Coventry University

MASTER OF SCIENCE BY RESEARCH

Effect of altering range of motion on muscle activation patterns when using the MuJo™ Shoulder Machines

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Award date:
2017

Awarding institution:
Coventry University

Effect of Altering Range of Motion on Muscle Activation Patterns when Using the MuJo™ Shoulder Machines.

By

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September 2017

Coventry University

A thesis submitted in partial fulfilment of the University's requirements for the Degree of Master of Science by Research

Certificate of Ethical Approval

Applicant:

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Project Title: The Evaluation of MuJo™ Fitness Equipment for the Assessment and Treatment of Upper Quadrant Function.

This is to certify that the above-named applicant has completed the Coventry University Ethical Approval process and their project has been confirmed and approved as Medium Risk

Date of approval:

20 January 2017

Project Reference Number:

P49614

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1RM = One repetition maximum

ARV = Average rectified value

CV = Coefficient of variation

DF = Degrees of freedom

EMG = Electromyography

MVC = Maximal voluntary contraction

RMS = Root mean square

ROM = Range of motion

UQ = Upper quadrant

Acknowledgements

I would first like to express my sincere gratitude to my supervisory team, for their collective assistance and direction, and also their much-appreciated individual input. Dr Sam Oxford for providing me with the opportunity to undertake this Master’s, and his guidance throughout, not least on the widely-acknowledged perils of electromyography! Dr Neil Clarke for his continued patience, willingness to still answer all my panicked questions, and always imploring me to relax. Dr Peter Mundy for also always being willing to answer my panicked questions, encouraging me to pursue biomechanics, and of course being right that I would end up pursuing biomechanics. I hope I’ve made you all proud.

Thank you also to the entire Sports Science department at Coventry University; it has been an amazing four years and I’m grateful to everyone that has taught and helped me throughout my time here. Thank you for every opportunity and piece of advice that has helped me reach each goal I have aimed for. I would also like to thank MuJo™ for graciously funding this Master’s; I will be forever appreciative of this opportunity.

Finally, I can never convey how thankful I am to have such incredible family and friends, who have continuously striven to provide me with every opportunity possible, alongside the tools to succeed. None of this would have been possible without you and your continuous support and patience, thank you.

1.0 Abstract

Resistance equipment is often restricted to a single plane of movement; however, multi-planar movements may be more effective, by facilitating the development of motor strategy and intermuscular coordination. Due to their moving axis cam technology, the MuJo™ External Shoulder Machine and Internal Shoulder Machine enable such movement. Furthermore, the range of motion (ROM) through which the shoulder travels can be adjusted to target specific muscles, which may have useful implications within a rehabilitation environment. However, little is currently known about the consequences of varying the ROM whilst using the devices. Therefore, the purpose of this study was to determine the effect of altering ROM on shoulder muscle activity during exercises performed on the MuJo™ Shoulder Machines.

Following institutional ethics approval, nine recreationally-active male participants (means ± SDs: age: 25 ± 5 years; body mass: 77.06 ± 11.06 kg; height: 1.76 ± 0.09 m) performed abduction and external rotation, and adduction and internal rotation of the shoulder at twelve different ranges of motion, in a randomised, counterbalanced order. Surface electromyography (EMG) was collected from the upper trapezius, anterior and posterior deltoids, infraspinatus, pectoralis major, and latissimus dorsi. Muscle activity was normalised to the peak activity from a one repetition maximum test, also performed on the machines. The effect of abduction/adduction and rotation angle on normalised peak EMG was analysed via a two-way analysis of variance with repeated measures for each muscle; effect sizes were calculated using partial eta squared (η^2_p).

For the External Machine, a significant main effect for abduction in the upper trapezius was found ($F_{(1.1,\ 8.7)}$=15.34, P=0.003, η^2_p=0.66). Electromyography amplitude was significantly higher at 90° of abduction than at 60° and 30°, and significantly higher at 60° than at 30°. For the anterior deltoid, EMG amplitude was significantly higher at 90° than at 60° of abduction ($F_{(2,16)}$=7.17, P=0.006, η^2_p=0.47). A significant main effect for rotation in the latissimus dorsi was found ($F_{(3,24)}$=7.96, P=0.001, η^2_p=0.50), with EMG amplitude significantly lower at 0° than at both 90° and 60°. For the Internal Machine, a main effect for rotation in the pectoralis major was observed ($F_{(3,\ 24)}$=6.98, P=0.002, η^2_p=0.47), with EMG amplitude significantly lower at 60° than 30° of rotation. No significant interactions or main effects were observed in the remaining muscles on either machine.

In conclusion, altering the ROM results in some changes in muscle activity during abduction and rotation, perhaps indicating a greater requirement for stabilisation during the less constrained repetitions. Further studies incorporating kinematics and inverse dynamics may provide deeper understanding into the effects on motor strategy that may occur when exercising with this device.

2.0 Introduction

The musculature of the upper quadrant (UQ) is used for a variety of everyday and sport specific tasks, yet until the 1970s, there was a distinct dearth of research concerning the body's response to dynamic UQ exercise (Sawka, 1986). Powerful movements of the UQ may be required for many industrial, agricultural, and military activities (Newton *et al.*, 1997; Sawka, 1986). Furthermore, the successful development and improvement of the UQ may be considered integral to many sports, including gymnastics (Marinšek, 2010), swimming (Hawley *et al*., 1992), paddle sports (i.e., canoeing and kayaking) (Ackland *et al*., 2003), and throwing sports such as javelin, hammer throw, and handball (Cook, 2006; Hermassi *et al.*, 2010). Kinematics, kinetics, and muscle activations via EMG measurements have been investigated prolifically in overhead throwing activities (DiGiovine *et al*., 1992; Escamilla *et al*., 1998; Escamilla *et al*., 2007; Fleisig *et al*., 1996). Peak scapular muscle activity is reported to be high during the cocking and deceleration phases of overarm throwing; additionally, during these phases, high rotator cuff activity has been noted, to help attenuate forces of around 80-120% of body weight (Escamilla and Andrews, 2009).

