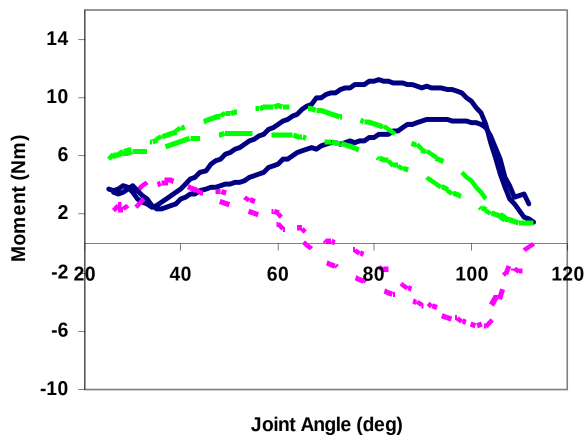


Figure 6: Average data from six consecutive repetitions of elastic resistance exercise from one subject employing moderate acceleration movement control strategy from Figure 5 above. Shown are the moment of elastic (dashed light green line), Iα (dotted light pink line), and the resultant joint moment (solid dark blue line).

Repetition Interval Determination

Force and acceleration data organized in time series format were examined in order to determine the 1) start, 2) mid, and 3) end point of each repetition. Repetition mid point occurs when the shoulder joint angle was at a position of maximal internal rotation and also corresponds to the period when shoulder internal rotators muscle activity transitions from primarily concentric to primarily eccentric.

Peak Iα

All repetitive or cyclical movements will demonstrate a fundamental four-phase acceleration pattern (Webber 2004). These four phases were readily observed in each repetition of all movement strategies of the elastic exercise (see **Figure 7** below). Nomenclature has been established (Webber 2004) to identify each phase in the acceleration waveform (P1, P2, P3, & P4). P1 and P4 were shoulder internal rotation directed accelerations (negative), and P2 and P3 were shoulder external rotation directed accelerations (positive). P1 represents forearm acceleration at the initiation of movement, P2 represents slowing down of the forearm towards full shoulder internal rotation. Both P1 and P2 occur as a result of concentric shoulder internal rotator muscle activity. P2 acceleration occurs opposite to the direction of shoulder motion (internal rotation). Displacement and acceleration do not always act in the same direction (Enoka 2002). P3 represents forearm acceleration at the initiation of shoulder external rotation from full internal rotation, P4 represents slowing down of the forearm it comes back to stop at the original starting position. Both P3 and P4 occur as a result of eccentric shoulder internal rotator muscle activity. The quantity of acceleration (acceleration magnitude x acceleration duration) required to slow the forearm down is equal and opposite to the quantity of acceleration required to speed the forearm up (i.e. P1 and P4 Iα = P2 and P3 Iα).

The minimum (P1 & P4) and maximum (P2 & P3) Iα values in each repetition were determined from the data using minimum and maximum lookup functions in the spreadsheet.

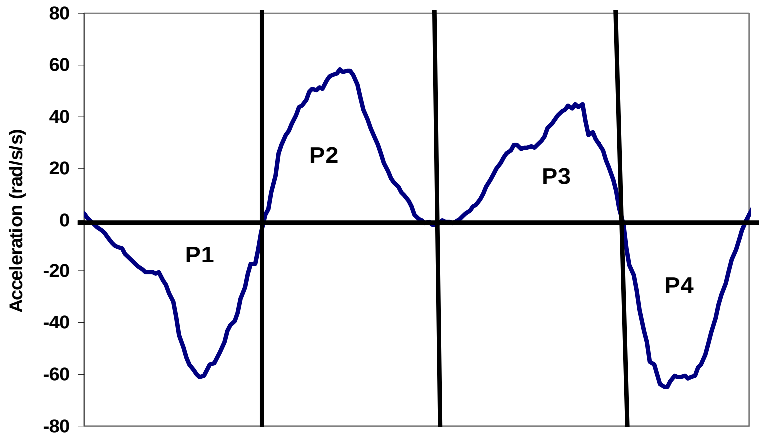


Figure 7: Forearm acceleration during one repetition of shoulder internal rotation exercise in the moderate acceleration movement strategy using blue elastic resistance. Four phases of forearm acceleration are labeled (P1, P2, P3, and P4) and are demarcated by black vertical lines.

Iα Influence on RJM

RJM was determined by the Newtonian equation; RJM = Iα – Melastic. When forearm acceleration was occurring in the direction of shoulder internal rotation (P1 and P4 Iα, negative acceleration in **Figure 7**), Iα adds to the moment of elastic to determine the required RJM. In this scenario the shoulder RJM required to produce forearm motion would be greater than the shoulder RJM required to overcome the elastic moment alone. When forearm acceleration was occurring in the direction of shoulder external rotation (P2 and P3 Iα, positive acceleration in **Figure 7**), Iα decreases from elastic moment to determine the required RJM. In this scenario the forearm is accelerating in the same direction that the elastic moment is acting (towards shoulder external rotation), and the shoulder RJM required to produce motion would be less than the RJM required to overcome the elastic moment alone. Higher positive acceleration (directed towards shoulder external rotation) decreases the magnitude of shoulder RJM required to produce motion because the forearm is accelerating in the same direction the elastic moment is acting and therefore the shoulder rotator muscles do not have to overcome the elastic moment but rather are required to control the forearm moving in the same direction the elastic is pulling.

Peak Resultant Joint Moment

Iα peaks at P1, P2, P3, and P4 resulted in corresponding RJM peaks which occurred at or near the same time period as Iα peaks. RJM peaks were temporally offset from Iα peaks as a result of the influence of elastic moment. P1 and P4 RJM in each repetition were looked up using the minimum and maximum function.

Mechanical Work Analysis

Average mechanical work (Joules) performed by the shoulder internal rotator muscles on the elastic band over the operational range of the exercise was determined by calculating the integral of resultant joint moment with respect to joint angle (area under the RJM curve).

Statistical Analysis

At each joint angle, the Melastic, Iα, and RJM data from each subject were arranged in a spreadsheet to compute the across subject joint angle average (Grand Mean). The individual data was preserved for within subject inferential testing. In order to compare changes in the Melastic, Iα, and RJM data between the six movement strategies, paired t-tests were performed at each joint angle using the individual subject data. This resulted in 180 paired t-tests results derived from ten comparisons (ten subjects) at each joint angle over the angular excursion of the exercise. That is there were 180 comparisons as there was a nominal 90° range of motion – 90°concentric and 90° eccentric (total of 180 joint angles).

Paired comparisons were performed between

1. Low and moderate acceleration strategies,
2. 2:2 and <1:1 cadence strategies,
3. <1:1 cadence strategy and moderate acceleration strategy, and
4. Low acceleration strategy with 0% and 30% elastic starting strain.
5. Moderate acceleration strategy with 0% and 30% elastic starting strain.

In this study, 180 paired t-tests were performed to identify regions of difference over the entire range of motion (180 degrees). Inherent to repeated inferential testing is the possibility that purely random effects will generate Type I errors (stating a significant difference exists when it actually occurred as a result of random variation between the two groups). This random effect is normally acceptable at the 0.05 level meaning that there is a 5% likelihood that a difference observed is simply due to chance. Since we are performing 180 inferential tests (a paired t-test at each joint angle over a 180 degree range of motion), there is an increased chance of Type I errors occurring. In order to effectively guard against committing a Type I error, a technique was employed to only accept trends in data that occurred over 3 consecutive points as opposed to accepting any single comparison showing a significant result at an individual joint angle. Using a mathematical probability model (3000 replications of 180 joint angles), we determined that the likelihood of having three consecutive random Type I errors (this would apply to Type II errors as well!) is 0.007%. For two in a row, the likelihood is 0.14%. Three was chosen as this effectively guards against accepting a Type I error. Using this approach we in fact are just examining the trend in the p-values as an indicator of the likelihood of Type I errors (or even Type II errors). There were strong trends in the patterns of the p-values; with no indication of meaningful random effects (i.e. the effect of random factors was negligible in the statistical interpretation of the data). If a traditional Bonferonni level was employed, a ridiculous significance level of 0.000278 (0.05/180) would need to be applied which would produce Type II errors. This technique has been previously employed in image analysis (Hiemstra 2000).

Results

The mean (±SD) physical characteristics for subjects (n=10) were 27.4 (2.6) years of age, 1.73 (0.1) m height and 70.4 (11.7) kg body mass. Five subjects (3 female and 2 male) had recent experience in upper extremity resistance training as defined by participating in upper extremity resistance training a minimum of 4 days/month within the past two months.

The results are organized in the following manner:

1) Low acceleration and moderate acceleration movement strategy comparisons
2) 2:2 and <1:1 cadence strategy comparisons
3) Moderate acceleration movement strategy and <1:1 cadence strategy comparisons
4) Low acceleration strategy: 0% and 30% elastic start strain comparisons
5) Moderate acceleration strategy: 0% and 30% start strain comparisons
6) Range of motion, repetition duration, and repetition velocity analysis

<u>Low Acceleration vs. Moderate Acceleration Movement Strategies</u>

1. <u>$I\alpha$</u>

$I\alpha$ in the moderate acceleration condition was significantly different from $I\alpha$ in the low acceleration strategy through 150° of 180° (83%) of angular excursion of the exercise (**Figure 8**). Non significant $I\alpha$ differences occurred:

i. At the beginning and end of the repetition (shoulder rotation ~115º-120º) where acceleration was close to zero
ii. Close to the mid point of the concentric and eccentric phases of the repetition where acceleration is at or close to zero. This occurs as the direction of forearm acceleration changes from negative to positive in the concentric phase (corresponding to a change from shoulder internal rotation directed acceleration to shoulder external rotation directed acceleration), and from positive to negative in the eccentric phase (corresponding to a change from shoulder external rotation directed acceleration to shoulder internal rotation directed acceleration).

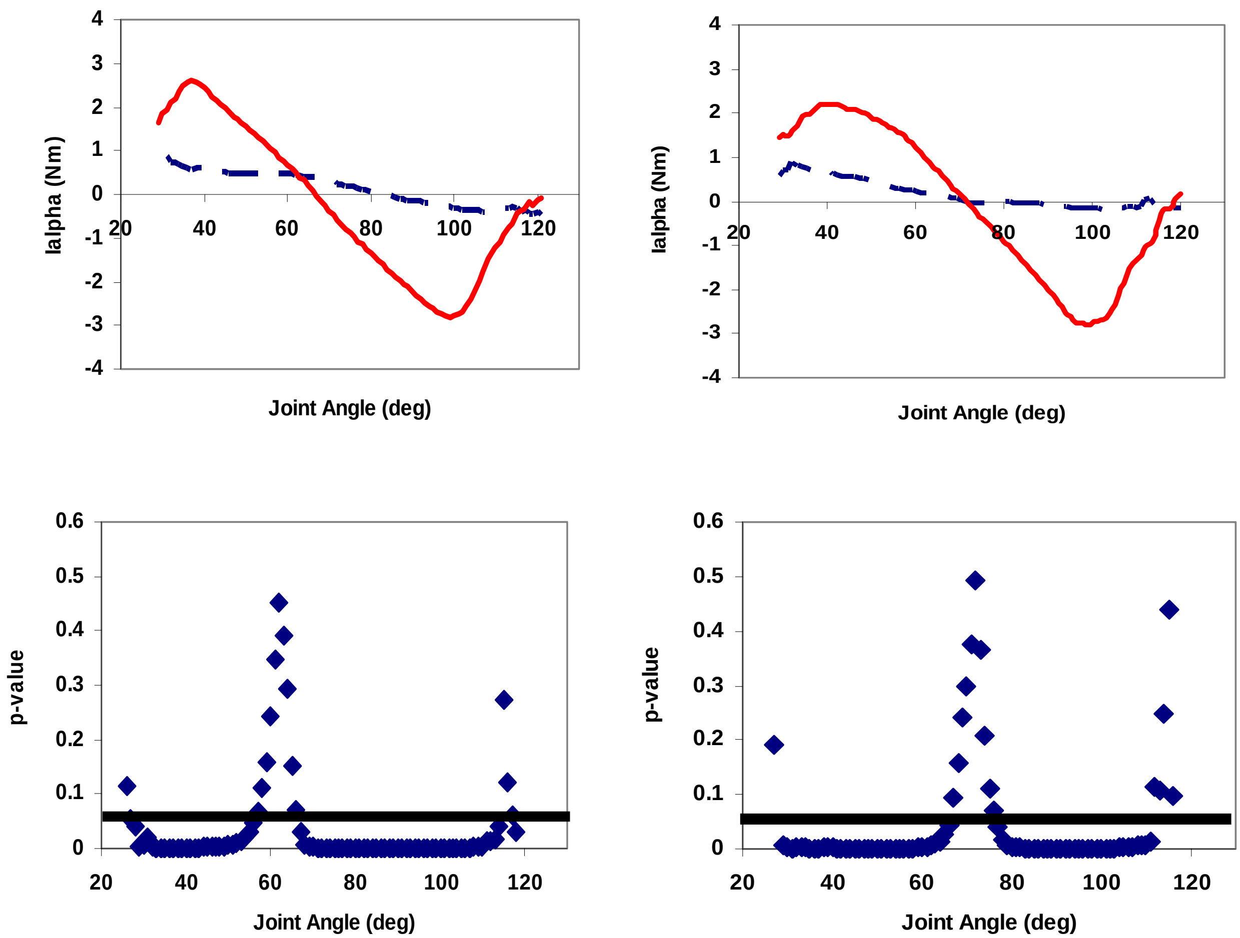


Figure 8: Average acceleration dependent moment (Iα) for the low acceleration movement strategy (dashed dark blue line) and the moderate acceleration movement strategy (solid light red line) during the concentric (left graph) and eccentric (right graph) phases of elastic resistance exercise. Results of joint angle matched paired t-tests are shown in the corresponding graphs below for the concentric and eccentric phases. Statistically significant Iα differences ($p<0.05$) between the moderate and low acceleration movement strategies are shown by scatter points below the wide black horizontal line at 0.05 and covers 84% (76° of 90º) of the comparisons in the concentric phase and 82% (74° of 90º) of the comparisons in the eccentric phase. Regions of non significance correspond to epochs of low acceleration magnitude at the ends at mid points of the repetition.

2. <u>Peak Iα</u>

Between repetition Iα peak differences were assessed using repeated measures ANOVA. No significant differences were found in either condition. Paired t-tests were then used to compare average Iα peaks between low acceleration and moderate acceleration strategies. All Iα peaks in the moderate acceleration strategy were significantly greater than corresponding Iα peaks in the low acceleration strategy (**Figure 9**) ($p<0.001$). Absolute values of peak P1 and P4 Iα were used in the comparisons. Between-strategy Iα peaks analysis provides information at four points (joint angles) of the exercise. This analysis alone would fail to show potential between-strategy Iα differences occurring at 176 joint angles (176º) of the range of motion as is shown in the joint angle matched paired t-test comparisons in **Figure 8**. The quantity of information increases by 4500% (2 peaks vs. 90 degrees comparisons or 4 peaks vs. 180 degrees) with time series normalization multiple comparison method.

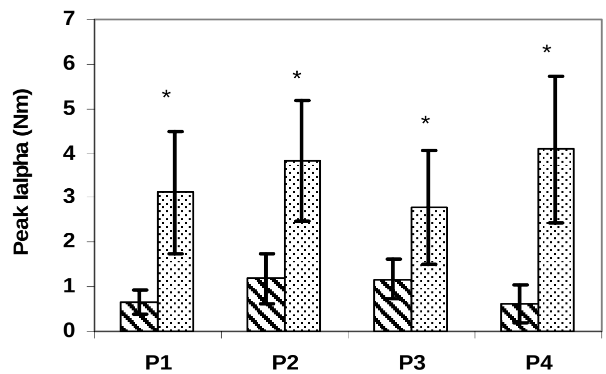


Figure 9: Peak Iα (± SD) in the low acceleration strategy (dark striped bar) and moderate acceleration strategy (dotted bar). Significant differences are indicated by *.

3. RJM

Elastic moment was not significantly different between the moderate acceleration and low acceleration strategies through 150º of 180º. RJM differences between strategies were therefore primarily the result of Iα differences. RJM in the moderate acceleration strategy was significantly different from RJM in the low acceleration strategy through 142° of 180° (79%) of angular excursion of the exercise (**Figure 10**). Regions of non significant RJM differences occurred at the start and end points of the repetition as well as the mid point in the concentric and eccentric phases. In the first period of the concentric phase (from 120° to ~ 60°), RJM in the moderate acceleration strategy was up to 50% higher than joint angle matched RJM in the low acceleration strategy. Greater P1 Iα magnitudes in the moderate acceleration strategy during this period as subjects were yanking on the band accounts for the greater resulting RJM. In the second period of the concentric phase (~60° to 30°), RJM in the moderate acceleration strategy was up to 32% lower than RJM in the low acceleration strategy. Greater P2 Iα magnitudes in the moderate acceleration strategy directed towards shoulder external rotation (positive acceleration) during this period accounts for lower resulting RJM. In the first period of the eccentric phase (30° to ~ 75°), RJM in the moderate acceleration strategy is up to 31% lower than RJM in the low acceleration strategy. Greater P3 Iα magnitudes in the moderate acceleration strategy directed towards shoulder external rotation (positive acceleration) accounts for lower resulting RJM. In the second period of the eccentric phase (~75° to 120°), RJM in the moderate acceleration strategy is up to 111% higher than RJM in the low acceleration strategy. Greater P4 Iα magnitudes in the moderate acceleration strategy directed towards shoulder external rotation (negative acceleration) accounts for higher resulting RJM.

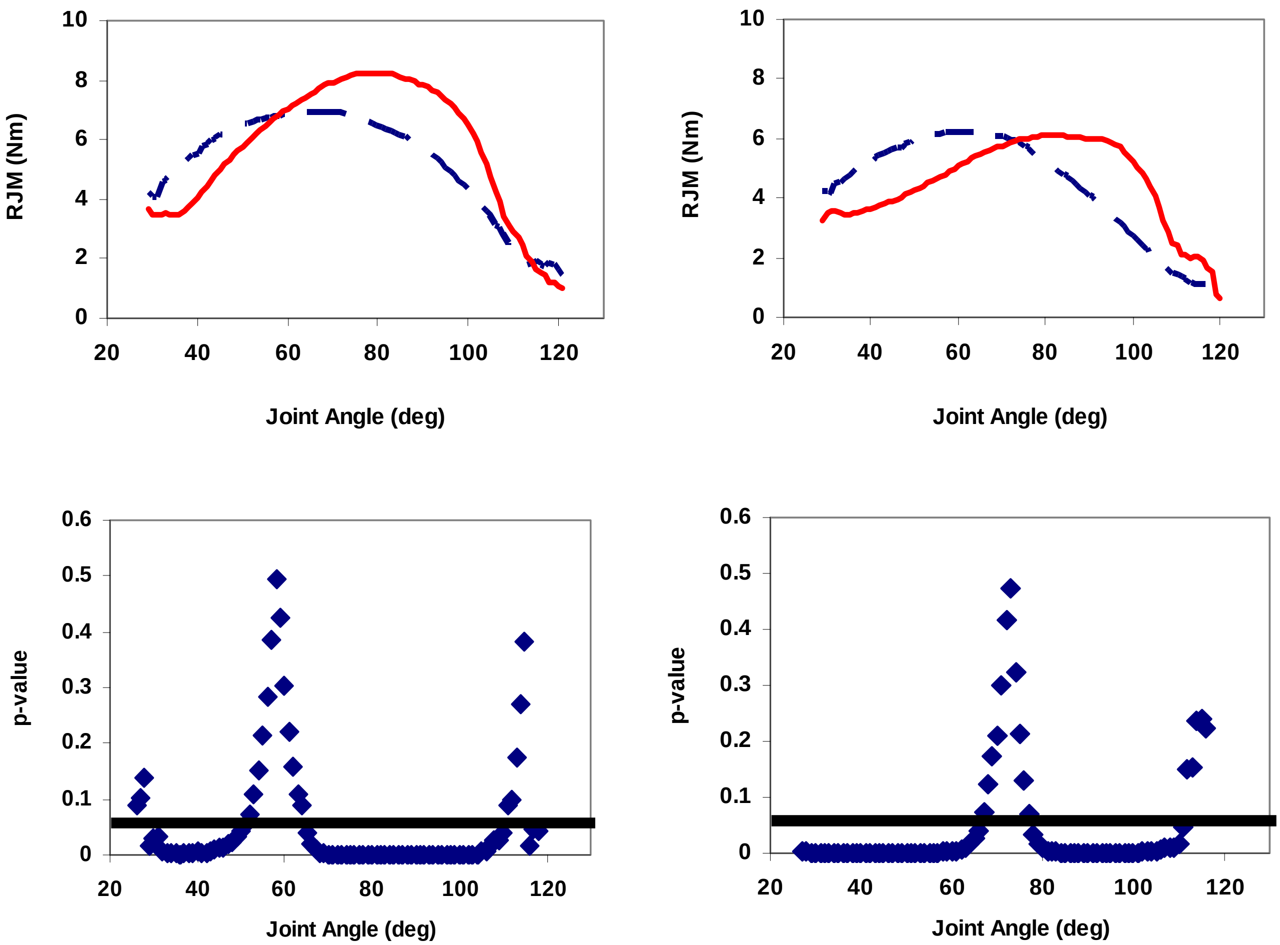


Figure 10: Average shoulder RJM for the low acceleration strategy (dashed dark blue line) and moderate acceleration strategy (solid light red line) during the concentric (left graph) and eccentric (right graph) phases of exercise. Results of joint angle matched t-tests are shown in the corresponding graphs below for the concentric and eccentric phases. Statistically significant RJM differences ($p<0.05$) between the moderate acceleration and low acceleration strategies are shown by scatter points below the wide horizontal line at 0.05 and account for 76% (68º of 90º) of concentric phase comparisons, and 82% (74º of 90º) of eccentric phase comparisons. Regions of non significance occur at the start and end of repetition as well as at joint angles where the RJM curves intersect as Iα transitions from positive to negative or negative to positive.

4. Peak RJM

Between repetition peak P1 and P4 RJM differences were assessed using repeated measures ANOVA. A high degree of inter repetition consistency was found in both conditions with two exceptions. Peak P1 RJM in the fourth repetition of the low acceleration strategy (7.22 ± 1.38 Nm) was significantly lower than the first (7.74 ± 1.23 Nm, $p<0.05$) and third (7.45 ± 1.43 Nm, $p<0.05$) repetitions. Maximum peak P1 RJM difference (0.52 Nm) was not considered sufficient to preclude collapsing repetitions into a single mean. Peak P1 RJM in the sixth repetition of the moderate acceleration strategy (8.26 ± 1.44 Nm) was significantly lower than the fifth repetition (8.81 ± 1.63 Nm, $p<0.05$). This difference (0.55 Nm) was sufficiently small to collapse repetitions into single mean. Paired t-tests were then performed comparing peak P1 and peak P4 RJM between conditions (**Figure 11**). Peak P1 RJM in the moderate acceleration strategy (8.87 ± 1.75 Nm) was significantly greater than peak P1 RJM in the low acceleration strategy (7.39 ± 1.41 Nm, $p<0.01$). Angle of peak P1 RJM was different between conditions (~85° moderate acceleration, ~70° low acceleration). Peak P4 RJM in the moderate acceleration strategy (6.94 ± 1.32 Nm) was not significantly different from peak P4 RJM in the low acceleration strategy (6.55 ± 1.21 Nm). Angle of peak P4 RJM was different between conditions and accounted for the lack of significant peak P4 RJM differences (see **Figure 10** above, eccentric phase). Peak P1 and P4 RJM analysis provides information at two points (joint angles) of the exercise. This analysis alone would fail to show potential between strategies RJM differences occurring at 178° of the range of motion as is shown in the joint angle matched paired t-test comparisons in **Figure 10**. We would conclude from a peak P4 analysis that there was little difference in RJM between low acceleration and moderate acceleration movement strategies.

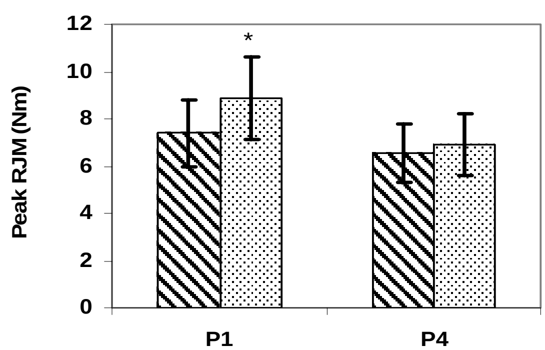


Figure 11: Peak RJM (± SD) in the low acceleration strategy (dark striped bar) and moderate acceleration strategy (dotted bar). Significant differences are indicated by *.

5. Relative RJM differences

Relative between conditions RJM differences were determined by dividing RJM in the moderate acceleration strategy by RJM in the low acceleration strategy. Results represent RJM in the moderate acceleration strategy as a percentage of RJM in the low acceleration strategy (**Figure 12**). In the concentric phase RJM in the moderate acceleration strategy varied from 50% higher to 32% lower than RJM in the low acceleration strategy. In the eccentric phase RJM in the moderate acceleration strategy varied between 31% lower and 111% higher than RJM in the low acceleration strategy.

Greater relative RJM differences occurred in the eccentric phase due to:

- Lower elastic moment magnitudes occurred in the eccentric phase (see Moment of Elastic Curve Shift on pages 57-58). Iα therefore contributes relatively more to RJM in the eccentric phase resulting in greater between conditions RJM differences.
- Lowest elastic moment magnitudes occurred in the second period of the eccentric phase from ~75º to 120º (see **Figure 27**). In this period Iα has the greatest relative contribution to RJM and between condition RJM differences (which are determined almost exclusively be Iα differences) are also the greatest.

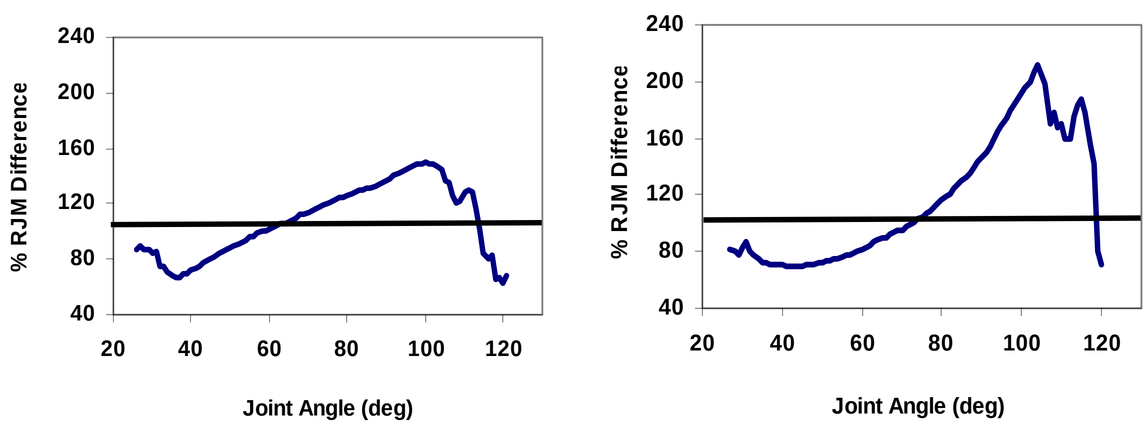


Figure 12: RJM in the moderate acceleration strategy as a percentage of RJM in the low acceleration strategy displayed in the concentric (left graph) and eccentric (right graph) phases of exercise. Relative to low acceleration strategy, the moderate acceleration strategy results in two epochs of greater shoulder loading (120º to ~60º concentric, ~75º to 120º eccentric) and two periods of reduced shoulder loading (~60º to 30º concentric, 30º to ~75º).

6. Mechanical Work

The area under the elastic moment curve (elastic moment integral) in the moderate acceleration and low acceleration movement strategies were determined using the Trapezoid Law. Elastic moment was not significantly different between conditions however very small elastic moment integral differences were found between strategies in the concentric (0.02 Joules) and eccentric (0.12 Joules) phases. Elastic moment integral was equalized between strategies by applying these small offset values and then used to re calculate RJM at all joint angles. Area under the RJM curves in both strategies was then determined using the Trapezoid Law (**Figure 13**). Average mechanical work performed in a single repetition was equivalent in both strategies (16.4 Joules). Greater work was performed in the concentric phase (9.1 Joules) than the eccentric phase (7.3 Joules) as a result of the downwards elastic moment curve shift (see Moment of Elastic Curve Shift on pages 57-58, **Figure 32**). The distribution of work was different between strategies. In the moderate acceleration strategy, 16% more work (5.9 Joules vs. 5.1 Joules) was performed during the first ¼ of the repetition, and 38% more work (3.7 Joules vs. 2.7 Joules) was performed in the last ¼ of the repetition. These periods of higher relative work in the moderate acceleration strategy occurred when forearm acceleration was directed towards shoulder internal rotation (P1 and P4 $I\alpha$). In the middle ½ of the repetition (from ~60º shoulder rotation to full internal rotation and then back to ~75º shoulder rotation), 21% less work (6.9 Joules vs. 8.6 Joules) was performed in the moderate acceleration strategy. During the middle ½ of the repetition forearm acceleration was directed towards shoulder external rotation (P2 and P3 $I\alpha$). Greatest relative work differential between movement strategies occurred in the last ¼ of repetition when elastic moment magnitude was lowest and $I\alpha$ therefore had the greatest contribution to RJM.

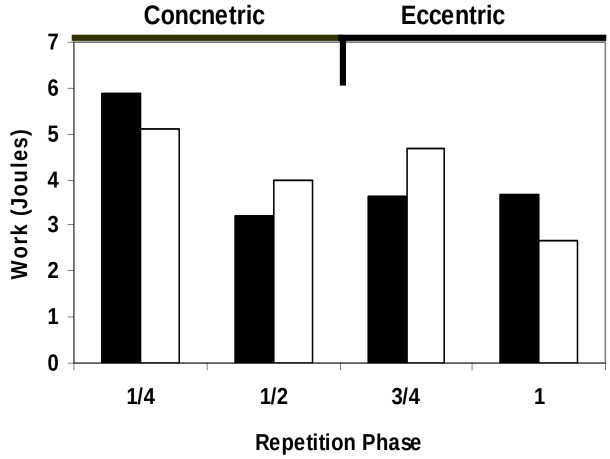


Figure 13: Average work distribution over one repetition in moderate acceleration (dark bars) and low acceleration movement strategies.

2:2 vs. <1:1 Cadence Strategies

1. Iα

The acceleration dependent moment (Iα) in the <1:1 cadence strategy was significantly different from the acceleration dependent moment (Iα) in the 2:2 cadence strategy through 135º of 180º (75%) of angular excursion of the exercise (**Figure 14**). Non significant Iα differences occurred:

i. At the start and end points of the exercise (~116º-120º) where acceleration was at or close to zero.

ii. Close to the mid point of the concentric and eccentric phases of the repetition where acceleration is at or close to zero This occurs as the direction of acceleration changes from negative to positive in the concentric phase (corresponding to a change from shoulder internal rotation directed acceleration to shoulder external rotation directed acceleration), and from positive to negative in the eccentric phase (corresponding to a change from shoulder external rotation directed acceleration to shoulder internal rotation directed acceleration).

In the <1:1 cadence strategy Iα magnitudes were lower in the eccentric phase compared to concentric phase resulting in lower between strategies Iα differences during the eccentric phase.

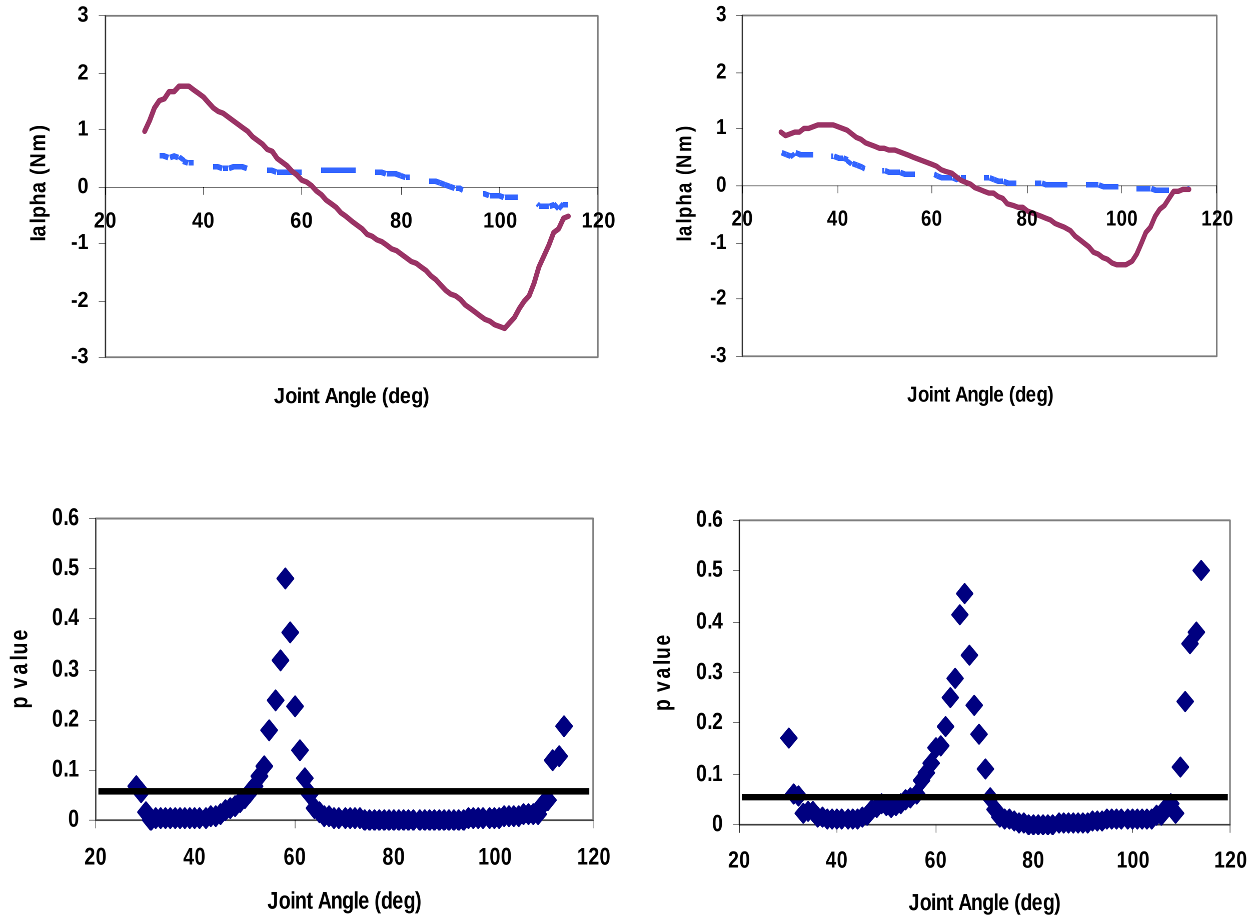


Figure 14: Average acceleration dependent moment ($I\alpha$) for the <1:1 (solid dark wine line) and 2:2 (dashed light blue line) cadence strategies during the concentric (left graph) and eccentric (right graph) phases of exercise. Results of joint angle matched dependent t-tests are shown in the corresponding graphs below for the concentric and eccentric phases. Statistically significant $I\alpha$ differences ($p<0.05$) between conditions are shown by scatter points below the wide horizontal bar at 0.05 and covers 79% (71° of 90°) of comparisons in the concentric phase and 71% (64° of 90°) of comparisons in the eccentric phase. Regions of non significance correspond to epochs of low acceleration at repetition start, end, and mid points.