Strength and power in the UQ may also be considered determinants of successful performance in sports not dominated by overarm throwing, such as American Football (Fry and Kraemer, 1991) and Rugby (Baker, 2001). Additionally, overhead throwing activities are associated with shoulder and other UQ injuries (Braun *et al.*, 2009; Meister, 2000; Wilk *et al.*, 2002); a relatively high incidence of UQ injuries has also been found in Rugby Union (Bui-Mansfield *et al.*, 2007; Brooks *et al.*, 2005a; Brooks and Kemp, 2011; Headey *et al.*, 2007) and Rugby League (Gabbett, 2000; Gabbett and Jenkins, 2011). Most common within Rugby Union matches was injury to the acromioclavicular joint, yet the most severe injury was shoulder dislocation or instability (Headey *et al.*, 2007). Different injury locations in training have been noted for different positions; backs had a higher incidence of shoulder injury, but forwards had a higher prevalence of neck and spinal damage (Brooks *et* al., 2005b). Within swimmers, shoulder pain is the most commonly reported orthopaedic affliction (Wanivenhaus *et al.*, 2012), with shoulder impingement and muscle overuse being frequent issues. Additionally, within kayaking, 61% of injuries reported were to the upper extremity, with almost half of those occurring within the shoulder (Fiore and Houston, 2001). As levels of UQ strength and power may distinguish between players of different levels within the aforementioned sports, and may play some protective role against sustaining injury, it is argued that training for the UQ ought to be integral to optimum preparation of these athletes (Baker, 2001).

Increased frequency, duration, and intensity of exercise have been associated with less chronic pain within recreationally active populations (Landmark *et al*., 2011). In particular, for 20-64 year olds, pain was 10-12% lower with regular exercise of moderate intensity (Landmark *et al*., 2011), indicating that consistent bouts of exercise may play some protective role against chronic musculoskeletal pain. However, incidence of injury in recreational populations has also been noted to be high, particularly for those aged 5-24 years (Conn *et al*., 2003); injury to the upper extremity was found to be 31.2%, with 26.4% of those being to the shoulder or upper arm (Conn *et al*., 2003). Consequently, UQ training should also be a focus for recreational populations, in order to better allow the body to withstand repeated load and stressors.

Occurrence of musculoskeletal pain is also high within the general, inactive population; occurrence has been found to be higher in women, increase with age, and be more prevalent in psychologically stressed populations (Pribicevic, 2012). Work-related aspects are considered to be a great risk factor for developing musculoskeletal impairment, such as poor posture, performing the same activity repeatedly (such as typing or driving) (van der Windt *et al*., 2000), and working with arms above shoulder height (Leclerc *et al*., 2004; Pope *et al*., 1997). Location can also play a role, as it can often determine the occupational range of a population; for example, a rural environment may mean that many people will be employed in mostly physically demanding roles, thereby placing greater stress on workers (Harkness *et al*., 2005). However, even in less physical roles, there are often greater psychosocial demands within the workplace than before, which has also been shown to be a risk factor for onset of UQ pain (Harkness *et al*., 2003; van der Windt *et al*., 2000).

Whilst incidence of dysfunction has been shown to increase with age, shoulder instability is also a common cause of pain and dysfunction in younger adults (Song *et al*., 2015). Shoulder dislocations account for 4% of all injuries for people ages 20-30 years old, with most of these occurring during sport participation (Aune *et al*., 2012). However, when measuring the prevalence of musculoskeletal pain in different body regions (including the shoulder) within different age groups, Parsons *et al*. (2007) found that shoulder pain was most prevalent in 45-64 year-olds. Additionally, a higher incidence of chronic pain was observed in 55-64 year-olds (50%) than in 18-24 year-olds (23%), with wrist, elbow, shoulder, neck, and lower back pain peaking in the 45-54 or 55-64 age brackets (Parsons *et al*., 2007). The retrospective survey used to collect this data may be subject to recall bias (Raphael, 1987), yet the study nevertheless helps demonstrate the extent of the prevalence of musculoskeletal pain. Such high incidence of dysfunction and pain has vast economic implications, both on healthcare services and affected individuals due to absence from work (Pribicevic, 2012). Consequently, developing appropriate techniques to treat and prevent UQ impairment will have potentially extensive benefits for athletic, injured, and the general population as a whole.

The MuJo™ Multiple Joint Fitness System comprises resistance equipment that incorporates moving cam technology, to allow multi-directional movement. The upper body resistance machines resemble a traditional chest-press machine (see appendix 1 for illustration of the machines) but are designed to more closely replicate the shoulder's range of motion, instead of being restricted to one movement plane. As such, it is possible to incorporate many muscles of the shoulder and back at one time whilst exercising on one machine. Due to their unique multi-axial cam technology, the devices are purported to independently load and train multiple joints and muscles. In this way, the system integrates the unconstrained nature of free weights with the control provided by conventional resistance machines. Due to the independently moving parts, resistive force is always applied to the part that is moving rather than fixed, meaning the user should experience consistent loading throughout ROM. Further, due to the control afforded, good form and technique can be maintained, without limiting movement to a single joint and/or plane of motion (MuJo™ Mechanics Ltd., 2012). It is suggested that this multi-planar movement facilitates the training of movement patterns, and more of the muscle group is used and developed. Many people suffering from musculoskeletal impairment or injury undergo treatment or rehabilitation at healthcare facilities rather than hospitals. As such, the MuJo™ system could potentially offer a modality for treatment within a health-care setting; however, the effects of these machines have not been fully elucidated, thus their potential benefits or limitations currently remain unclear.

3.0 Literature Review

3.1 Introduction

This section aims to first provide a brief outline of the anatomy of the UQ, and then critically analyse the current literature regarding UQ exercise in reference to the limitations of present training modalities. The MuJo™ equipment is then presented as an alternative method of training; as such, the themes of this review are predominantly centred on those parameters that distinguish the MuJo™ equipment from traditional equipment. It is also argued that these aspects are integral to efficacious and efficient training. Finally, an exploration of the methods used within the study is included.