2. <u>Peak Iα</u>

Between repetition Iα peak differences were assessed for all four acceleration phases using repeated measures ANOVA. A significant repetitions effect occurred in the <1:1 cadence strategy (F=2.80, df =5) where peak P1 Iα in the second repetition (-3.10 ± 2.57 Nm) was greater than peak P1 Iα in repetition three (-2.67 ± 2.54 Nm, P<0.01) and repetition four (-2.58 ± 2.57 Nm, p<0.02). These peak P1 Iα differences were not sufficient to preclude collapsing data in the <1:1cadence strategy. Paired t-tests were then used to compare average Iα peaks between 2:2 and <1:1 cadence strategies. All Iα peaks in the <1:1 cadence strategy were significantly greater than corresponding Iα peaks in the 2:2 cadence strategy (p<0.001) (**Figure 15**). Absolute values of peak P1 and P4 Iα were used in the comparisons.

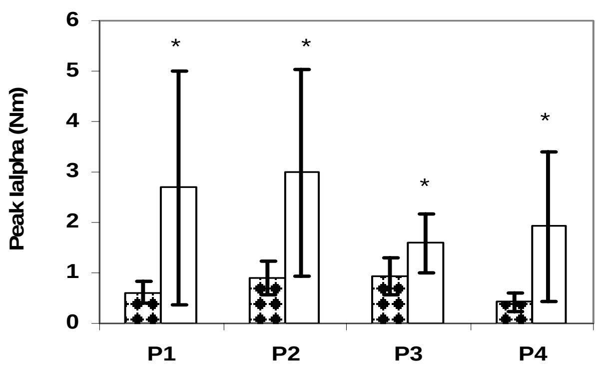


Figure 15: Peak Ialpha (± SD) in the 2:2 (dark triangle bar) and <1:1 (light bar) cadence strategies. Significant differences are indicated by *.

3. <u>RJM</u>

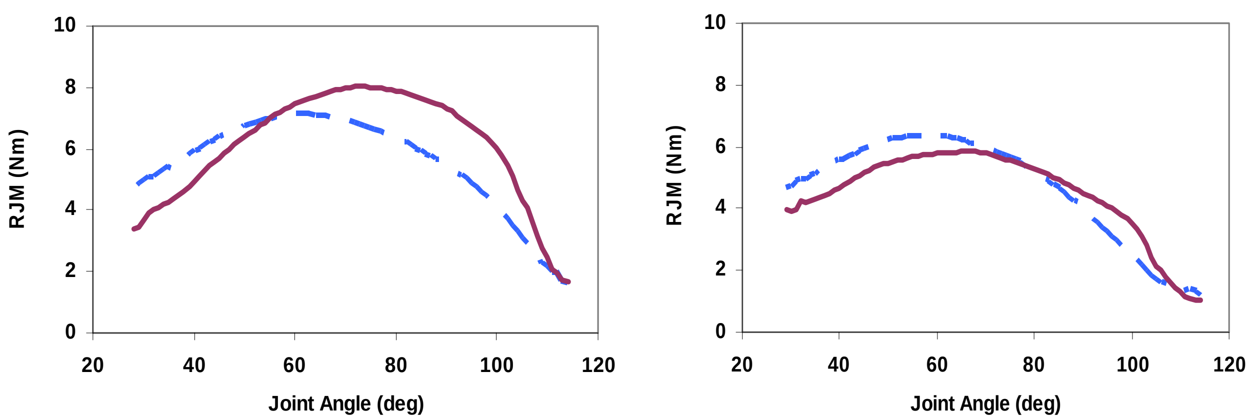
Elastic moment was not significantly different between the 2:2 and <1:1 cadence strategies. RJM differences between strategies were therefore the result of between-strategy Iα differences. RJM in the <1:1 cadence strategy was significantly different from the RJM in 2:2 cadence strategy through 108° of 180° (60%) of angular excursion of the exercise (**Figure 16**). Regions of non significant RJM differences occurred at:

i. At the start and end of the repetition (~110º-120º) where Iα in both strategies was at or close to zero.
ii. At or near the mid point in the concentric and eccentric phases where Iα in the <1:1 strategy transitions from negative to positive (concentric phase) and positive to negative (eccentric phase).

In the first period of the concentric phase (from 118° to ~56°), RJM in the <1:1 cadence strategy was up to 47% higher than joint angle matched RJM in the 2:2 cadence strategy. Greater P1 Iα magnitudes in the <1:1 cadence strategy during this period - as subjects were yanking on the band- accounts for the greater resulting RJM. In the second period of the concentric phase (~56º to 28º), RJM in the <1:1 cadence strategy was up to 30% lower than RJM in the 2:2 cadence strategy. Greater P2 Iα magnitudes in the <1:1 cadence strategy (positive acceleration directed towards shoulder external rotation) during this period accounts for reduced RJM. In the first period of the eccentric phase (28° to ~73°), RJM in the <1:1 cadence strategy was up to 22% lower than RJM in the 2:2 cadence strategy. Greater P3 Iα magnitudes in the <1:1cadence strategy (positive acceleration directed towards shoulder external rotation) accounts for lower resulting RJM. In the second period of the eccentric phase (~73° to 118°), RJM in the <1:1 cadence strategy was up to 41% higher than RJM in the 2:2 cadence strategy. Greater P4 Iα magnitudes (negative acceleration directed towards shoulder external rotation) accounts for higher resulting RJM.

On average the magnitude and volume of RJM differences between cadence strategies was lower in the eccentric phase as a result of:

- Iα magnitudes in the <1:1 cadence strategy were lower in the eccentric phase relative to concentric phase (see **Figure 14**). This resulted in lower relative Iα and subsequent RJM differences between cadence strategies (Iα in the 2:2 strategy was similar in concentric and eccentric phases).



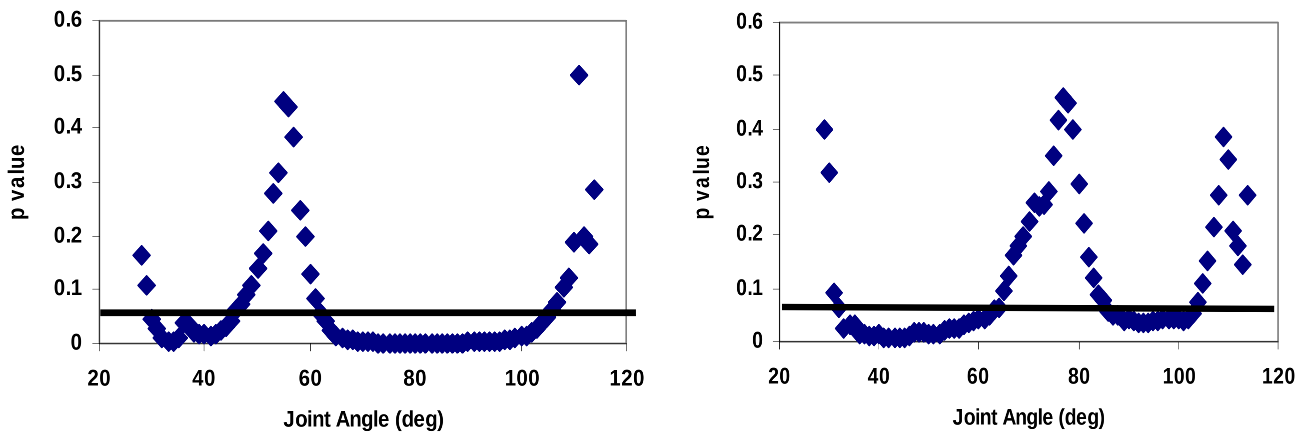


Figure 16: Average shoulder RJM for the <1:1 (solid dark wine line) and 2:2 (dashed light blue line) cadence strategies during the concentric (left graph) and eccentric (right graph) phases of exercise. Results of joint angle matched dependent t-tests are shown in the corresponding graphs below for the concentric and eccentric phases. Statistically significant RJM differences (p<0.05) between conditions are shown by scatter points below the wide horizontal bar at 0.05 and accounts for 68% (61º of 90º) of concentric phase comparisons and 52% (47º of 90º) of eccentric phase comparisons. Regions of non significance correspond to the start and end of repetition as well as where the RJM curves cross when Iα transitions from positive to negative or negative to positive.

4. <u>Peak RJM</u>

Between repetition peak P1 and P4 RJM differences were assessed using repeated measures ANOVA. No significant differences were detected in either cadence strategy. Paired t-tests were then performed comparing peak P1 and P4 RJM between cadence strategies (**Figure 17**). Peak P1 RJM in the <1:1 cadence strategy (8.56 ± 2.21 Nm) was significantly greater than peak P1 RJM in the 2:2 cadence strategy (7.53 ± 1.28 Nm, p<0.01). Peak P4 RJM in the <1:1 cadence strategy (6.41 ± 0.93 Nm) was not significantly different from peak P4 RJM in the 2:2 cadence strategy (6.66 ± 1.19 Nm). Peak RJM analysis alone would indicate no between cadence strategy RJM difference in the eccentric phase which is clearly not the case (see **Figure 16**).

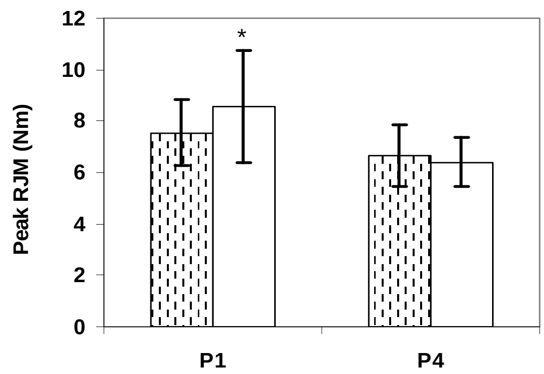


Figure 17: Peak RJM (± SD) in the 2:2 (dotted bar) and <1:1 (light bar) cadence strategies. Significant differences are indicated by *.

5. Relative RJM Differences

Relative between cadence strategy RJM differences were determined by dividing RJM in the <1:1 cadence strategy by RJM in the 2:2 cadence strategy. Results represent RJM in the <1:1 cadence strategy as a percentage of RJM in the 2:2 cadence strategy. In the concentric phase, RJM in the <1:1 cadence strategy varied from 47% higher to 30% lower than RJM in the 2:2 cadence strategy. In the eccentric phase, RJM in the <1:1 cadence strategy varied from 22% lower to 41 % higher than RJM in the 2:2 cadence strategy (**Figure 18**). Greater between-strategy Iα differences in the concentric phase resulted in greater relative between-strategy RJM differences during this period. The period of greatest RJM differences occurring in the eccentric phase (~90º to 115º) corresponds to the range of motion were elastic moment magnitude is lowest and between-strategy Iα differences have the greatest influence on between-strategy RJM differences.

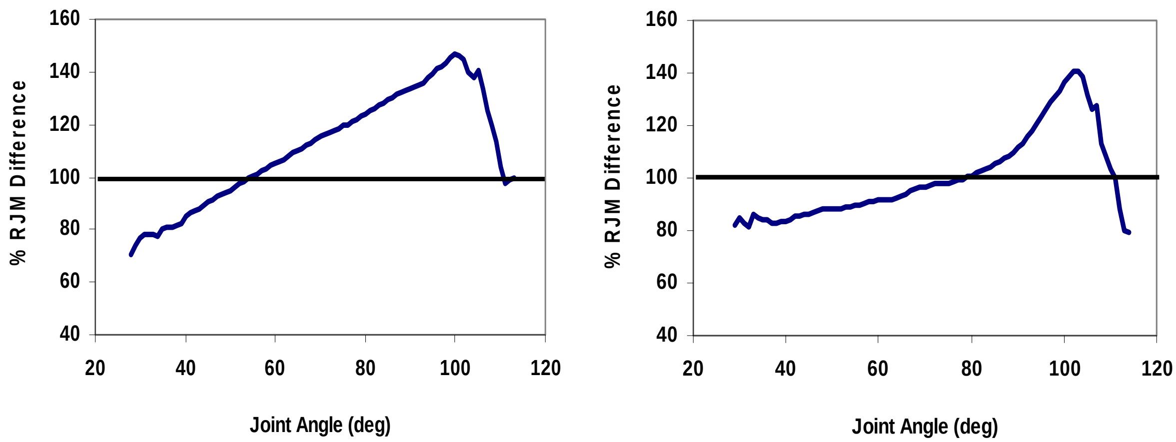


Figure 18: RJM in the <1:1 cadence strategy as a percentage of RJM in the 2:2 cadence strategy displayed in the concentric (left graph) and eccentric (right graph) phases of exercise. Relative to 2:2 cadence strategy, the <1:1 cadence strategy results in two epochs of greater shoulder loading (118º to ~60º concentric, ~80º to 118º eccentric) and two periods of reduced shoulder loading (~60º to 28º concentric, 28º to ~75º).

7. Mechanical Work

The area under the moment of elastic curve (elastic moment integral) in the 2:2 and <1:1 cadence strategies were determined using the Trapezoid Law. Elastic moment was not significantly different between conditions however very small elastic moment integral differences were found between conditions in the concentric (0.16 Joules) and eccentric (0.33 Joules) phases. Elastic moment integral was equalized between conditions by applying these small offset values and then used to re calculate RJM at all joint angles. Area under the RJM curves in both conditions was then determined using the Trapezoid Law (**Figure 19**). Total average mechanical work performed in a single repetition was equivalent in both conditions (16.2 Joules). Greater work was performed in the concentric phase (8.9 Joules) than the eccentric phase (7.3 Joules) as a result of elastic moment curve shift (see Moment of Elastic Curve Shift on pages 57-58). The distribution of work was different between conditions. In the <1:1 cadence strategy 14% more work (5.6 Joules vs. 4.9 Joules) was performed during the first ¼ of the repetition and 14% more work (3.0 Joules vs. 2.7 Joules) was performed in the last ¼ of the repetition. These periods of higher relative work in the <1:1 cadence strategy occurred when forearm acceleration was directed towards shoulder internal rotation (P1 and P4 Iα). In the middle ½ of the repetition (from ~60º shoulder rotation to full internal rotation and then back to ~73º shoulder rotation) 12% less work (7.6 Joules vs. 8.7 Joules) was performed in the <1:1 cadence strategy. During the middle ½ of the repetition forearm acceleration was directed towards shoulder external rotation (P2 and P3 Iα).

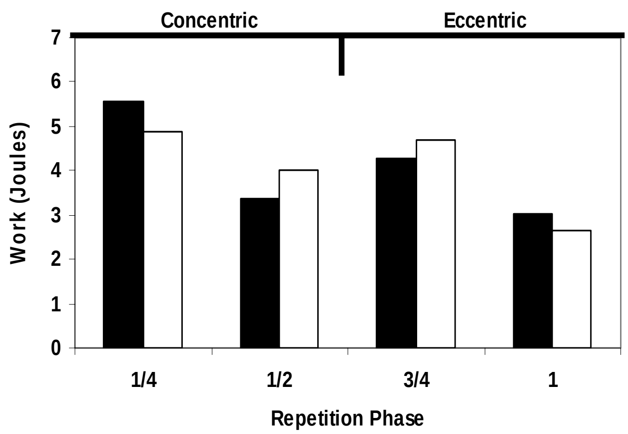


Figure 19: Average work distribution over one repetition in <1:1 (dark bars) and 2:2 cadence strategies.

<u>Moderate Acceleration Movement Strategy vs. <1:1 Cadence Strategy</u>

1. <u>Iα</u>

The pattern and magnitude of Iα in the moderate acceleration and <1:1 cadence strategies were similar and did not differ significantly at any joint angles in the concentric phase of the exercise. In the eccentric phase, Iα magnitude in the moderate acceleration strategy was significantly greater than Iα in the <1:1 cadence strategy at 41° of 90° (46%) of range of motion (**Figure 20**). Relatively lower Iα in the <1:1 cadence strategy during the eccentric phase was not unexpected based on lower average velocity during this phase as a result of cueing instructions. Non significant Iα differences in the eccentric phase occurred:

i. At the start and end of the eccentric phase (from 30º to 35ºand ~112º to 119º)
ii. At or close to the eccentric phase mid point (~70º-75° shoulder rotation) where Iα in both strategies were at or close to zero as the direction of acceleration transitioned from positive to negative (corresponding to a change from shoulder external rotation directed acceleration to shoulder internal rotation directed acceleration).

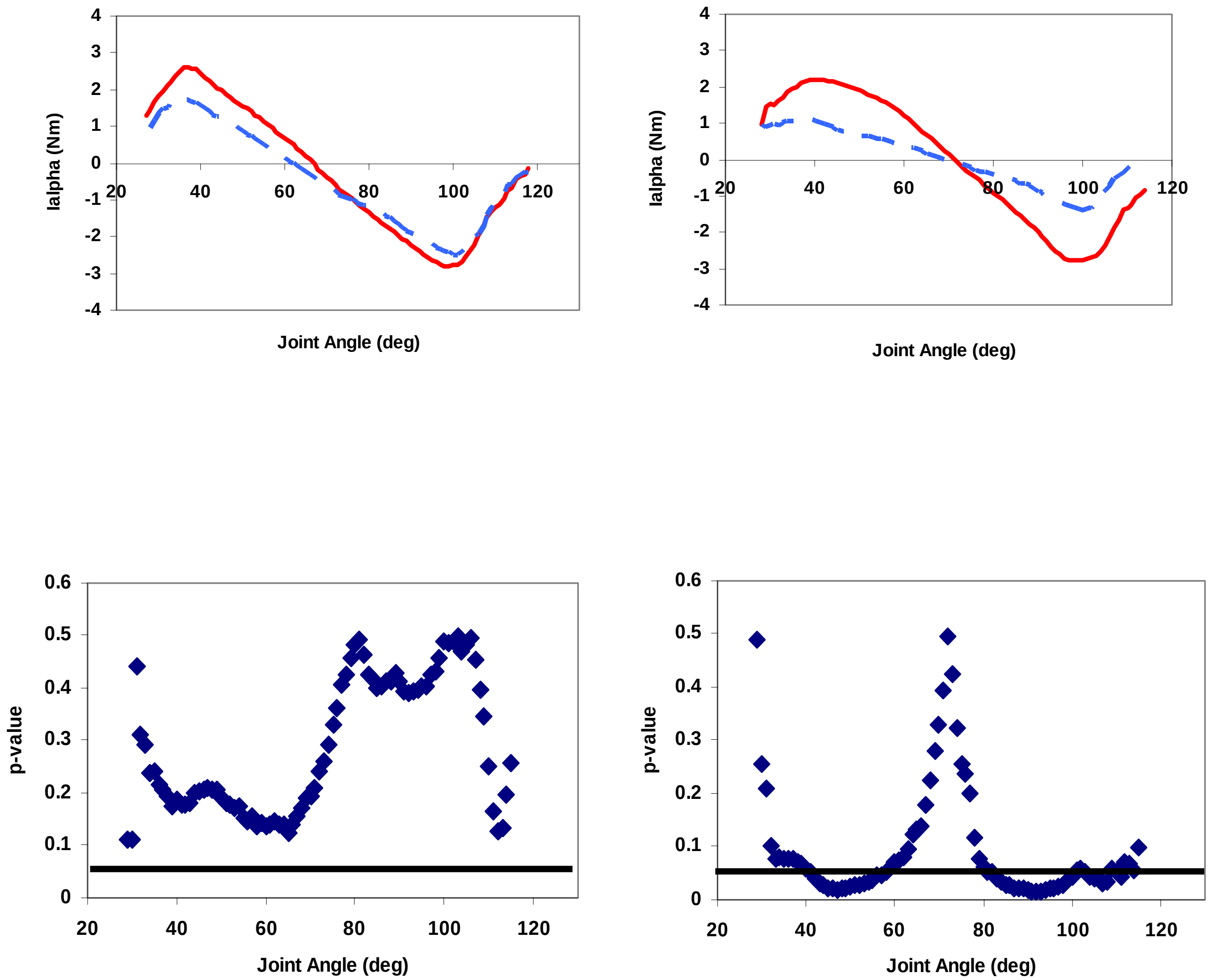


Figure 20: Average acceleration dependent moment (Iα) in the moderate acceleration strategy (solid light red line) and <1:1 cadence strategy (dashed light blue line) during the concentric (top left graph) and eccentric (top right graph) phases of exercise. Results of the joint angle matched dependent t-tests are shown in the corresponding graphs below for the concentric and eccentric phases. Statistically significant Iα differences ($p<0.05$) between conditions are shown by scatter points below the wide horizontal line at 0.05 and covers 46% (41° of 90°) of the comparisons in the eccentric phase. No significant Iα differences occurred in the concentric phase.

2. <u>Peak Iα</u>

Paired t-test were used to compare average Iα peaks between moderate acceleration and <1:1 cadence strategies. No significant Iα differences were found at peak P1 and peak P2 (**Figure 21**). The strategy to "yank" on the elastic to initiate the exercise (peak P1 Iα) and subsequent slowing down after “yank” (peak P2 Iα) were similar in both strategies. In the eccentric phase, peak P3 and peak P4 Iα in the moderate acceleration strategy were significantly greater than peak P3 and peak P4 Iα in the <1:1 cadence strategy (peak P3 difference = 1.3 Nm, $p<0.006$, peak P4 difference = 2.2 Nm, $p<0.001$). This finding was not surprising due to the lower average velocity in the eccentric phase of the <1:1 cadence strategy based on cueing instructions. Absolute values of peak P1 and peak P4 Iα were used in the comparisons.

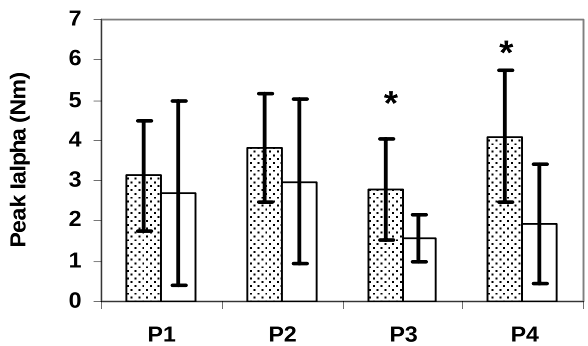


Figure 21: Peak Iα (± SD) in the moderate acceleration strategy (dotted bar) and <1:1 cadence strategy (light bar). Significant differences are indicated by *.

3. <u>Peak P1 Iα vs. Peak P3 Iα</u>

The average acceleration employed to initiate the concentric (peak P1 Iα) and eccentric (peak P3 Iα) phases of the exercise in the moderate acceleration strategy and the <1:1 cadence strategy were compared using paired t-tests. In the moderate acceleration strategy, average peak P1 acceleration (3.12 ± 1.37 Nm) was 11% higher than average peak P3 acceleration (2.78 ± 1.26 Nm, $p<0.001$). In the <1:1 cadence strategy, average peak P1 acceleration (2.68 ± 2.30 Nm) was 41% higher than average peak P3 acceleration (1.59 ± 0.58 Nm, $p<0.001$). Greater peak P1 acceleration in the <1:1 cadence strategy was anticipated based on the cueing instruction (which resulted in significantly higher average velocity during the concentric phase compared to the eccentric phase). The 11% peak P3 acceleration reduction in the moderate acceleration strategy was not anticipated and may have been the result of an attempt by subjects to increase the motion control of their arm during the eccentric phase as a protective mechanism to minimize the potential of the forearm “snapping back” as it was moving in the same direction as the line of pull of the elastic band, thus reducing the potential for injury or trauma at the shoulder. In the moderate acceleration strategy, subjects adopted a movement strategy to reduce peak arm acceleration in the eccentric phase even when instructed to perform eccentric phase in the same manner as the concentric phases (i.e. “tick-toc”). On average subjects tended to yank harder on the elastic band (peak P1) than they did to release it back (peak P3) in both strategies.

In both conditions peak P1 and peak P3 Iα were highly variable indicating a range of movement strategies employed to achieve the instructed condition. In particular the “yank” in the <1:1 cadence strategy was highly variable between subjects as indicated in **Figure 22** below.

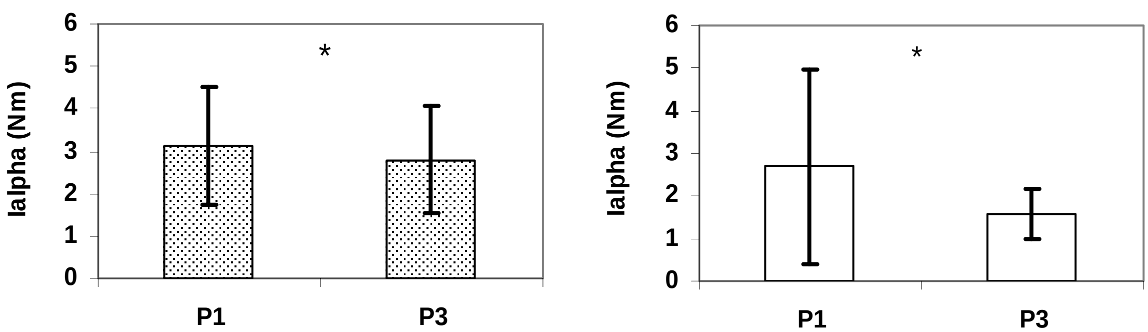


Figure 22 Average (±SD) peak P1 and peak P3 Iα in the moderate acceleration strategy (left graph) and <1:1 cadence strategy (right graph). Significant differences are indicated by *.

4. <u>RJM</u>

Elastic moment was not significantly different between the moderate acceleration strategy and <1:1 cadence strategy through the range of motion of the exercise. Between conditions RJM differences were therefore the result of Iα differences. No significant RJM differences were found in the concentric phase (**Figure 23**). Absence of significant between conditions Iα differences in the concentric phase accounted for this finding.

In the eccentric phase, RJM magnitude in the moderate acceleration strategy was significantly different from RJM in the <1:1 cadence strategy through 45° of 90° (39%) of range of motion. Non significant differences occurred at the start, mid point and end of the eccentric phase when Iα magnitudes in both strategies was at or close to zero and therefore Iα in both strategies was equivalent or close to equivalent. In the first period of the eccentric phase (from 28° to ~70°), RJM in the moderate acceleration strategy was up to 25% lower than joint angle matched RJM in the <1:1 cadence strategy. Greater P3 Iα magnitudes (positive acceleration) in the moderate acceleration strategy accounted for resulting reduced RJM. In the second period of the eccentric phase (~70° to 118°), RJM in the moderate acceleration strategy was up to 95%

higher than RJM in the <1:1 cadence strategy. Greater P4 Iα magnitudes (negative acceleration) in the moderate acceleration strategy accounted for higher resulting RJM.

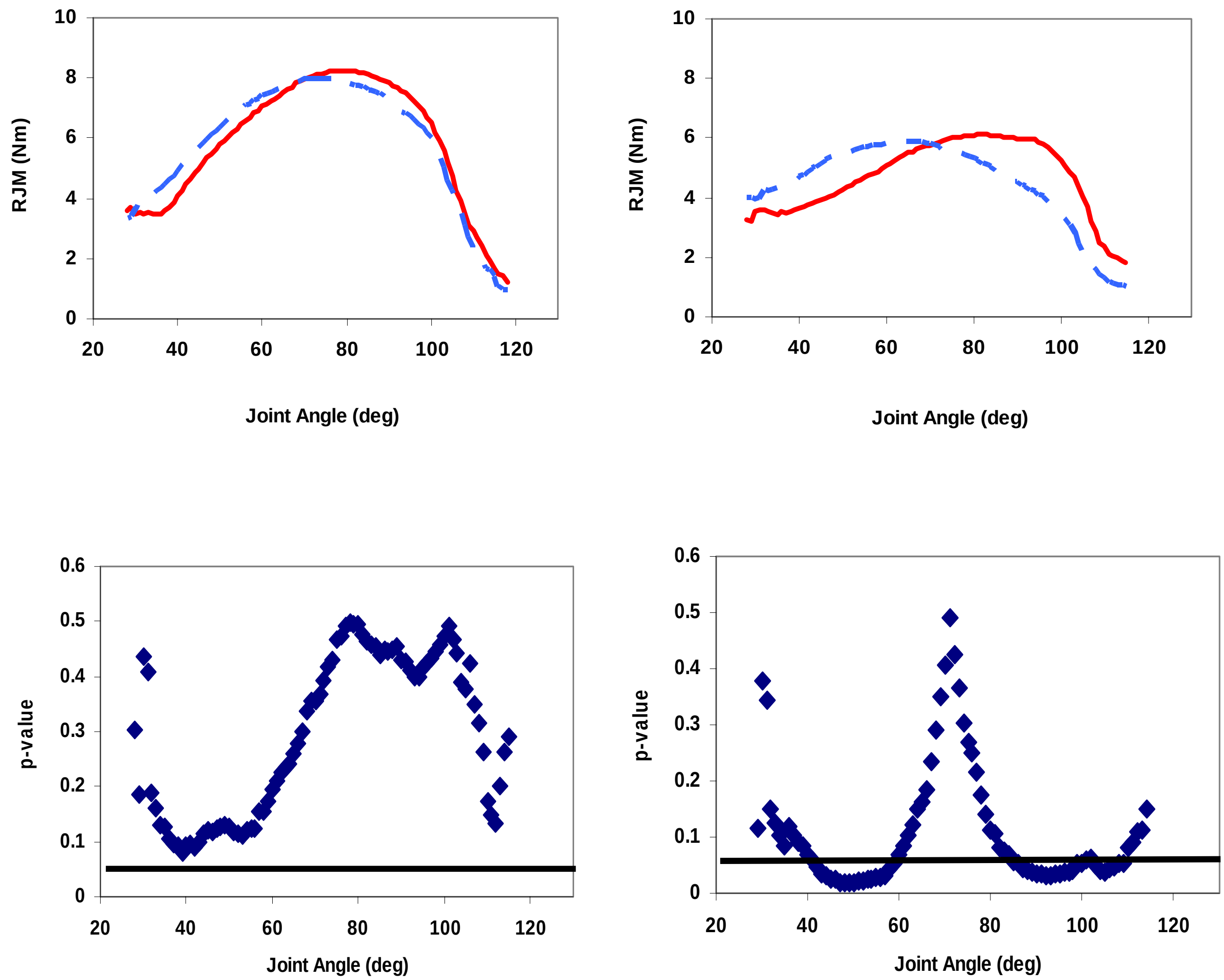


Figure 23: Average shoulder RJM in the moderate acceleration strategy (solid light red line) and <1:1 cadence strategy (dashed light blue line) in the concentric (top left graph) and eccentric (top right graph) phases of exercise. Results of joint angle matched dependent t-tests are shown in the corresponding graphs below for the concentric and eccentric phases. Statistically significant Iα differences ($p<0.05$) between strategies are shown by scatter points below the wide horizontal line at 0.05 and accounts for 39% (35º of 90º) of eccentric phase comparisons. No significant differences were found in the concentric phase.

5. Peak RJM

Paired t-tests were performed comparing peak P1 and peak P4 RJM between strategies. Peak P1 RJM in the moderate acceleration strategy (8.87 ± 1.75 Nm) was not significantly different from peak P1 RJM in the <1:1 cadence strategy (8.56 ± 2.20 Nm) (**Figure 24**). Peak P4 RJM in the moderate acceleration strategy (6.94 ± 1.32 Nm) was significantly greater than peak P4 RJM in the <1:1 cadence strategy (6.40 ± 0.93 Nm, $p<0.01$). Angle of peak P4 RJM was different between strategies (see **Figure 23**).

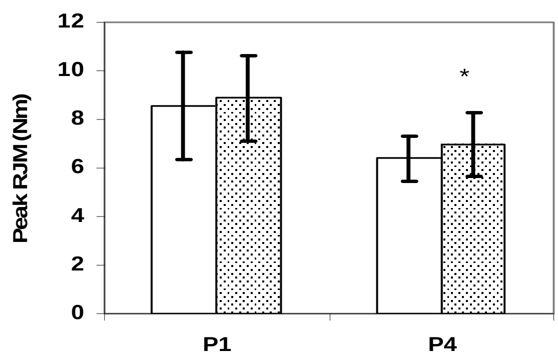


Figure 24: Average (± SD) peak RJM in the <1:1 cadence strategy (light bar) and moderate acceleration strategy (dotted bar). Significant differences are indicated by *.

6. Relative Differences

Relative between conditions RJM differences were determined by dividing RJM in the moderate acceleration strategy by RJM in the <1:1 cadence strategy. Results represents RJM in the moderate acceleration strategy as a percentage of RJM in the <1:1 cadence strategy (**Figure 25**). In the concentric phase, RJM in the moderate acceleration strategy varied from 44% higher to 20% lower than RJM in the <1:1 cadence strategy. In the eccentric phase, RJM in the moderate acceleration strategy varied from 25% lower to 95% higher than RJM in the <1:1 cadence strategy. Greatest between-strategy Iα differences occurred in the eccentric phase and resulted in greatest relative RJM differences. The highest relative RJM differences occurred in the last ¼ of the repetition (from ~90° to 120°) where elastic moment magnitudes are the lowest and between condition Iα differences have the greatest effect on between condition RJM differences.

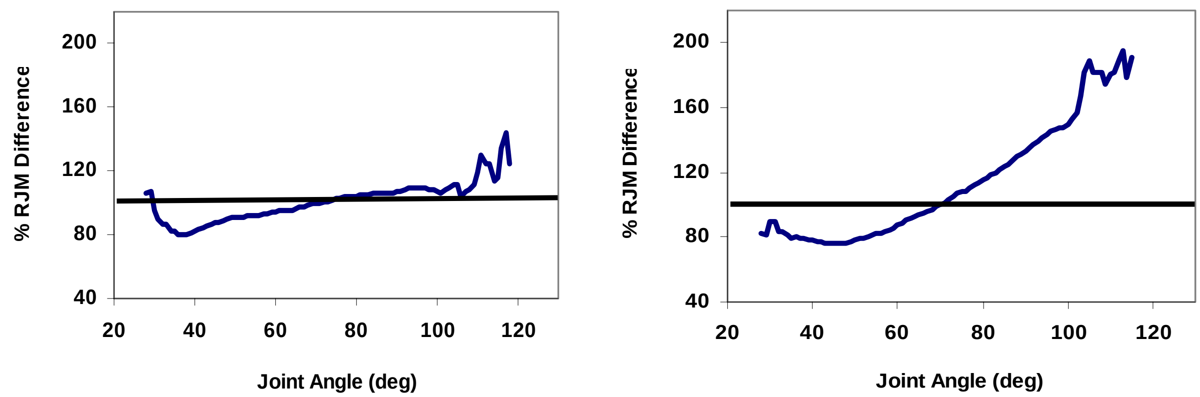


Figure 25: RJM in the moderate acceleration strategy as a percentage of RJM in the <1:1 cadence strategy displayed in the concentric (left graph) and eccentric (right graph) phases of exercise. Greater between-strategy Iα differences in the eccentric phase resulted in greater between-strategy RJM differences.