3.2 Shoulder anatomy and function

The shoulder complex consists of four articulations, namely glenohumeral, acromioclavicular, sternoclavicular, and scapulothoracic. In order to ensure fluid, coordinated movement, intact articulations and the preservation of power in their respective muscles is required (Inman et al. 1944). Whilst the shoulder has the most mobility of any joint in the body, the absence of bony restrictions results in its concomitantly being the least stable joint (Jordan *et al*., 2012). The stability is instead provided by soft tissues, and the integrity that is provided by its ligaments is reinforced by the two layers of musculature crossing the joint. Due to their angles of pull, the rotator cuff muscles provide vital glenohumeral stability, both from passive tension and dynamic contraction (Hay and Reid, 1988; Jordan *et al*., 2012). As such, a collective reduction in rotator cuff force or an isolated reduction from any of its muscles can increase humeral head translation, and hence risk of subluxation (Dark *et al*., 2007). Optimal coordination of the shoulder results in a compromise between maintaining stiffness and producing the required joint torque (Blache *et al*., 2015); as such, coordination due to muscle contractions plays a large part in stabilising the shoulder (Kronberg *et al*., 1990).

The stability of the shoulder is a result of interactions between static and dynamic restraints that precisely centre the humeral head within the glenoid fossa (Hurov, 2009). The static facets are provided by adhesive and cohesive forces inherent to synovial fluid, negative pressure within the joint capsule, and the presence of ligaments and tendons (Hurov, 2009). It appears that ligaments provide stability at extreme ranges of motion, and each ligament has its own role through a particular range of motion (Curl and Warren, 1996). The morphology of ligaments also provides a mechanical advantage via the crimping nature of their collagen fibres; their ability to absorb energy before fibres undergo full tension enables high energy storage capacity over a large range of joint positions (Bigliani et al., 1996). Dynamic constraint arises from active muscle, by way of concavity compression, muscle stiffness, and tendon compliance (Hurov, 2009). Concavity compression is predominantly provided by the rotator cuff musculature, facilitated by two important characteristics of their morphology: they have a large cross-sectional area and relatively short individual fibre lengths (Hurov, 2009; Veeger *et al*., 1991). Cross-sectional area and fibre length are determinants of potential force and excursion of the fibre respectively, hence the rotator cuff muscles are capable of generating high forces to compress the humeral head into the glenoid, whilst also constraining humeral head translation (Hurov, 2009). In addition, when the strain energy stored by tendons is released, it may be added to this muscle force, further facilitating the active stabilisation of the glenohumeral joint.

The first comprehensive insight into the function of the shoulder was provided by the pioneering work of Inman and colleagues (1944). The authors stated that a minimum of three forces are required to ensure glenohumeral equilibrium, derived from the weight of the extremity, the abducting musculature that is primarily represented by the deltoid, and the resultant of these forces acting through the centre of rotation in an opposing direction to that of the deltoid (Inman *et al.*, 1944). The authors also concluded that the use of EMG is considered a useful method of investigating the activity of individual muscles, as well as their role within a functional group performing coordinated movement (Inman *et al*., 1944). It is important to note that the shoulder operates as a kinetic chain, with muscles functioning in combination to result in the desired movements (Jordan *et al*., 2012; Schenkman and Rugo De Cartaya, 1987; Smith *et al*., 1996). As such, muscles possess functional roles and will be recruited correspondingly depending upon the specific task being performed. In brief, an agonist or prime mover muscle is one that provides the main force causing motion at a joint; conversely, an antagonist produces an opposite action to the desired movement (Jordan *et al*., 2012). Muscles may also act as synergists, and act in one of two ways; assisting or helping synergists act in pairs to produce the required motion, whilst undesired motions are cancelled out (Schenkman and Rugo De Cartaya, 1987). Alternatively, neutralisers are synergists that specifically cancel out undesired movements caused by the agonist. Additionally, muscles can provide a stabilising force, which prevent unwanted movements in directions other than the intended action (Schenkman and Rugo De Cartaya, 1987).

The trapezius comprises the outermost layer of the posterior muscles that act directly on the scapula; in particular, it elevates, retracts, and rotates the scapula (Jordan *et al*., 2012; Youdas *et al*., 2012). Together with the serratus anterior and levator scapulae, it provides passive support for the shoulder, facilitates active elevation, and is the upper component of the force couple required for scapular rotation (Inman *et al*., 1944; Schenkman and Rugo De Cartaya, 1987). Inman *et al*. (1944) additionally maintained that the upper fibres may also provide a postural function, as they exhibit an action potential whilst the arm is at rest. However, other authors have contradicted this claim, demonstrating that when participants were told to relax, the upper trapezius activity completely ceased (Bearn, 1961; Ballesteros *et al*., 1965). Therefore, whilst the trapezius may provide some postural function, it is not merely a postural muscle and has other important functions that must be considered (Basmajian and De Luca, 1985).

Of the aforementioned layers of shoulder musculature, the exterior layer consists of the deltoid and the pectoralis major (Hay and Reid, 1988; Jordan *et al*., 2012). The deltoid gives the shoulder its normal rounded contour, and is formed of three heads, each of which is activated differently for specific activities. It also forms a force couple with the rotator cuff by working simultaneously with the supraspinatus to result in shoulder flexion (de Witte *et al*., 2014; Jordan *et al*., 2012). However, each muscle is thought to contribute different amounts to glenohumeral equilibrium, in that a significant increase in deltoid activation has been observed as glenohumeral elevation moment increased (de Witte *et al*., 2014). Further, as the arm is increasingly abducted to 90°, the deltoid's moment arm improves, thereby allowing it to produce a larger force compared to the supraspinatus; however, this also results in increased shear force created by the deltoid (Jordan *et al*., 2012). The deltoid is also thought to have a dynamic stabilising role, particularly with the shoulder in 90° of abduction as the alignment of their fibres is then positioned to provide a transarticular compressive force (Boettcher *et al*., 2010).