Low Acceleration Movement Strategy: 0 % vs. 30% Starting Elastic Strain

1. Iα

Iα did not differ significantly between 0% and 30% starting elastic strain throughout the angular excursion of the exercise (**Figure 26**).

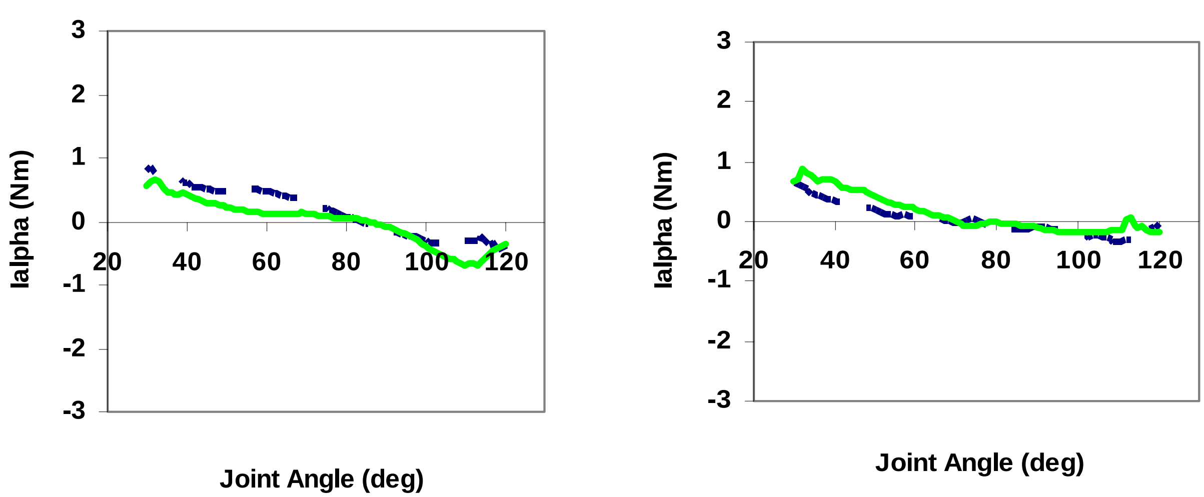


Figure 26: Iα in the low acceleration strategy with 0% (dashed dark blue line) and 30% start stain (solid light green line) in the concentric (left graph) and eccentric (right graph) exercise phases. Iα was not significantly different between conditions.

2. <u>Elastic Moment</u>

Elastic moment in the low acceleration strategy was compared with starting elastic strain at O% and 30% though 180° of angular excursion. Paired t-tests were performed at all joint angles to determine elastic moment differences. 30% elastic start strain resulted in significantly greater elastic moment magnitudes at all joint angles compared to 0% start strain ($p<0.001$). The pattern of the elastic moment remained unchanged between conditions **(Figure 27).** In the eccentric phase of both conditions elastic moment curve shifted downwards. This was consistent in all other movement strategies and was the result of lower elastic recoil force in the eccentric phase.

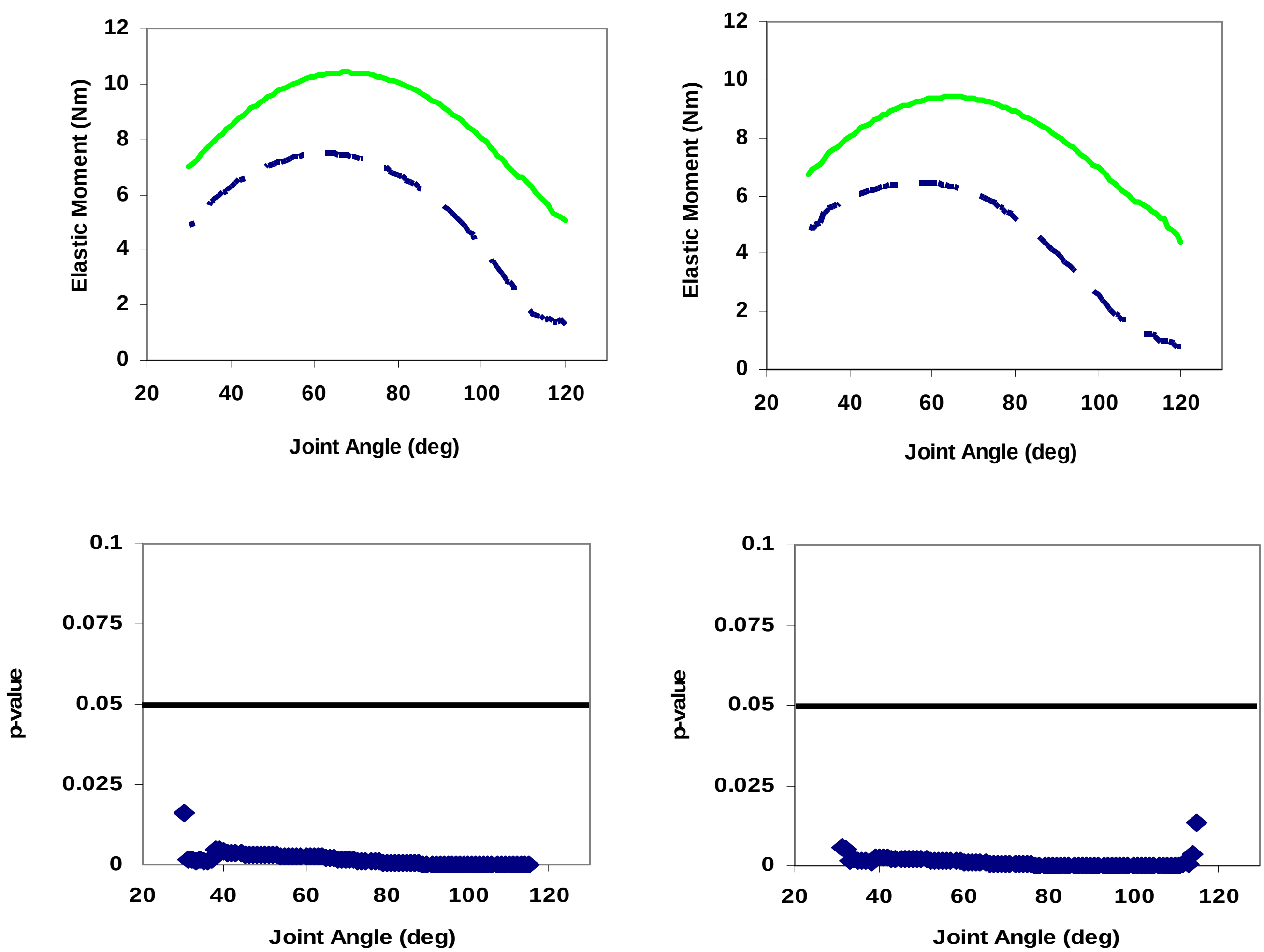


Figure 27: Average elastic moment in the low acceleration strategy with 0% (dashed dark blue line) and 30% start stain (solid light green line) in the concentric (left graph) and eccentric (right graph) exercise phases. Results of joint angle matched dependent t-tests are shown in the corresponding graphs below for the concentric (left) and eccentric (right) phases. Elastic moment was significantly greater in the 30% elastic strain strategy at all 180° tested as indicated by scatter points below p-value of 0.05.

3. RJM

Paired t-tests were performed comparing RJM in the two elastic strain strategies at all joint angles. RJM in the 30% condition was significantly greater than RJM in the 0% condition at all joint angles tested ($p<0.001$). Results corresponded to elastic moment comparisons between conditions described in the section above.

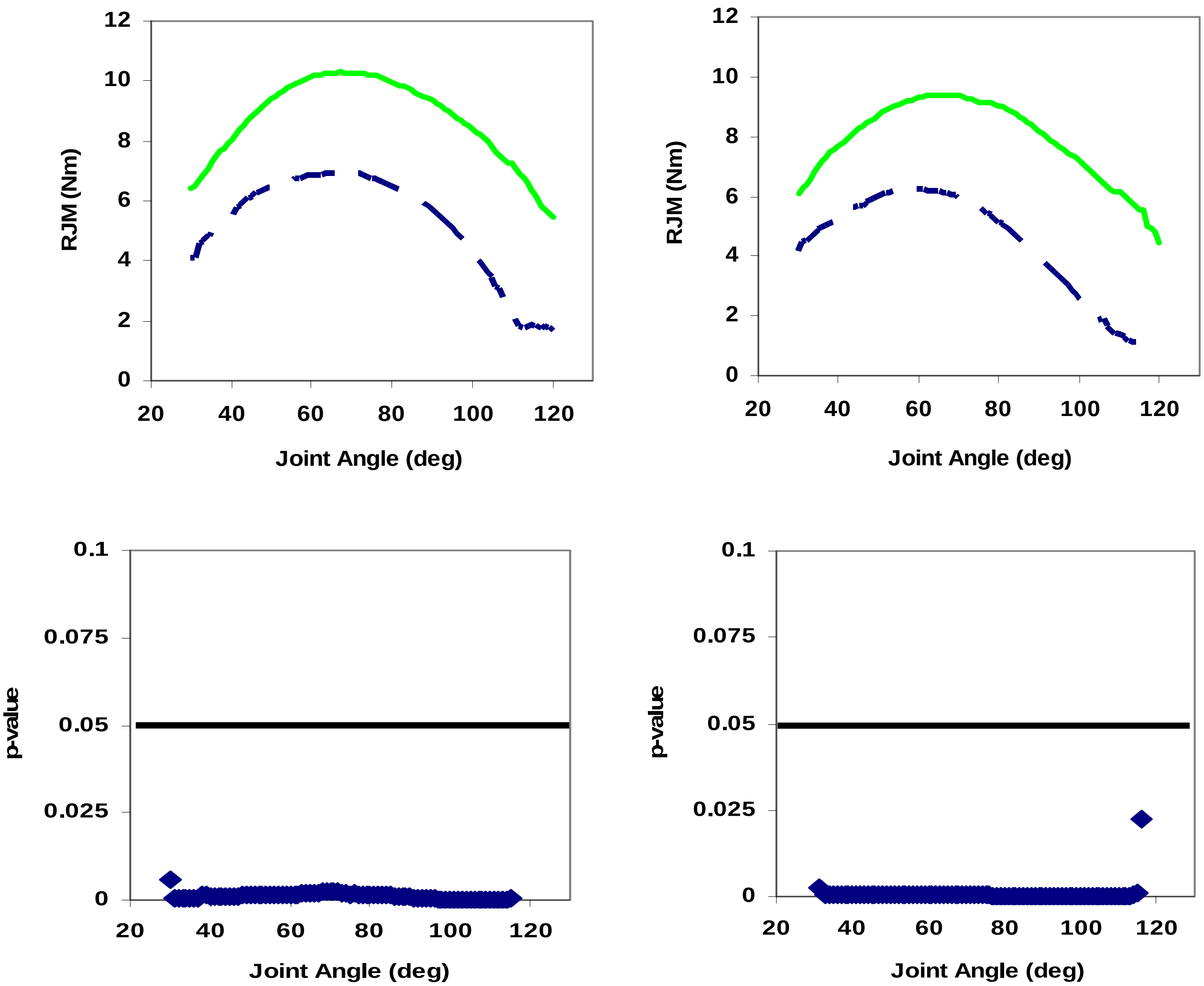


Figure 28: RJM in the low acceleration strategy with 0% (dashed dark blue line) and 30% start stain (solid light green line) in the concentric (left graph) and eccentric (right graph) exercise phases. Results of joint angle matched dependent t-tests are shown in the corresponding graphs below for the concentric (left) and eccentric (right) phases. RJM was significantly greater in the 30% elastic strain strategy at all 180º tested as indicated by scatter points below p-value of 0.05.

4. Relative Differences

Relative between-strategy RJM differences were determined by dividing RJM in the 30% elastic strain strategy by RJM in the 0% elastic strain strategy. Results represent RJM in the 30% elastic strain strategy as a percentage of RJM in the 0% elastic strain strategy **(Figure 29).** RJM in the 30% elastic strain strategy was greater than RJM in the 0% elastic strain strategy by up to 272% in the concentric phase and 430% in the eccentric phase. For both the concentric and eccentric phases, the greatest relative RJM differences occurred between 120° and 85° range of motion.

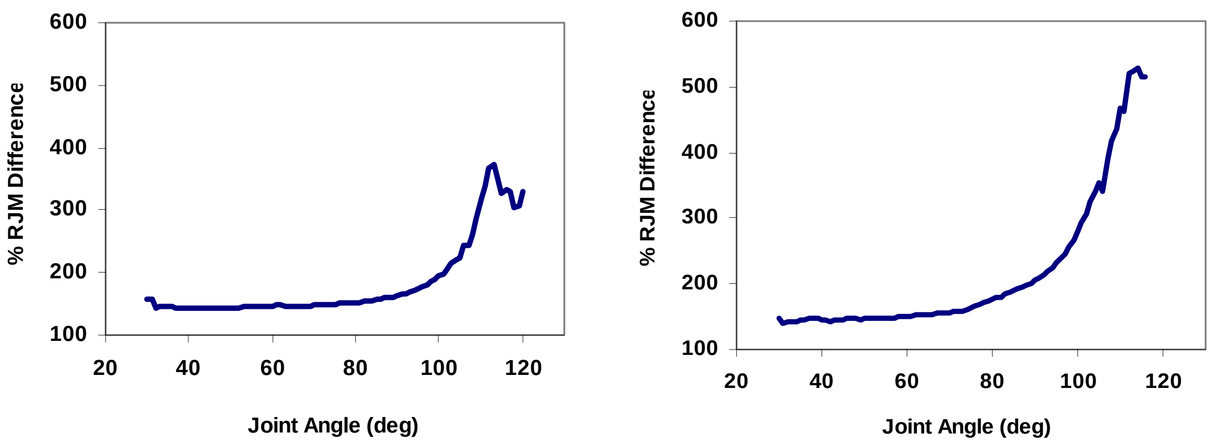


Figure 29: RJM in the 30% strategy as a percentage of RJM in the 0% strategy displayed in the concentric (left graph) and eccentric (right graph) exercise phases. Greatest RJM differences occurred between 120° and ~85° range of motion.

Elastic Moment: Effects of Start Strain vs. Elastic Color Progression

The moment of elastic produced by the other five commercially available elastic bands (yellow, red, green, black, and silver) during shoulder the rotation exercise was determined from published stress/strain regression equations and elastic moment arm values calculated from this study. Elastic moment curves were compared to elastic moment curves produced from the blue elastic band in the low acceleration movement control strategy with 0% and 30% elastic start strain. Elastic moment curves in the concentric phase of shoulder rotation exercise are displayed in **Figure 30**. Apparent in the graph below is that elastic band blue with 30% start strain produces a moment curve that is intermediate to the silver and black elastic moment curves with 0% start strain. 30% start strain could be used as a form of elastic load progression and would provide an additional elastic load gradation as shown below.

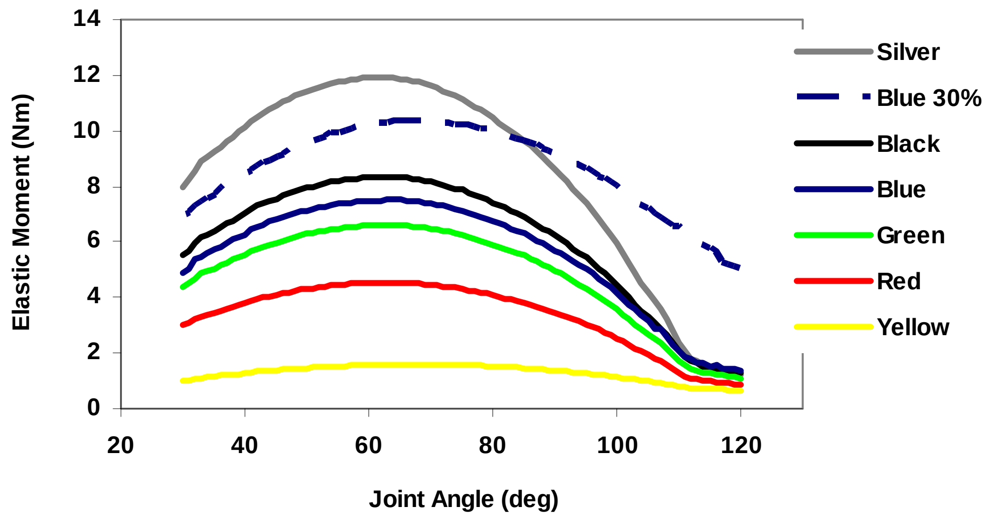


Figure 30: Elastic moment in the concentric phase of shoulder internal rotation exercise for six colors of elastic bands with 0% start strain (solid lines) and elastic moment of the blue band with 30% start strain (dashed blue line).

Moderate Acceleration Movement Strategy: 0 % vs. 30% Starting Elastic Strain

Iα and RJM differences between the low acceleration and moderate acceleration strategies with elastic start strain at 30% were similar to Iα and RJM differences between the low acceleration and moderate acceleration strategies with elastic start strain at 0%. The movement strategy employed did not change significantly as a result of increased load from greater elastic moment at 30% strain. In the moderate acceleration strategy, the overall pattern of Iα was similar at both starting elastic strains **(Figure 30)**, however there was a small Iα difference at the initiation of the exercise (from 120º to ~100º) where P1 Iα magnitude was greater with 30% elastic strain. Overall, moderate acceleration strategy did not become higher acceleration strategy under higher elastic loading conditions. In the low acceleration strategy, the pattern and magnitude of Iα was similar at both starting strains (**See figure 26 above**). Low acceleration strategy did not become moderate acceleration strategy under higher elastic load conditions.

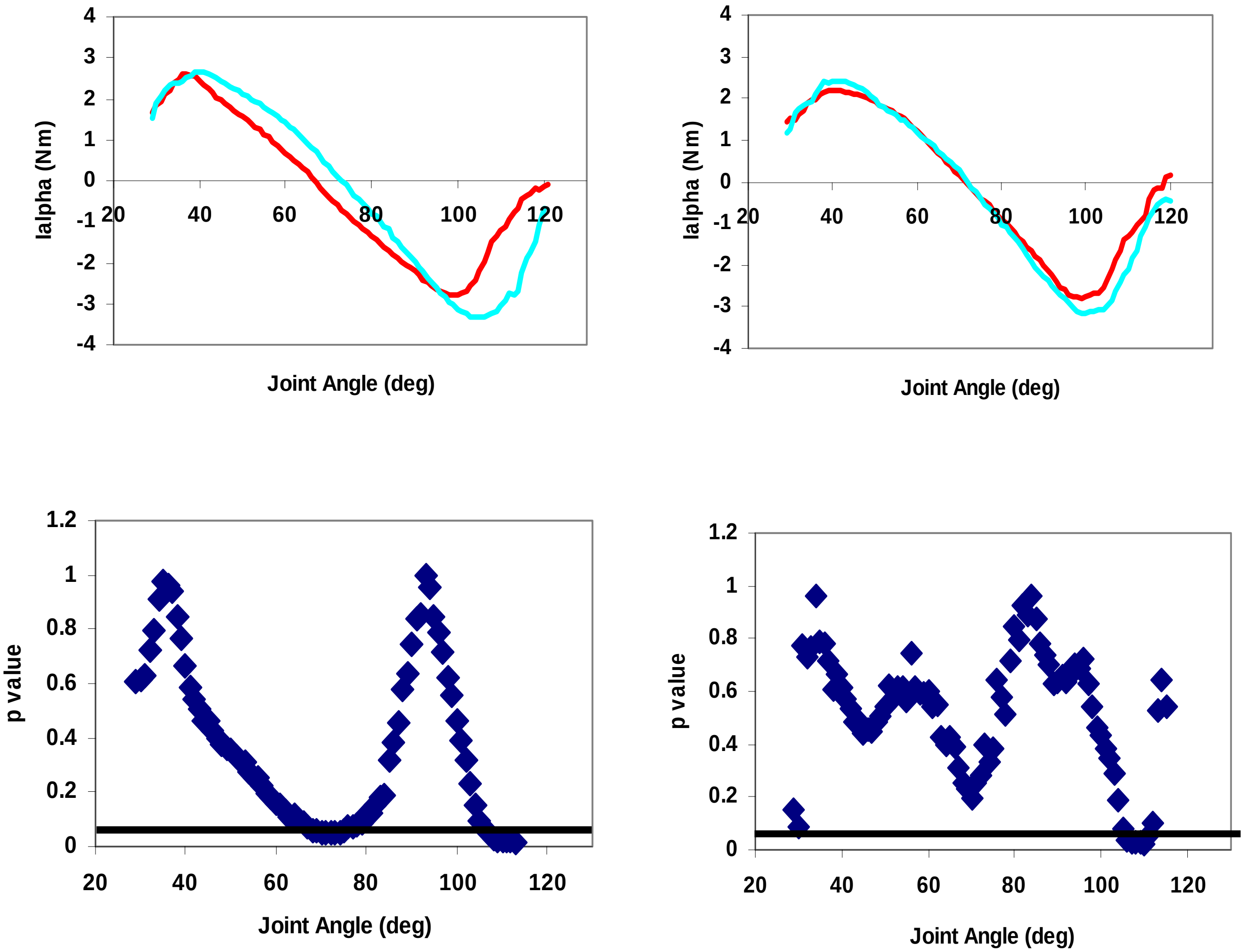


Figure 31: Iα in the moderate acceleration strategy with 0% elastic start strain (dark red line) and 30% elastic start strain (light blue line). Results from concentric (left graph) and eccentric (right graph) are shown. Results of joint angled matched paired t-test are displayed below for concentric (left graph) and eccentric (right graph) phases. Significant Iα differences are indicated by scatter points below the black horizontal line at 0.05

Acceleration vs. Velocity

Stepwise linear regression was performed to determine the relative contribution of elastic moment, peak acceleration ($I\alpha$), and average velocity to peak P1 and peak P4 RJM in all strategies. In the low acceleration and 2:2 cadence strategies elastic moment was the most powerful predictor of peak P1 and P4 RJM (low acceleration: r^2= 0.947 P1, r^2= 0.940 P4. 2:2: r^2= 0.961 P1, r^2= 0.959 P4). In the low acceleration and 2:2 cadence strategies peak RJM was almost exclusively determined by the elastic moment. In these conditions acceleration ($I\alpha$) was a relatively low and contributed very little to peak RJM.

In the moderate acceleration strategy a strong predictive relationship between peak $I\alpha$ and peak RJM was revealed at P1 (r^2=0.597) and P4 (r^2= 0.556). Average velocity was a non significant contributor to the variance observed in peak P1 and peak P4 RJM. $I\alpha$ was also the strongest predictor of RJM in the <1:1 cadence strategy in both concentric (r^2= 0.818 P1 RJM) and eccentric (r^2= 0.308 P4 RJM) phases.

Shoulder Rotation Range of Motion

Joint angles corresponding to the start, mid, and end points of all repetitions were determined from goniometer/time data in spreadsheet form. Average shoulder rotation range of motion in the concentric and eccentric phases and total angular excursion (concentric + eccentric range of motion) were established. In all movement strategies between repetitions range of motion differences was assessed using repeated measures ANOVA. Between repetitions range of motion was highly consistent with one exception in the 2:2 cadence strategy where the concentric phase range of motion in the first repetition (81.2 ± 5.6º) was significantly greater than the fourth (76 ± 5.2º, 4.9ºdifference, $p<0.002$), fifth (76.6 ± 5.7º, 4.5º difference, $p<0.009$), and sixth (76.9 ± 6.0º, 4.3º difference, $p<0.013$) repetitions. Between repetitions differences were sufficiently small to collapse repetitions into a single mean.

Paired t-tests were then used to compare concentric and eccentric phase range of motion within strategies. Small but significant range of motion differences were found in the low acceleration (2º difference, $p<0.001$,) and 2:2 (1º difference, $p<0.001$) strategies.

Between-strategy range of motion differences were assessed using repeated measures ANOVA. Significant differences occurred in the concentric (F=16.055, $p<0.001$, df 3, 59) and eccentric (F=21.057, $p<0.001$, df 3, 59) phases. Post Hoc Tukey's analysis revealed concentric phase range of motion in the moderate acceleration (MA) strategy (82 ± 5.0º) and <1:1 cadence strategy (81 ± 6.6º) were greater than low acceleration (LA) strategy (79 ± 5.4º) and 2:2 cadence strategy (78 ± 5.8º) range of motion ($p<0.001$). Moderate acceleration and <1:1 strategies range of motion and low acceleration and 2:2 strategies range of motion were not significantly different from each other. In the eccentric phase, range of motion in the moderate acceleration strategy (82 ± 5.6º) and <1:1 cadence strategy (80 ± 6.8º) were significantly greater than the low acceleration strategy (77 ± 5.3º) and 2:2 cadence strategy (76 ± 5.7º) range of motion ($p<0.001$). Range of motion in the moderate acceleration strategy and <1:1 cadence strategy, and low acceleration strategy and 2:2 cadence strategy were not significantly different from each other. Maximum between conditions range of motion differences (4º concentric phase, 6º eccentric phase) were sufficiently small not to influence/skew repetition velocity calculation.

Repeated measures ANOVA revealed significant between conditions total angular excursion differences ($F=20.671$, $p<0.001$, df3, 59). Post hoc Tukey's revealed angular excursion in the moderate acceleration strategy (164 ± 10.2º) and <1:1 cadence strategy (161 ± 13.0º) were greater than angular excursion in the low acceleration strategy (156 ± 10.4º) and 2:2 cadence strategy (155 ± 11.4º) ($p<0.001$). Angular excursion in the moderate acceleration strategy and <1:1 cadence strategy, and angular excursion, in the low acceleration strategy and 2:2 cadence strategy were not significantly different from each other.

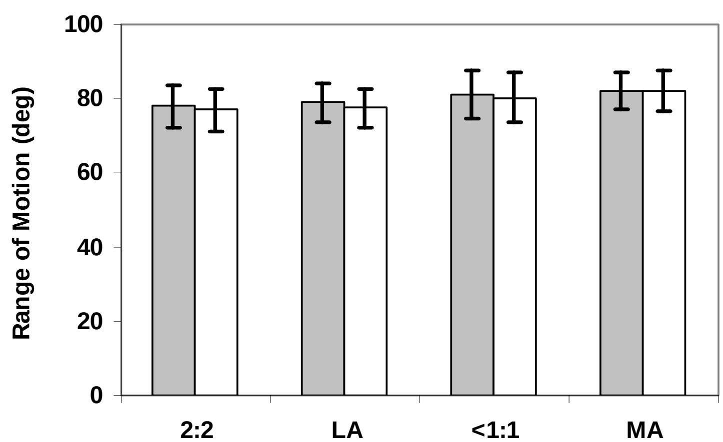


Figure 32: Average range of motion (±SD) in the concentric (dark bars) and eccentric phase in all movement strategies.

Moment of Elastic Curve Shift

Thera-Band® elastic bands have previously been shown to exhibit linear force strain characteristics (Hughes 1999; Patterson 2001). This relationship was observed in our study. We also observed a systematic shift in elastic band recoil force which occurred between the concentric and eccentric phases of the exercise. This force curve shift was consistent across trials and subjects. A small subtle systematic rotation of the trunk away from the band attachment point occurred in the concentric phase of the exercise. Concurrent translation of the elbow towards the band attachment point also occurred in the concentric phase. Both of these events resulted in higher elastic strain relative to matched joint angles in the eccentric phase. In the eccentric phase the opposite motions occurred with trunk rotation towards the band attachment site and concurrent elbow translation away from the band attachment point, resulting in reduced elastic strain. The elastic force curve shift was also observed in elastic moment data as would be expected (Melastic = recoil force x Moment arm).

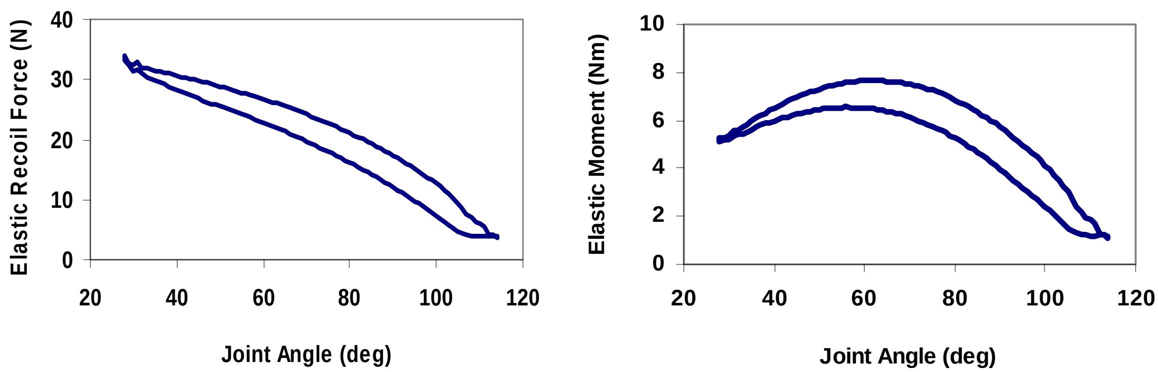


Figure 33: Elastic recoil force (left graph) and Moment of elastic (right graph) through 180° of angular excursion (90° concentric, 90° eccentric) in the 2:2 cadence strategy. Concentric phase is represented by the top blue line in both graphs. Elastic moment is lower in the eccentric phase at corresponding joint angles as a result of lower elastic recoil force.

Repetition Duration

The number of sampled data points in each repetition was multiplied by 0.005 seconds to determine repetition duration in seconds. All data was sampled at 200 Hz. Concentric and eccentric phase durations were delineated at repetition mid point.

Within strategies repetition duration differences were assessed using repeated measures ANOVA. Between repetitions duration was highly consistent in all strategies with one exception. Eccentric phase duration of the last repetition in the moderate acceleration strategy (0.58 ± 0.23s) was less than eccentric phase duration of all other repetitions, and was significantly less than repetition two (0.85 ± 0.14s, 0.27s difference, $p<0.004$) and repetition three (0.83 ± 0.15s, 0.24s difference, $p<0.004$). Subject altered their movement strategy in the last repetition in the anticipation that they would not be required to perform another repetition. Subjects tended to speed up in the last repetition to "get the set over with." In the moderate acceleration strategy and <1:1 cadence strategy, average duration of the concentric phase of the first repetition was shorter than duration of all subsequent repetitions. This difference did not reach statistical significance.

The first repetition may have been used to familiarize to a novel movement condition (no subjects had used fast cadence or moderate acceleration strategies previously in resistance training). Subjects tended to yank relatively harder on the first repetition and then subsequently slowed down to reduce loading and potential trauma to shoulder No evidence of effects of fatigue were found (i.e. there was no systematic trend to shorter repetition durations as the set of six progressed other than the last repetition of the moderate acceleration strategy described above). Repetitions were collapsed and separated into concentric and eccentric phases for all strategies.

Average repetition durations between strategies were compared using repeated measure ANOVA. Significant differences were found between all strategies in concentric and eccentric phases with the exception of the concentric phase in the moderate acceleration strategy and <1:1 cadence strategy. Concentric phase duration in the 2:2 cadence strategy (1.43 ± 0.31s) was greater than concentric phase duration in all other strategies ($p<0.001$). Concentric phase duration in the low acceleration strategy (1.09 ± 0.21s) was greater than concentric phase duration in the <1:1 cadence strategy (0.75 ± 0.25s, $p<0.001$) and moderate acceleration strategy (0.69 ± 0.17s, $p<0.001$). Concentric phase durations of the moderate acceleration strategy and <1:1 cadence strategy were not significantly different from each other. Eccentric phase duration in the 2:2 cadence strategy (1.62 ± 0.30s) was greater than eccentric duration in all other strategies ($p<0.001$). Eccentric phase duration in the low acceleration strategy (1.19 ± 0.28s, $p<0.01$) was greater than eccentric phase duration in the <1:1 cadence strategy (0.95 ± 0.22s, $p<0.01$), and moderate acceleration strategy (0.77 ± 0.20s, $p<0.001$). Eccentric phase duration in the <1:1 cadence strategy was greater than eccentric phase duration in the moderate acceleration strategy ($p<0.001$). Results of repetition duration analysis were not unexpected based on cueing instructions.

Average concentric and eccentric phase duration differences within strategies were assessed using paired t-test. Concentric phase duration was significantly less than eccentric phase duration in all strategies. Results are summarized below:

- Moderate acceleration strategy: Concentric duration (0.69 ± 0.17s): eccentric duration (0.77 ± 0.20s, 11% difference, $p<0.001$).
- Low acceleration strategy: Concentric duration (1.09 ± 0.21s): eccentric duration (1.19 ± 0.28s s, 9% difference, $p<0.00015$).
- 2:2 cadence strategy: Concentric duration (1.43 ± 0.31s): eccentric duration (1.62 ± 0.30s s, 11% difference, $p<0.001$).
- <1:1 cadence strategy: Concentric duration (0.75 ± 0.25s): eccentric duration (0.95 ± 0.22s s, 22% difference**,** $p<0.001$).

The greatest differential of mean duration between concentric and eccentric phases (22%) occurred in the <1:1 cadence strategy. This was not surprising as this was the only strategy in which subjects were instructed to perform the concentric phase faster than eccentric phase. Differences observed in other strategies were not expected and may have been the result of a tendency to reduce speed when moving in the same direction as the elastic band is pulling to minimize the risk of the arm snapping back with potential resulting injury. Correspondingly average P3 and P4 Iα which occurred in the eccentric phase were lower than average P1 and P2 Iα in all conditions.

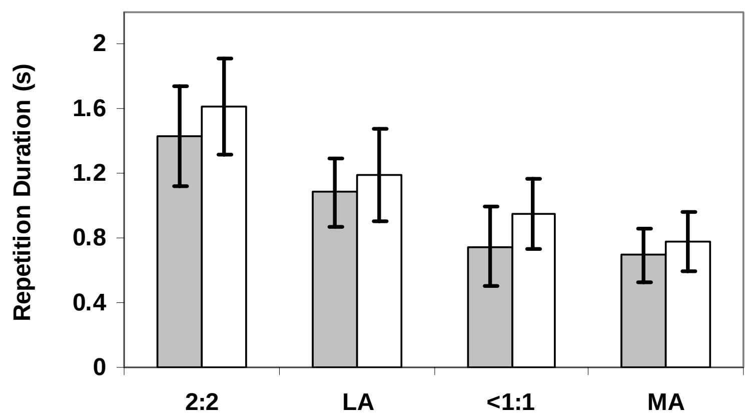


Figure 34: Average (±SD) repetition duration for all strategies in concentric (dark bars) and eccentric phases. Repetition durations were different between all strategies in concentric (p<0.001) and eccentric (p<0.001) phases except in the concentric phase in the moderate acceleration and <1:1 strategies. All four strategies demonstrated significant concentric-eccentric differences.

Adherence to Cuing

Average repetition duration data was examined to determine adherence to cuing. Average repetition duration in the <1:1 cadence strategy (0.75 ± 0.25s concentric, 0.95 ± 0.22s eccentric) very closely matched the desired cadence subjects were instructed to perform. Average repetition duration in the 2:2 cadence strategy (1.43 ± 0.31s concentric: 1.62 ± 0.30s eccentric) was lower than the cadence subjects were instructed to perform

Relationship between Repetition Duration and Repetition Velocity

As a result of low range of motion variation, repetition velocity was determined primarily by repetition duration. Stepwise linear regression was performed between repetition velocity, duration and range of motion for the concentric and eccentric phases in all conditions (scatter plots and regression). Repetition duration correlated strongly with repetition velocity in all strategies with R^2 values ranging between 0.91 and 0.97.