The anterior head serves to flex, adduct, and internally rotate the humerus at the glenohumeral joint (Youdas et al., 2012). It is highly active during resisted flexion, and also assists in horizontal adduction (Hay and Reid, 1988); studies have also noted activity during scapation (Decker *et al*., 1999; Townsend *et al*., 1991; Youdas *et al*., 2012). The middle head's primary function is abduction, with its greatest activity occurring between 90° and 180° (Basmajian and De Luca, 1985; Hay and Reid, 1988; Inman *et al*., 1944; Jordan *et al*., 2012); however, all parts of the deltoid may be considered synergists for abduction (Smith *et al*., 1996). The posterior head acts as an extensor, and also assists the teres minor and infraspinatus in external rotation (Hay and Reid, 1988; Jordan *et al*., 2012). Additionally, regarding the deltoid's potential stabilising role, activity generated by the posterior deltoid during internal rotations may be indicative of its ability to counteract the rotational perturbation, hence maintaining joint integrity (Day *et al*., 2012). It has been argued that the three deltoid heads can be further divided into seven segments, classified as 'prime mover', 'synergist', or 'antagonist' segments based upon their lines of action relative to the movement direction (and therefore, specific task to be undertaken) and their periods of activation (Brown *et al*., 2007). For example, prime mover segments for a specific task, such as middle deltoid during shoulder abduction, had larger moment arms and more favourable mechanical lines of action for that task (Brown *et al*., 2007).

The pectoralis major consists of a clavicular and a sternocostal head, which converge at the sternoclavicular joint (Jordan *et al*., 2012). It is a prime mover for adduction and internal rotation of the humerus, but each head has secondary functions; the clavicular head flexes whilst the sternocostal head extends the humerus (Hay and Reid, 1988; Jordan *et al*., 2012). The muscle has a large adductor moment arm, with its muscle segments possessing effective mechanical lines of action for its tasks (Brown *et al*., 2007). The level of activity of the muscle has been contested, with Scheving and Pauly (1959) maintaining that internal rotation must be performed against resistance for it to be active, yet de Sousa *et al*. (1969) argued that the clavicular head is almost always active. Nevertheless, all authors stated that the sternocostal head is inactive except during adduction. This is in contrast to more recent arguments that it contributes to shoulder extension (Hay and Reid, 1988; Jordan *et al*., 2012), a discrepancy that may be due to increased understanding with further investigation of muscle activity. However, whilst the sternocostal head has been considered to be a shoulder extensor, Brown *et al*. (2007)

found that no segments of the pectoralis major contributed to shoulder extension, apart from a segment of the clavicular head that acted as an antagonist during extension. Furthermore, this section of the pectoralis major had a similar onset, peak intensity, and duration of activity to the extension prime mover segments, contrasting with the usual delayed onsets and duration of activity otherwise seen in antagonist sections (Brown *et al*., 2007). It was argued that the function of this activity may have been to protect the anterior shoulder from injury during forceful activation, made possible by the muscle's horizontal fibre orientation (Brown *et al*., 2007).

The infraspinatus functions as the primary external rotator of the humerus (Jordan *et al*., 2012), and together with teres minor and subscapularis, forms the inferior component of the force couple at the shoulder joint (Inman *et al*., 1944). As such, it acts to neutralise the compression force of the deltoids, and helps produce the external rotation necessary for full abduction (Schenkman and Rugo De Cartaya, 1987). Additionally, the infraspinatus may act as a depressor of the humerus during abduction by exerting an inferiorly directed force (Inman *et al*., 1944; Neumann, 2013). Due to its important stabilising role, an adequate level of strength in the infraspinatus is considered central to glenohumeral joint integrity, yet it is commonly involved in rotator cuff injuries (Hughes *et al*., 2014). The higher, transverse section of the infraspinatus may predominantly serve a supportive role, whereas the lower, oblique section is considered to be stronger and thus contribute to shoulder abduction (Nimura *et al*., 2015). Indeed, if isometric external rotation is performed against resistance, the oblique part of the infraspinatus is recruited more preferentially than the transverse section, regardless of shoulder position (Hughes *et al*., 2014). The latissimus dorsi acts to adduct and internally rotate the humerus (Jordan *et al*., 2012), and is a powerful extensor (Hay and Reid, 1988). When dividing the latissimus dorsi into 6 segments, all demonstrated both extension and adduction moment arms, and appeared to act as prime mover segments for adduction (Brown et al., 2007). Further, the latissimus dorsi acts as a synergist during internal rotation, and may have a stabilising role during flexion, with increasing activity as flexion angle increases (Kronberg *et al*., 1990).

The force-producing capability of a muscle is determined by many factors, including its architectural characteristics, as well as the length-tension and force-velocity relationships. The length-tension relationship states that greatest force can be produced when a fibre is at approximately its resting length, due to maximum actin and myosin filament interaction, resulting in optimal cross-bridge formation (Lorenz and Campello, 2012). At the single fibre level, this relationship is determined exclusively through cross-bridge interaction (Brughelli and Cronin, 2007). Therefore, as fibres lengthen or shorten, fewer cross-bridges can form leading to reductions in active tension. However, when considering this relationship for the entire muscle, both active and passive tension contribute; at shorter muscle lengths, all force is derived from cross-bridge formation, whereas at longer lengths, much of the force is due to passive components (Brughelli and Cronin, 2007; Lorenz and Campello, 2012). In shoulders with greater than normal ROM, loading the joint through extremes of motion could stretch some muscles excessively whilst others may not be lengthened enough; this could result in muscle activity at inefficient positions of their length-tension curves, leading to reduced force production (Weldon and Richardson, 2001). Further, it is also important to consider the joint-torque relationship; as joint angle changes throughout a movement, both the force produced and the length of the moment arm will change, thus affecting torque development (Brughelli and Cronin, 2007). Consequently, these relationships must be considered when attempting to analyse shoulder movement.