Repetition Velocity

Repetition range of motion (deg) was divided by repetition duration (sec.) to determine average angular forearm velocity in each repetition. Between repetition differences within strategies were assessed using repeated measures ANOVA. Significant repetition effects occurred in the eccentric phase of the moderate acceleration strategy (F= 7.015, df5, p<0.001) where repetition six (146 ± 33º/s) was greater than repetition one (103 ± 33º/s, p<0.037), two (97 ± 14ºs, p<0.009), and three (104 ± 24º/s, p<0.009). This corresponds to repetition duration analysis where repetition six in the moderate acceleration strategy was less than all other repetitions. In the moderate acceleration strategy and <1:1 cadence strategy, average velocity in the concentric phase of the first repetition was greater than all subsequent repetitions. These differences did not reach statistical significance different due to high variation and also correspond to results in repetition duration analysis.

Average repetition velocity in the concentric and eccentric phases was compared within strategies using paired t-tests. Concentric velocity was significantly greater than eccentric velocity in all strategies. Results are summarized below:

- Low acceleration strategy: Concentric (75.5 ± 16.2º/s): Eccentric (68.8 ± 17.4º/s), 9 % difference, p<0.001
- Moderate acceleration strategy: Concentric (124 ± 26.5º/s): Eccentric (114 ± 34.1º/s), 8 % difference, p<0.001
- 2:2 cadence strategy: Concentric (57.3 ± 14.4º/s): Eccentric (49.4 ± 10.7º/s), 14% difference, p<0.001
- <1:1 cadence strategy: Concentric (121 ± 42.8º/s): (88.9 ± 20.6º/s), 26% difference p<0.001.

These findings also correspond to results in repetition duration analysis.

Average repetition velocity between strategies was compared using repeated measures ANOVA. Results revealed significant differences in concentric and eccentric phases. Post hoc Tukey's analysis revealed average velocity in the concentric phase of the 2:2 cadence strategy (57.3 ± 14.4º/s) was significantly lower than average velocity in all other strategies (p<0.001). Concentric phase velocity in the low acceleration strategy (75.5 ± 16.2º/s) was significantly lower than concentric phase velocity in the 1:1 cadence strategy (121 ± 42.8º/s) and moderate acceleration strategy (124 ± 26.5º/s) (p<0.001). Concentric phase velocity in the <1:1 cadence strategy and moderate acceleration strategy were not significantly different from each other (p=0.25). Eccentric phase velocity in the 2:2 cadence strategy (49.4 ± 10.7º/s) was significantly lower than eccentric phase velocity in all other strategies (p<0.001). Eccentric phase velocity in the low acceleration strategy (68.8 ± 17.4º/s) was significantly lower than eccentric phase velocity in the <1:1 cadence strategy (88.9 ± 20.6º/s. p<0.001) and moderate acceleration strategy (114 ± 34.1º/s) (p<0.001). Eccentric phase velocity in the <1:1 cadence strategy was significantly lower than eccentric phase duration in the moderate acceleration strategy (p<0.001). All of these observations corresponded to results in the repetition duration analysis section.

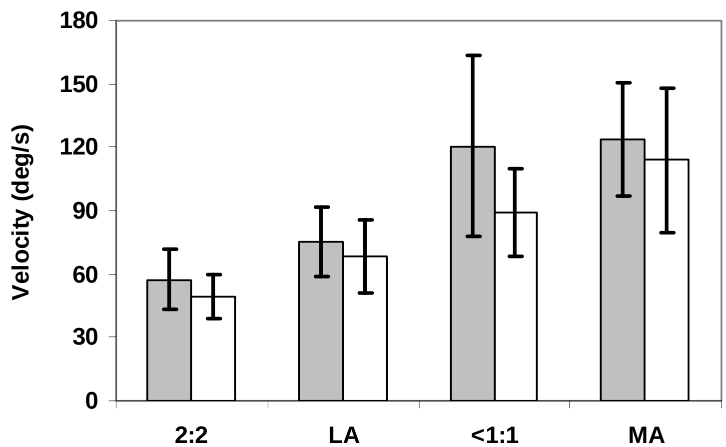


Figure 35: Average (±SD) repetition velocity for the concentric (dark bars) and eccentric phase of all movement strategies. Velocity was significantly different between all strategies in the concentric (p<0.001) and eccentric (p<0.001) phases except in the concentric phase of the moderate acceleration strategy and <1:1 cadence strategy.

Maximum Isometric Shoulder Internal Rotation Moment

Average maximum isometric shoulder internal rotator moment in subjects with recent upper extremity resistance training (32.8 ± 8.9 Nm, n=5) did not differ significantly from the average maximum isometric shoulder internal rotation moment in subjects without experience in upper extremity resistance training (29.0 ± 1.4 Nm, n=5) (independent t-test). There was no difference between resistance and non resistance trained groups when maximum isometric shoulder internal rotation moment was normalized to body weight (0.44 Nm/kg trained vs. 0.44 Nm/kg untrained).

RJM Relative to Maximum Internal Rotator Moment

Theoretically, optimal neuromuscular loading from resistance exercise occurs when the external moment most closely matches individual maximum moment generating capabilities at each joint angle through the range of motion of the exercise. The RJM/joint angle curves resulting from the elastic resistance exercise in the low and moderate acceleration strategies were compared to the maximum shoulder internal rotator moment curves obtained from a group of 27 normal subjects with no previous shoulder pathology tested on a Kin Com 500H dynamometer at 90º/s and 120º/s (Lori DePauw, M.Sc. Thesis). Maximum shoulder internal rotator moments did not differ significantly at these two test velocities. RJM in the low acceleration strategy ranged from 16% (at 110º shoulder rotation) to 42% (at ~40º shoulder rotation) of maximum moment through the range of motion of the exercise. RJM in the moderate acceleration ranged from 13% (at ~120º) to 60% (at ~80º) of maximum moment through the range of motion (**Figure 35**). RJM at 60% of maximum represents an equivalent to a 14 RM weight.

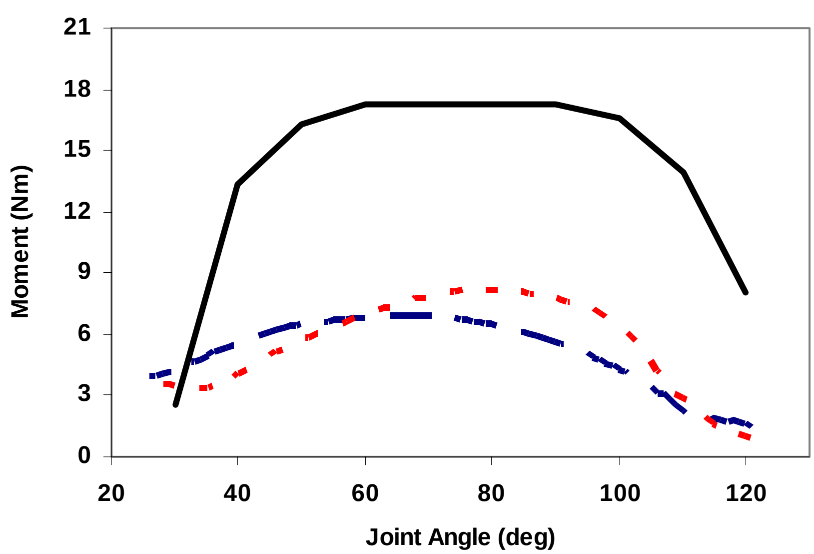


Figure 36: RJM in the moderate (dotted light red line) and low (dashed dark blue line) acceleration strategies in the concentric phase of shoulder rotation exercise. Maximum shoulder rotation concentric moment determined at 120º/s is displayed by the dark black line.

Summary of Results

1. RJM in the moderate acceleration strategy was significantly greater than RJM in the low acceleration strategy at 82% of joint angles in the initial phase of exercise from ~120º to ~60º range of motion. (**Figure 10**).
    I. RJM in the moderate acceleration strategy was up to 50% greater than RJM in the low acceleration strategy in this period (at ~98º of shoulder rotation).
    II. Peak RJM in the moderate acceleration strategy was 22% greater than peak RJM in the low acceleration strategy in this phase (**Figure 11**).
    III. Iα in the moderate acceleration strategy was up to 860% greater than Iα in the low acceleration strategy in this period (-2.9 Nm vs. -0.30 Nm). Iα in the moderate acceleration strategy was significantly greater than Iα in low acceleration strategy through 91% of joint angles in this period (**Figure 8**).
2. RJM in the <1:1 cadence strategy was significantly greater than RJM in the 2:2 cadence strategy at 84% of joint angles in the initial phase of exercise from ~115º to ~56º range of motion. (**Figure 16**).
    I. RJM in <1:1 cadence strategy was up 47% greater than RJM in the 2:2 cadence strategy in this period (at 100º shoulder rotation).
    II. Peak RJM in the <1:1 cadence strategy was 13% greater than peak RJM in the 2:2 cadence strategy in this period (**Figure 17**)
    III. Iα in the <1:1 cadence strategy was up to 1160% greater than Iα in the 2:2 cadence strategy in this period (-2.90 Nm vs. -0.23 Nm) (**Figure 14**). Iα in the <1:1 cadence strategy was significantly greater than Iα in the 2:2 cadence strategy at 84% of joint angles in this period (**Figure 14**).
3. In the second period of the concentric phase (from ~60º to 30º), and the initial period of the eccentric phase (30º to ~75º), RJM in the moderate acceleration strategy was significantly lower than RJM in the low acceleration strategy at 81% of joint angles (**Figure 10**)
    I. RJM in the moderate acceleration strategy was up to 32% lower than RJM in the low acceleration strategy in this period (at joint angle ~38º)
    II. Iα in the moderate acceleration strategy was up to 330% greater (<u>positive acceleration,</u> 2.6 Nm vs. 0.6 Nm) than Iα in the low acceleration in this period

(**Figure 8**). Iα in the moderate acceleration strategy was significantly greater than Iα in the low acceleration strategy at 85% of joint angles in this period (**Figure 8**).

4. In the second period of the concentric phase (from ~ 56º to 28º), and initial period of the eccentric phase (28º to ~73º), RJM in the <1:1 cadence strategy was significantly lower than RJM in the 2:2 cadence strategy at 83% of joint angles (**Figure 16**)
   I. RJM in the <1:1 cadence strategy was up to 30% lower than RJM in the 2:2 cadence strategy in this period (at joint angle ~30º).
   II. Iα in the <1:1 cadence strategy was up to 300% greater (positive acceleration, 1.75 Nm vs. 0.43 Nm) than Iα in the 2:2 cadence strategy in this period (**Figure 14**). Iα in the <1:1 cadence strategy was significantly greater than Iα in the 2:2 cadence strategy at 86% of joint angles in this period (**Figure 14**).
5. RJM in the moderate acceleration strategy was not significantly different from RJM in the <1:1 cadence strategy at any joint angles in the concentric phase of exercise (**Figure 23**).
6. In the initial period of the eccentric phase (from 30º to ~70º) RJM in the moderate acceleration strategy was significantly lower than RJM in the <1:1 cadence strategy at 55% of joint angles (**Figure 23**)
   I. RJM in the moderate acceleration strategy was up to 25% lower than RJM in the <1:1 cadence strategy in this period.
   II. Iα in the moderate acceleration phase was up to 200% greater (positive acceleration, 1.7 Nm vs. 0.56 Nm) than Iα in the <1:1 cadence strategy in this period. Iα in the moderate acceleration strategy was significantly greater than Iα in the <1:1 cadence strategy at 56% of joint angles in this period (**Figure 20**).
7. In the second period of the eccentric phase (from ~70º to 120º) RJM in the moderate acceleration strategy was significantly greater than RJM in the <1:1 cadence strategy at 62% of joint angles (**Figure 23**).
   I. RJM in the moderate acceleration strategy was up to 95% greater than RJM in the <1:1 cadence strategy in this period (at ~110º).
   II. Peak RJM was 11% higher (6.9 Nm vs. 6.3 Nm) in the moderate acceleration strategy in this period.
   III. Iα in the moderate acceleration strategy was up to 107% greater (-2.24 Nm vs. -1.08 Nm) than Iα in the <1:1 cadence strategy in this period. Iα in the moderate

acceleration strategy was significantly greater than $I\alpha$ in the <1:1 cadence strategy at 75% of joint angles (**Figure 20**).

8. RJM in the low acceleration strategy with 30% elastic start strain was significantly greater than RJM in the low acceleration strategy with 0% start strain through 100% of angular excursion of the exercise (180° of 180°) (**Figure 28**).
    I. RJM in the 30% strain condition was up to 500% greater than RJM in the 0% strain condition. Greatest RJM differences occurred between 120º to 80º range of motion.
    II. At all joint angles, RJM differences were the result of significant elastic moment differences between elastic start strain conditions.
9. Brief instruction followed by short demonstration and verbal correction resulted in effective acceleration control. We were able to induce distinct limb acceleration control in the low and moderate acceleration strategies. Load can be controlled through this technique, which has not been previously shown.
10. Brief instruction followed by short demonstration and verbal correction was effective in achieving the desired cadence in the <1:1 cadence strategy.
11. Brief instruction followed by short demonstration and verbal correction was less effective in achieving the desired cadence in the 2:2 cadence strategy.
12. No significant change in limb acceleration ($I\alpha$) occurred when low acceleration strategy was performed at 30% elastic start strain (**Figure 26**).
13. A small increase in limb acceleration ($I\alpha$) occurred at the initiation of exercise when moderate acceleration strategy was performed at 30% elastic start strain compared to 0% elastic start strain. The overall movement strategy (acceleration profile) was similar with 0% and 30% elastic start strains with moderate acceleration strategy cueing (**Figure 30**).
14. Average mechanical work was equivalent in the low acceleration and moderate acceleration movement strategies. More work was performed in the concentric phase (9.1 Joules) relative to the eccentric phase (7.3 Joules) in both strategies due to the downwards shift of the elastic moment curve during the eccentric phase**.** Relative to low acceleration strategy, moderate acceleration strategy resulted in 15% more work in the first ¼ of repetition (5.9 Joules vs. 5.1 Joules), 19% less work in the middle ½ of repetition (6.9 Joules vs. 8.6 Joules), and 37% more work in the last ¼ of repetition (3.7 Joules vs. 2.7 Joules) (**Figure 13**). Greatest relative work differences occurred in the last ¼ of repetition when elastic moment magnitudes were lowest and $I\alpha$ differences between strategies had the greatest influence on RJM and the resulting work performed. Absolute work difference between strategies was zero at the end of repetition, however relative

work differences vary over the four phases of the repetition (corresponding to P1, P2, P3, and P4 $I\alpha$) as a function of the pattern of the RJM curves which are not symmetrical and change over the course or the repetition and are also shifted between concentric and eccentric phases.

15. Average mechanical work was equivalent in the 2:2 and <1:1 cadence strategies. More work was performed in the concentric phase (8.9 Joules) relative to the eccentric phase (7.3 Joules) in both strategies due to the downwards elastic moment curve shift during the eccentric phase. Relative to 2:2 cadence strategy, <1:1 cadence strategy resulted in 14% more work in the first ¼ of repetition (5.6 Joules vs. 4.9 Joules) , 13% less work in the middle ½ of repetition (7.6 Joules vs. 8.7 Joules), and 11% more in the last ½ of repetition (3.0 Joules vs. 2.7 Joules) (**Figure 19**). Absolute work differences between strategies was zero at the end of repetition, however relative work differences vary over the four phases of the repetition (corresponding to P1, P2, P3, and P4 $I\alpha$) as a function of the pattern of the RJM curves which are not symmetrical and change over the course or the repetition and also are shifted between concentric and eccentric phases.

Discussion

In practice and in research on resistance exercise, the load is often over-simplified as simply arising from the mass (or weight) of a barbell or dumbbell, the thickness of elastic resistance (coded by color), or mass used in CAM or pulley based exercise machines. The characterization of load based on these parameters alone omits the contribution of limb or trunk acceleration among other factors such as moment of weight, inertia, etc. These factors are clearly identified by Newtonian mechanics as a contributor to the instantaneous load.

One aspect that is often discussed but rarely quantified in resistance training settings to control load is movement strategy. Movement strategy is poorly defined in the exercise literature and is primarily limited to the control of average limb velocity during exercise (e.g. velocity control with isovelocity dynamometer, cadence control in resistance exercise). Limb motion reflects the loading of the tissues as joint torques are proportional to limb acceleration (not velocity per se). The results of this study clearly indicate that movement strategy in terms of acceleration has a substantial influence on RJM and must therefore be considered in elastic exercise prescription and guidelines.

In the <u>moderate acceleration</u> strategy which manifested as higher acceleration magnitude in relation to the <u>low acceleration strategy</u>, three prominent changes in the joint moments were revealed:

1. Peak RJM (P1) increased by 20% (8.9 Nm vs. 7.4 Nm).
2. Angle of peak RJM shifted from ~70º to ~85º in the concentric phase (peak P1 RJM) and from ~60º to ~90º in eccentric phase (peak P4 RJM).
3. RJM was reduced by up to 32% during the middle repetition range of motion corresponding to P2 and P3 Iα.