3.3 Upper Quadrant Training

Current resistance training guidelines state that programmes should train all major muscle groups, on two to three days a week; furthermore, programmes should include single- and multi-joint exercises, particularly by following multi-joint with a single-joint exercise for the same muscle group (ACSM, 2014). Single-joint exercises are purported to cause a higher degree of muscle damage, and hence hypertrophy, than do multi-joint exercises (Soares *et al*., 2015). Additionally, it has been suggested that, due to their lesser complexity, single-joint exercises may rely less upon neural factors and hence be easier to learn (Gentil *et al*., 2013). However, because of the isolation of the muscle involved, much less weight can be lifted with a single-joint exercise (Ingham, 2006). Nevertheless, single-joint exercises may be useful in correcting an imbalance between muscle groups that may pose an increased risk to injury (Gentil *et al*., 2015; Giannakopoulos *et al*., 2004). It is recommended that isolation exercises should be used to initially strengthen the weaker muscle, but then should be supplemented with more complex exercises to further enhance the surrounding musculature (Giannakopoulos *et al*., 2004).

Multi-joint exercises are recommended because they allow for multiple muscle groups to be recruited, and target both agonist and antagonist muscle groups effectively (ACSM, 2014; Gentil *et al*., 2015). Whilst it is commonly recommended to include both single- and multi-joint exercises in a programme, lack of time is often cited as a barrier to exercise (Gómez-López *et al*., 2011), thus including both types could result in a duration that may dissuade participation. As such, the current literature has attempted to determine whether a protocol including both is superior to that of only performing multi-joint alone (de Franca *et al*., 2015; Gentil *et al*., 2013; Gentil *et al*., 2015; Giannakopoulos *et al*., 2004). Generally, it is stipulated that both protocols are equally effective at increasing muscle size, strength, and motor unit activation (de Franca *et al*., 2015; Gentil *et al*., 2015). Consequently, although it has been stated that single-joint exercises are required in order to optimally stimulate all working muscles, multi-joint exercises alone may constitute a sufficient stimulus for adaptation. Therefore, the most time-efficient, and yet still effective, approach may be to focus on multi-joint exercises to promote hypertrophy and strength. Such an approach may improve and encourage long-term adherence to exercise.

A further benefit of multi-joint exercises arises from their incorporation of multiple degrees of freedom (DF); thus, the way in which muscles are trained and utilised in a task can be manipulated due to the multitude of kinematic patterns possible (Galloway and Koshland, 2002). As the number of muscles acting across a joint is generally in excess of the number of DF in that joint, a single joint torque can be determined by numerous muscle activation patterns (Gielen et al., 1998). Furthermore, various muscle activation patterns can result in the same desired behaviour, a phenomenon known as the motor-equivalence problem (Bernstein, 1967 cited in Gielen *et al*., 1998). This is somewhat emphasised by the fact that participants demonstrate the flexibility to use many muscles in various different combinations, depending upon the specific motor task and/or the instructions they receive (Tax *et al*., 1990a, 1990b; Theeuwen *et al*., 1994). For example, muscle activation patterns for the same muscle can differ for different types of contraction (e.g. isometric or concentric). This flexibility may indicate strategies to reduce the abundant DF in the motor system, and develop more stable and functional states of coordination (Davids *et al*., 2008).

When approaching a novel or less familiar task, it may be therefore beneficial to allow participants to explore with this flexibility, to find a solution and develop coordination strategies to increase movement economy and efficiency (Bernstein, 1967 cited in Gielen *et al*., 1998). However, the use of traditional resistance equipment will reduce the number of DF in multi-joint exercises, and therefore restrict the available muscle activation patterns and forces to be used in the movement execution and stabilisation (Cacchio *et al*., 2008). Comparisons between constrained and less constrained equipment have shown larger increases in one repetition maximum (1RM) performance, higher levels of muscle activity, and adaptive change in muscle activation pattern for the unconstrained equipment (Cacchio *et al*., 2008; Signorile *et al*., 2017), and a proximal-to-distal sequence of activation in the trunk and arm musculature (Koyama *et al*., 2010). Therefore, increasing the DF within resistance equipment may result in muscle activations and kinematics that better replicate the activation patterns observed for throwing and striking (Koyama *et al*., 2010). The proximal-to-distal sequence within throwing is considered necessary to generate increasingly large extension velocities, and to prevent a premature termination of action (Schenau, 1989). As adaptations arising from training are specific to the physiological and technical nature of the training stress, implementing exercises that involve the specific movements and techniques within the respective sport facilitates the development of relevant musculature and neuronal mechanisms (Gabbett *et al*., 2009; Muller *et al*., 2000; Young, 2006). Further, an unconstrained method of training allows for a more effective improvement of intermuscular coordination, due to the available development of motor strategy. (Cacchio *et al*., 2008). Less restricted training enables the body to explore various coordinative solutions in order to optimise techniques (Davids *et al*., 2008).

An additional variable that may affect the efficacy of training is the range of motion (ROM) over which weight is lifted. It has been speculated that training through an athlete's apparent optimal ROM can enable that athlete to lift loads heavier than those that can be lifted through a full ROM (Clark *et al*., 2008: 2011; Mookerjee and Ratamess, 1999). This is suggested to be due to the existence of a large deceleration phase within the full ROM repetition, resulting in a considerable part of the movement being performed way below maximum capability (Elliot *et al*., 1989; Lander *et al*., 1985; Newton *et al*., 1996). Conversely, performing movements over various ROM, with the consequence of targeting different joint angles, is purported to increase specificity (Clark et al., 2008, 2011); joint proprioceptors that deliver neural feedback ensure that strength gains are joint-angle specific (Ingham, 2006; Godfrey and Whyte, 2006). Accordingly, an intervention incorporating bench presses over variable ROM resulted in significantly higher isokinetic peak force values during the terminal phase of ROM, with the authors concluding an increased performance at shorter muscle lengths (Clark *et al*., 2011). Conversely, when comparing strength gains from resistance training over full, partial, or a combination of full and partial ROM repetitions, a statistically larger increase in 1RM bench press was observed in the full ROM group compared with the other two groups (Massey *et al*., 2005). This is perhaps due to the length-tension relationship; as joint angle changes throughout ROM, the number of crossbridges that can form will vary. At the ideal ROM, the muscle may be at its optimum length for force production due to maximum crossbridge formation (Pinto *et al*., 2012). This may not be possible at partial ROM, particularly if varying the range at which participants will train. Further, a greater increase in 1RM performance was observed when performing repetitions over a longer ROM compared with a shorter ROM, speculated to be due to greater morphological adaptations from a higher mechanical stress (McMahon *et al*., 2014). This may also be a result of the larger ROM involving a longer time under tension; responses to strength training are dependent upon both the intensity and duration of muscular tension (Crewther *et al*., 2005; Gentil *et al*., 2006). In addition, lifting over a full ROM requires the muscle to undergo tensile loading throughout a greater length, hence resulting in uniform loading (Ingham, 2006).

It is therefore argued that a more complete ROM ought to be advocated; even if absolute load may be slightly decreased, this will provide a greater internal stress and adaptation stimulus (McMahon *et al*., 2014). It must be noted however that strength gains may still be obtained from partial ROM repetitions and may be appropriate depending upon the end goal of the exercise. In this manner, partial repetitions may be advantageous when implemented as part of training to overcome an athlete's sticking point, or the aforementioned deceleration (Clark *et al*., 2011). If exercise is performed to failure, the sticking point can be defined as the point at which muscular failure occurs (Kompf and Arandjelovic, 2016). By increasing the possible force that can be produced against a desired load at this point, performance will improve, as strength at the specific ROM has been trained. As such, if strength at a particular joint angle is of importance, restricting the ROM may be more advantageous than if overall strength is to be improved.

Furthermore, restricting ROM to a safe range may be required when implementing a rehabilitation programme following injury, to avoid placing the structures under extraneous stress. Consequently, the choice of partial or full ROM repetitions ought to be made based on the individual desired outcome, and training ought to reflect the range of motion of the target activity (Godfrey and Whyte, 2006).

3.4 MuJo™

The MuJo™ System enables the integrated development of biarticular muscles and joints on one machine (MuJo™ Mechanics Ltd., 2012). The ability to load biarticular muscles may also be considered an improvement on traditional resistance equipment. Biarticular muscles are integral to well-coordinated movement, as the successful necessary transfer of joint rotation to translation of objects or the human body is a product of differing requirements. These arise from joint moments needed both to generate force and power, and also to determine the direction of the force that allows it to have application (Schenau et al., 1992). The systematic coactivation of monoarticular and biarticular muscles enables both these requirements to be met. Whilst monoarticular muscles appear to be concerned with the former role of force production, biarticular muscles are responsible for regulating the direction of the desired external force, by distributing the net moments across joints (Bolhuis et al., 1998). Therefore, by incorporating multi-joint movements that effectively train biarticular muscles, improvements in both force production and motor control may be observed.

The precise ROM over which the limb moves can be adjusted and predefined on the machines with the use of mechanical stops. This may be beneficial to target specific musculature, during rehabilitation from injury, for example. Performing controlled shoulder exercises over gradually increasing ROM is often recommended post-surgery or as part of a non-operative method of treatment (Escamilla *et al*., 2014; Manske, 2016; Neumann, 2013). Whilst the MuJo™ equipment is therefore novel and beneficial to sports and healthcare professionals, it is currently uncertain to what extent the mechanics of each exercise are affected by altering the ROM. Preliminary tests of the equipment demonstrate an increase in length-tension curves of biarticular muscles, postulated to be due to the addition of sarcomeres in series, known as sarcomerogenesis (Proske and Allen, 2005). This has the potential to enhance performance of the stretch-shortening cycle, due to a lower active stiffness at shorter muscle lengths and increased passive stiffness at longer muscle lengths (Brughelli and Cronin, 2007). Shifts in length-tension curves are thought to be a result of eccentric training (Malliaropoulos *et al*., 2012), with the potential to both increase athletic performance and reduce injury risk (LaStayo *et al*., 2003). However, there are a lack of randomised controlled training studies demonstrating such a shift (Brughelli and Cronin, 2007), and further eccentric exercise research is required to better understand how the neuromuscular system adapts to such training (Isner-Horobeti et al., 2013).

As each MuJo™ machine provides extensive exercising possibilities, it may be necessary to determine whether small alterations in ROM cause significant and quantifiable differences in activity of the shoulder musculature. Alongside this, it is essential to ascertain how an apparently healthy population responds to the exercises and equipment before they can be effectively and validly employed in a clinical environment. Therefore, a more comprehensive insight into the explicit outcomes of using these machines may be achieved by utilising EMG to observe the changes in activation of the specific muscles being used whilst exercising with the devices.

3.5 Electromyography

The ability to detect neuromuscular activity arises from the way in which muscle contraction is instigated. As skeletal muscle is constantly suffused in an ionic medium, there exists a voltage gradient across the fibre membrane; the inside of the fibre is around -90 mV compared with the outside (Kamen and Gabriel, 2010). Muscle tissue is therefore excitable, as even small differences in the resting membrane potential result in excitation of the sarcolemma. Once the membrane potential reaches a threshold value, a muscle fibre action potential (MFAP) is propagated along the sarcolemma, in both directions from the neuromuscular junction.