RJM represents a potential stimulus for neuromuscular adaptation in which higher RJM would result in greater stimulus for muscle hypertrophy and/or neural adaptations (e.g. greater motor unit recruitment, improved motor unit coordination, increased motor unit activation frequency etc.). Peak RJM in the moderate acceleration strategy changed by 20% and this represents a substantial increase in loading on the shoulder internal rotators. The shift in the RJM joint angle curve in the moderate acceleration strategy was likely the result of a change in neuromuscular activation strategy and would alter the loading pattern at the shoulder internal rotator muscles. Angle specific loading would be different compared to the low acceleration strategy, and the stimulus to neuromuscular adaptation would change such that greater stimulus would be imparted from 120° to ~70° of shoulder rotation, and lower stimulus would be imparted from ~70° to 30°

of shoulder rotation relative to the low acceleration movement strategy. RJM in the low acceleration movement strategy closely matches the elastic moment profile through the range of motion.

In the <1:1 cadence strategy which manifested as higher acceleration magnitude in relation to the 2:2 cadence strategy three prominent changes in the joint moments occurred:

1. Peak RJM (P1) increased by 15% (8.6 Nm vs. 7.5 Nm).
2. Angle of peak RJM shifted from ~65º to 80º in the concentric phase (peak P1 RJM), and from ~60º to 70º in the eccentric phase (peak P4 RJM).
3. RJM was reduced by up to 30% in the middle range of motion of the repetition corresponding to P2 and P3 $I\alpha$.

Peak RJM in the <1:1 cadence strategy increased by 15% and would therefore represent an increase in loading on the shoulder internal rotators. Relative to the 2:2 cadence strategy greater neuromuscular stimulus was imparted from 120° to ~65° of shoulder rotation, and lower neuromuscular stimulus was imparted from ~65° to 30°. In the low acceleration and 2:2 cadence strategies, load is well represented by the moment of elastic.

The imposition of higher accelerations (e.g. moderate acceleration strategy and <1:1 cadence strategy) resulted in a substantial change in the magnitude and pattern of neuromuscular loading during the exercise. Summarizing the effects of higher acceleration movement strategies in shoulder rotation elastic resistance exercise:

1. RJM was greater than RJM in low acceleration and 2:2 cadence strategies during P1 and P4 $I\alpha$ (from 120º to ~60º to 70º concentric phase and from ~ 70 to 75º to 120º eccentric phase) .
2. RJM was lower than RJM in low acceleration and 2:2 cadence strategies during P3 and P4 $I\alpha$ (from ~ 60º to 70º to 30º concentric phase and from 30º to ~ 70º to 75º eccentric phase).

The pattern and magnitude of RJM in the moderate acceleration strategy manifest a similar pattern and magnitude to RJM in the <1:1 cadence strategy in the concentric phase. Significant differences were apparent in the eccentric phase between these two strategies. These results would indicate that cadence control which is currently the only form of motion control employed in resistance exercise prescription is not necessarily the same as acceleration control.

Relative to 0% elastic start strain the imposition of 30% elastic start strain in the low acceleration strategy resulted in:

1. Peak RJM (P1) increased by 50% (10.3 Nm vs. 6.9 Nm).

2. Angle of peak P1 RJM remained unchanged with both start strain strategies (~66º concentric phase)
3. No evidence was found for elastic strain dependent movement strategy changes. The pattern and magnitude of Iα was not significantly different when low acceleration strategy was employed with either start strain.
4. Neuromuscular stimulation would be up to ~50% higher from ~80° to 30°, and up to 500% greater from ~80° to 120°.

The change in loading pattern resulting from elastic strain increase is important to know and could be utilized advantageously for targeted shoulder rehabilitation goals. Elastic resistance exercise guidelines (Page, P., Ellenbecker, T. 2003) recommend that resistance intensity progression be achieved exclusively from progressing to the next level (color coded thickness) of elastic band. These authors discourage elastic starting strain increases as a means of elastic load progression because they state that the kinematics of the exercise would change resulting in a less optimal elastic loading profile relative to the same exercise performed with 0% elastic start strain (Page, P., Ellenbecker, T. 2003). Our results clearly indicate that exercise kinematics did not change significantly with elastic start strain increases in both low and moderate acceleration strategies. Elastic start strain increases could be utilized as a means of load progression and may in fact be a superior method to band color (thickness) progression as the pattern of the elastic moment curve remained unchanged with 30% start strain in the low acceleration strategy, unlike the change in elastic moment curves reported by Hughes et al. (1999) when elastic band colors are progressed to the next level (**Figure 2**).

Currently elastic resistance literature provides information on elastic load based on:

1. the stress/strain profiles of elastics, and
2. the elastic moment joint angle graphs/profiles in static loading conditions based on two studies examining shoulder abduction and elbow flexion.

Visual inspection of variances of the RJM/ jt. angle plots revealed small inter-individual RJM differences in all four movement strategies. ANOVA testing revealed significant peak Iα and peak RJM differences between movement strategies. Based on these results we proceeded to a paired t-test analysis of the data incorporating our mathematical simulation to guard against Type 1 errors. By employing joint angle averaged data we were able to compute and compare the RJM in different elastic loading conditions over 180 joint angles thus providing the most comprehensive description of RJM ever in elastic resistance exercise. The RJM joint angle curves produced in this study provide much greater information on elastic loading during exercise than previous studies as they include both elastic moment and acceleration dependent moment (Iα) calculations. Based on this study one can clearly see that the pattern of load from elastic resistance can not be predicted by elastic stress strain or static elastic moment curves alone. Unlike previous literature which used a static analysis of elastic resistance exercise and only considered the moment of elastic force as a loading mechanism, we have clearly shown that the moment of elastic force is only one component to consider and does not predict the pattern and magnitude of loading when the exercise is done with accelerated movements (e.g. moderate acceleration and <1:1 strategies). The RJM curves in the moderate acceleration and <1:1 cadence are substantially different in pattern than those derived from the static analysis by Hughes et al (1999). To date, the optimal RJM profile from elastic resistance exercise for maximal training response and minimal risk of injury are not known. Indeed, the movement strategy dependent differences in loading may be a means to adapt the elastic exercise to specific functional demands. The progression from low acceleration to higher acceleration may be based upon task specificity.

Some authors have suggested that a fundamental advantage of elastic resistance over traditional isotonic or free weight resistance (e.g. dumbbells or barbells) is that individuals are not able to "cheat" by using limb momentum to complete the exercise, and therefore the load on muscles is maintained through the entire range of motion (Page, P., Ellenbecker, T. 2003). We have clearly shown that in the moderate acceleration and <1:1 cadence strategies the load during approximately half of the repetition is lower than elastic moment alone as a result of P2 and P3 $I\alpha$. The magnitudes of $I\alpha$ would increase with either the addition of an implement (e.g. dumbbell) or greater limb acceleration resulting in greater acceleration dependent moments. This would result in greater load at P1 and P4 $I\alpha$ and greater load reduction at P2 and P3 $I\alpha$. Most dynamic resistance exercise involves repetitive or cyclical movements in which the repetition is completed at or near the same position as the start. This necessarily requires four acceleration phases to complete the repetition as previously described (P1, P2, P3, and P4 $I\alpha$ **- Figure 7**). Based on Newtonian Mechanics, all dynamic resistance exercise (except in isokinetic conditions) will therefore involve periods of "cheating" corresponding to limb or trunk deceleration when the load will be reduced relative to the moment of external resistance produced from an elastic or dumbbell. Consideration should be given in resistance exercised guidelines to deceleration, and potentially attempts should be made to minimize it in certain scenarios in order prevent decreased neuromuscular loading at those regions of range of motion (e.g. ½ repetitions in the outer range of motion of the exercise, ballistic training, etc.).

In this study, the following hypotheses were supported:

1. RJM in the moderate acceleration strategy was greater than RJM in the low acceleration movement strategy at the initiation of movement.
2. RJM in the <1:1 cadence strategy was greater than RJM in the 2:2 cadence strategy at the initiation of movement.
3. RJM in the end of the concentric phase and beginning of eccentric phase of movement was diminished in the moderate acceleration movement strategy relative to the low acceleration movement strategy.
4. RJM during the end of the concentric phase and beginning of eccentric phase of movement was diminished in the fast cadence strategy relative to the slow cadence strategy.
5. RJM at the beginning of the eccentric phase was reduced in the moderate acceleration strategy relative to the fast cadence strategy.
6. RJM at the beginning of the eccentric phase was reduced in the moderate acceleration strategy relative to the fast cadence strategy due to deceleration effects.
7. RJM through the entire angular excursion of the exercise in the low acceleration strategy was higher with the 30% elastic start strain compared to 0% elastic start strain.
8. In the low acceleration strategy, 30% starting elastic strain did not result in a moderate acceleration movement pattern.
9. Brief instruction followed by short demonstration and verbal cuing was effective for acceleration control and resulted in a clear distinction between low and moderate acceleration strategies.
10. Brief instruction followed by short demonstration and verbal cuing was effective in achieving the desired cadence control in the <1:1 cadence strategy.

The following hypotheses were not supported:

1. RJM in the moderate acceleration movement strategy would be higher than the RJM in the fast cadence movement strategy at the initiation of exercise. RJM was not significantly different between these two strategies in the concentric phase.
2. In the moderate acceleration strategy, 30% starting elastic strain did not result in a significant change in the movement strategy (i.e. moderate acceleration did not shift to a higher acceleration pattern with 30% elastic start strain).

There was however a short period at the initiation of exercise (from~120º to 100º) where the RJM magnitude from 30% start strain was greater then RJM magnitude at 0% strain. Overall, the pattern and magnitude of RJM was similar between both starting strain conditions and differences could be largely attributed to strain effects.

This study provides the first comprehensive description of the loading characteristics of elastic exercise under various conditions. The imposition of higher acceleration in this cyclical elastic resistance exercise resulted in a consistent loading pattern that was significantly different from the same exercise performed with lower acceleration strategies through the majority of range of motion. Higher acceleration movement strategies resulted in two primary changes in the RJM profile relative to the lower acceleration movement strategies:

1. RJM was greater at the initial and ending periods of the repetition when forearm acceleration was directed away from the line of pull of elastic, and
2. RJM was reduced in the middle period of the repetition when forearm acceleration is directed towards the line of pull of elastic.

The load from higher acceleration movement strategies employed during this elastic resistance exercise was significantly altered resulting in a change in the neuromuscular loading pattern about the shoulder. This would result in significantly different stimulation to neuromuscular adaptation throughout the range of motion of the exercise. Static assessment of elastic loading is insufficient to characterize loading during elastic exercise. This study also revealed a shift in the position (joint angle) of peak RJM due to acceleration effects in moderate and 1:1 strategies.

Cadence control is not equivalent to acceleration control in terms of the magnitude and timing of load. Both cadence and acceleration movement strategies can be successfully cued. Cadence control manifests changes in load by a change in acceleration.

This is the first exercise study to utilize multiple comparison t-tests on time normalized data using regional correction method for type I errors. This methodology permitted angle by angle comparisons of the data resulting in comprehensive analysis of the RJM between conditions.

Future Research

Based on the results of this study future research should be directed towards investigating:

- Comparisons between elastic and dumbbell resistance exercise during a shoulder abduction exercise employing a <1:1 cadence has been previously reported (Kyle Turcotte, Masters Thesis, University of Manitoba, 2005). Results showed that all four Iα and RJM peaks were significantly different between resistance modes. Further analysis of the effect of movement strategy on RJM during free weight exercises is required. Movement strategy would potentially have greater effect on RJM in these conditions as a result of greater limb moment of inertia (I). The moment of inertia added to the exercising system in elastic is very low. An analysis of elastic loading with exercise performed at different joints is also required (e.g. knee, hip, shoulder abduction).
- The impact of metronome timing of cadence needs to be studied to determine if this induces higher acceleration effects than self-paced cadences.
- The extent of independence of average velocity (cadence) and acceleration magnitude needs to be further examined.
- The range of acceleration magnitudes used in real life settings (sport and rehabilitation) needs to be established.
- A study examining the differences in training adaptations (e.g. maximum isometric torque, velocity specific joint angle maximum torque, hypertrophy, functional performance measures) between moderate acceleration loading (or <1:1 cadence) and low acceleration (or 2:2 cadence) loading needs to be performed.

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Appendix A

RESEARCH PARTICIPANT INFORMATION AND CONSENT FORM

Title of Study: Characterization of loading about the shoulder induced by systematic variations of "Thera Band" Exercise.
Principle Investigator: Greg Hodges, MS 549 Health Sciences Centre General Hospital, 787-2389
Co-Investigator: Dr. Dean Kriellaars, RR303 Rehabilitation Hospital, 787-3505

You are being asked to participate in a research study. Please take your time to review this consent form and discuss any questions you may have with the study staff. This consent form may contain words that you do not understand. Please ask the study staff to explain any words or information that you do not clearly understand.

Purpose of Study
This research study is being conducted to determine the load produced on your shoulder from exercising with elastic tubing. Understanding the load induced by a particular exercise will help guide practitioners to prescribe exercise that is more efficient, safe and specific. In order to achieve the desired exercise response of improving strength, a better understanding of the dose (i.e. prescribed exercise) /response (load) relationship is important.

We propose to estimate the loading at the shoulder by using a common biomechanical mathematical modeling technique, which incorporates the arm weight, inertia, and acceleration with the resistance provided by the elastic tubing to give an overall estimate of load during this particular exercise.

A total of 20 participants will participate in this study

Study Procedures

All participants are being recruited for this study via word of mouth.
You will be asked to attend one exercise session at the University of Manitoba Bannatyne Campus, Human Performance Laboratory, Rehabilitation Hospital, Health Sciences Centre, Winnipeg. Total time for this session will not exceed one hour.

Day of the week and time of day of this session will be scheduled to accommodate your schedule.

At the exercise session, your body weight and height will be measured. The range of motion of your right shoulder will also be measured.

Prior to performing any exercise you will view a demonstration and description of the shoulder external rotation movement to be performed. You will then have a short warm up period consisting of 5 minutes of arm calisthenics and a series of general shoulder stretches. You will then be asked to perform four sets of twelve repetitions of the exercise. Input from the investigators may be provided to ensure safe and proper technique.

Prior to the exercise session you will be asked to perform two separate maximum isometric internal rotation strength tests by pulling inwards as hard as you can against an exercise handle secured to a force measurement gauge secured to an exercise rack. You will pull in for four seconds and then rest for two minutes before repeating. The exercise session proper consists of four sets of six repetitions of standing shoulder internal rotation with elastic tubing using different movement control strategies. You will hold one end of elastic tubing in your right hand, with the other end of tubing attached to the force-recording gauge secured to the exercise rack. From the starting position with your forearm horizontal, elbow at 90°, and shoulder externally rotated, you will pull out on the band through 90° of shoulder rotation range of motion. The initial tubing length or amount of stretch will be changed for one set by getting you to stand further away form the exercise rack. The more the tubing is stretched initially and through the range of motion, the greater the resistance it provides. A lightweight miniature accelerometer will be attached to the back of your right wrist during the exercises. A plastic electrogoniometer will be secured with tape to your forearm to measure the range of motion during the exercise.

The exercise session will be stopped if:
-You wish to stop for any reason
-You exhibit signs of pain or severe discomfort
-You use unsafe technique in the performance of the exercise.

The researcher may decide to take you of this study if you are unable to perform the exercise properly (i.e. safely).

You can stop participating at any time. However if you decide to stop participating in this study, we encourage you to talk to the study staff first.

Risks and Discomforts
After the exercise session, it is common to feel some discomfort in the shoulder muscles involved in performing the exercise. This very minor discomfort will likely last up to 5 days peaking at 2 days after the exercise. This is a normal consequence of exercise and is called delayed onset muscle soreness.

Minor discomfort may be felt during the exercise session (as with any form of exercise). However, if obvious pain arises at any time during the session, the session will be discontinued.

Benefits
There may or may not be direct benefit to you from participating in this study. We hope the information learned from this study will benefit other people in the rehabilitative and resistance training settings.

Costs
All the procedures, which will be performed as part of this study, are provided at no cost to you.

Payment
You will receive no payment or reimbursement for any expenses related to participating in this study.

Confidentiality
Information gathered in this research study may be published or presented in public forums, however your name and other identifying information will not be used or revealed.
Absolute confidentiality cannot be guaranteed. Your personal information may be disclosed if required by law.

The University of Manitoba Health Research Ethics Board may review records related to this study for quality assurance purposes.

Voluntary Participation/Withdrawal from the Study
Your decision to take part in this study is voluntary. You may withdraw from the study at any time. Your decision not to participate or to withdraw from the study will not affect your care at this center.
If the study staff feels that it is in your best interest to withdraw you from the study, they will remove you without your consent.

We will tell you about any new information that may affect your health, welfare or willingness to stay in this study.

Medical Care for Injury Related to the Study
You are not waiving any of your legal rights by signing this consent form nor are you releasing the investigators or the sponsors from their legal and professional responsibilities.

Questions
You are free to ask any questions that you may have about your treatment and your rights as a research participant. If any questions arise during or after the study or if you have a research related injury, contact the study staff: Dr. Dean Kriellaars at 787-3505 or Greg Hodges at 787-2389.

For questions about your rights as a research participant you may contact The University of Manitoba, Bannatyne Campus Research Ethics Board Office at 789-3389.

Statement of Consent

I have read this consent form. I have had any questions regarding the study answered by the study staff in a language I understand. The risks and benefits have been explained to me. I understand that my participation in this study is voluntary and that I may choose to withdraw at ant time.

I understand that information regarding my personal identity will be kept confidential, but that confidentiality is not guaranteed. I authorize the inspection of any of my records that relate to this study by The University of Manitoba Research Ethics Board, for quality assurance purposes.

By signing this consent form I have not waived any of the legal rights I have as a participant in a research study.

Participant signature___________________________ Date____________________

Participant printed name____________________________

I, the undersigned, have fully explained the relevant details of this research study to the participant named above and believe that the participant has understood and has knowingly given consent.

Signature_________________________ Date________________

Printed Name_____________________________

Role in the Study___________________________________