The motor unit comprises a single motor neuron and all the muscle fibres it innervates; as such, multiple muscle fibres are innervated by one motor neuron. The resulting individual MFAPs are summed both spatially and temporally, resulting in the motor unit action potential (MUAP) (Kamen and Gabriel, 2010). It is this MUAP that is recorded by the electrodes, and its amplitude and duration is determined by the individual MFAPs that make it up. The generation of action potentials is fundamentally an ionic process;

therefore, the conduction velocity is determined by the exchange rate of ions across the membrane, as well as differing histochemical and architectural characteristics (Kamen, 2014)

3.5.1. Detection

The EMG signal is inherently complex and is affected by a multitude of physiological aspects, as well as the characteristics of the equipment used to detect, record, and process it. The signal has both positive and negative components, and at each time point is the electrical sum of all active motor units (Kamen, 2014). Whilst both needle and fine wire electrodes can be used, their process is invasive, and thus surface electrodes are most commonly used in sports kinesiology environments. In order to improve replicability and comparison between studies, it is recommended that certain details ought to be systematically reported in each. These are: the type and material of electrodes, inter-electrode distance, location of electrodes and preparation of skin, type of filter used in the amplifier, the lower and upper cut-off frequency, input impedance, and the processing and normalisation methods (Clarys *et al*., 2010). Silver or silver chloride electrodes are most often used, and a bipolar configuration (consisting of two recording electrodes and one placed on an electrically neutral site) is most appropriate for dynamic activity (Kamen, 2014). Electrodes should be placed halfway between the most distal end plate zone and the distal tendon; attaching electrodes on the geometric belly of the contracted muscle, with the detection surface parallel to the length of the muscle, is considered a reliable method of placement (Clarys, 1983). Electromyographical signals should be sampled at a minimum of 1000Hz, and an analogue-to-digital converter with at least 12 bits should be used (Grimshaw *et al*., 2007).

The input impedance of the amplifier should ideally be at least ten times greater than the impedance of the electrodes (Clarys and Cabri, 1993). As it is not always possible to determine the electrode impedance, it is advised to have a relatively high amplifier input impedance, preferably greater than 100 Megaohms (Grimshaw *et al*., 2007). Additionally, the amplifier will possess a common-mode rejection ratio (CMRR), denoting the capability of the amplifier to attenuate signals common to both inputs (Basmajian and De Luca, 1985; Kamen, 2014); the CMRR should ideally be between 80 and 90 dB (Winter *et al*., 1980). Whilst EMG amplifiers are differential amplifiers, it is possible to use a double differential, involving three recording electrodes. The difference between electrodes one and two and between two and three is calculated, and the resulting two signals are input into a third amplifier and further differentiated (Grimshaw *et al*., 2007). This can help to attenuate the presence of crosstalk, the phenomenon whereby electrical activity from adjacent and/or distant muscles is recorded alongside the activity from the muscle of interest (Kamen and Gabriel, 2010). This occurs because muscle is a volume conductor and so electrical signals may travel indiscriminately throughout muscle; however, it is exacerbated by high levels of subcutaneous fat. The double-differentiating technique is the most consistent technique for reducing crosstalk (van Vugt and van Dijk, 2001).

To ensure an accurate detection, recording, and analysis of the EMG signal, it is important to maximise the fidelity of the signal. There are two aspects that can undermine this endeavour: the signal-to-noise ratio (the ratio of energy in the EMG signal to the energy in the noise signal), and the distortion of the signal (De Luca, 2002). Increasing the signal-to-noise ratio can be achieved by filtering the maximum amount of noise whilst also conserving as much of the EMG signal as possible (De Luca *et al*., 2010). Alongside several intrinsic and extrinsic sources of low-frequency noise that may cause issues, movement artefact noise is also of concern. This originates at the electrode-skin interface, occurring when the muscle moves under the skin, and also when a movement occurs at the electrode-skin interface, due to an impulse travelling through the muscle and skin underlying the sensor (De Luca *et al*., 2010). Movement artefact can also be caused by movement of the cable connecting the electrode to the amplifier. Additionally, noise is more likely during dynamic contractions, and hence may lead to false conclusions, particularly if the signal is used to discern information on the physiological and anatomical properties of muscles (De Luca *et al*., 2010). Determining the band-pass frequencies is important, as the aim is to both reduce noise and yet retain as much information required from the signal as possible; the power of the EMG signal lies mostly within the 5-500 Hz range (Merletti, 1999). The low-pass filter frequency should be set to where the noise amplitude exceeds that of the EMG signal; this is usually around 400-450 Hz (De Luca *et al*., 2010). Determination of the high-pass cut-off frequency however, is dependent upon the application and the muscles being investigated. Whilst for more vigorous activities it is recommended to have a higher cut-off frequency, 20 Hz provides a good compromise for ensuring acquisition of the maximum amount of desired information from the EMG signal (De Luca *et al.*, 2010). Furthermore, in order to preserve the EMG signal and minimise distortion, the signal should be processed linearly, and should not be clipped, i.e. there should be no unnecessary filtering (De Luca, 2002).

3.5.2 Processing

The EMG signal is often processed in one of two ways, either in the time-domain or frequency-domain, the choice of which will depend upon the aim of the research. Temporal processing is mostly used when the objective is regarding motor-coordination, quantifying an amount of activity, or how activity changes over time; frequency-domain processing, however, is more appropriate for the study of muscle fatigue (Clarys and Cabri, 1993). Because the signal is an alternating current signal, and thus its mean would be zero, it must first be rectified (Kamen, 2014). This ideally involves full-wave rectification, by reversing the sign of the negative voltages, thus preserving the energy of the signal (Basmajian and De Luca, 1985) and producing the absolute value of the EMG (Winter *et al.*, 1980). Following this, the average rectified value (ARV) can be calculated, by integrating the full-wave rectified signal over a time period or window, and then dividing the integrated EMG by that time window (Burden, 2008; Clarys and Cabri, 1993).

Equation 3.1. Average rectified value EMG.

$$ARV = \frac{1}{T}\sum_{t=1}^{T}|X(t)|dt$$

Where *X(t)* is the EMG signal, and *T* is the time over which the ARV is calculated.

Alternatively, the root mean square (RMS) can be calculated; this is the square root of the average power of the raw EMG calculated over a specific time period (Burden, 2008). Because this involves squared values of the original EMG signal, it does not require full wave rectification (Kamen, 2014).

Equation 3.2. Root mean square EMG.

$$RMS = \sqrt{\frac{1}{T}\int_{0}^{T} X^2(t)dt}$$

Whilst both methods are commonly used within the literature and acknowledged as acceptable, they often only provide information on a specific time period. This may not be representative of the full waveform, and using single maximal measures is not infallible against movement artefacts (Clarys and Cabri, 1993). It is possible to instead process by making successive calculations of the ARV or RMS throughout the duration of the signal. This produces a type of moving average, with the smoothness of the resulting curve depending upon the window width used; a longer window will result in a smoother curve and vice versa (Basmajian and De Luca, 1985; Burden, 2008; Winter et al., 1980). Consequently, the window width to be used will depend upon the contraction type and the intended application.

3.5.3 Normalisation

If the EMG signal has been processed in the time domain, it is then necessary to normalise it; this rescales the signal from millivolts to a percentage of a reference value (Ball and Scurr, 2008). Normalisation is necessary due to the inherent variability of the EMG signal, as the signal is affected by a multitude of technical, anatomical, and physiological facets (Burden, 2010). These factors can be categorised into three groups: causative, which are factors that have a basic effect on the signal and may be further divided into extrinsic or intrinsic aspects; intermediate, those that are indicative of physiological phenomena influenced by the causative and which have an effect on the deterministic factors; and deterministic, which have a direct influence on the detected signal (De Luca, 1997). These parameters are listed in Table 3.1.

Table 3.1. Factors that may influence the recorded EMG signal (De Luca, 1997; De Luca *et al*., 2010).

Causative		**Intermediate**	**Deterministic**
Extrinsic	Intrinsic		
• Electrode configuration • Electrode location with respect to the myotendonous junction, & lateral edge of the muscle. • Electrode orientation with respect to the muscle fibres • Power-line noise • Cable motion artefact	• Number of active motor units • Muscle fibre type composition • Muscle blood flow • Fibre diameter • Depth & location of active fibres with respect to electrode detection surfaces • Amount of tissue between muscle surface and electrodes • Firing characteristics of motor units • Motor unit twitch.	• Band pass filtering aspects of the electrodes • Detection volume of electrodes • Superposition of action potentials • Spatial filtering effect due to relative position of electrode and active muscle fibres.	• Number of active motor units • Motor unit force-twitch • Mechanical interaction between muscle fibres • Motor unit firing rate • Number of detected motor units • Amplitude duration and shape of motor unit action potentials • Recruitment stability of motor units.

As a result of these factors, if the EMG signal is not normalised, misinterpretation and the drawing of false conclusions are likely. As such, normalisation reduces variability, and allows for comparisons of amplitude information from multiples muscles, thereby improving absolute EMG reliability (Ball and Scurr, 2008; Kamen and Gabriel, 2010).

Both SENIAM (Surface EMG for Non-Invasive Assessment of Muscles) and the Journal of Electromyography and Kinesiology recommend the use of a maximal voluntary contraction (MVC) as a reference contraction with which to normalise (Hermens et al., 1999). Often this is obtained from a static contraction, which is appropriate in all static investigations; in dynamic activities, however, it is problematic. One limitation of this method is that different maximal values can be found at different angles of movement; further, values obtained from dynamic contractions can often exceed 100% of MVC, thus questioning linearity (Clarys et al., 2010).

Alternative methods of normalisation have therefore been suggested and developed for use in sport and/or kinesiological investigation (Ball and Scurr, 2010; Burden, 2010; Clarys et al., 2010). These include highest peak activity in dynamic conditions ($peak_{task}$), mean activity in dynamic conditions ($mean_{task}$), and peak EMG from a submaximal non-isometric voluntary contraction (dynamic-sub_{mvc}) (Burden, 2010). Dynamic-sub_{mvc} is most appropriate only if all of the muscles under investigation will be activated to the same maximal level during the task. $Peak_{task}$ and $mean_{task}$ are most effectual improving group homogeneity, by removing true biological variation (Burden, 2010). However, it must be noted that all normalisation methods achieve this, albeit to a lesser extent.

3.6 Conclusion

This section of the thesis aimed to briefly discuss the anatomy of the UQ, alongside aspects of current training that may be considered problematic, and the methods to be used within the present study. It was concluded that multi-joint exercises incorporating multiple DF may be the most effective, yet many traditional resistance machines restrict the DF available; consequently, the potential to develop motor control may be compromised. In response to these issues, the MuJo™ equipment was presented as an alternative method of training. Due to the unique multi-axis cam technology and independently moving parts, the MuJo™ machines allow multiple joints and muscles to be trained on one machine, over a specified ROM. However, it was noted that little is known about how altering the ROM affects mechanics whilst using the device, and that further information is required to determine how an apparently asymptomatic population responds to the equipment.

3.7 Aims

The aim of this thesis was to investigate the effect of altering the ROM on the activation of the musculature of the UQ, whilst exercising on the MuJo™ External Shoulder Machine and MuJo™ Internal Shoulder Machine.

3.8 Objectives

This was determined by:

- Recruiting participants to perform repetitions of external rotation and abduction, and internal rotation and adduction, at twelve different ranges of motion on the MuJo™ External Shoulder Machine and MuJo™ Internal Shoulder Machine respectively.
- Using surface EMG to quantify the changes in muscle activation occurring as a result of the alterations in ROM.

3.9 Hypotheses

It was hypothesised that:

1. those repetitions with less ROM would result in the agonist or prime mover muscles being predominantly activated;
2. those repetitions with more ROM would demonstrate higher levels of activation in supporting musculature such as synergists and antagonists;
3. those repetitions with greater ROM would consist of a longer active phase duration than those that were more restricted.

4.0 Methods

4.1 Participants:

Following an *a priori* power analysis (effect size *f*=0.40; α=0.05; β=0.80), nine male participants (mean ± SD: age: 25 ± 5 years; body mass: 77.06 ± 11.06 kg; height: 1.76 ± 0.09 m) were recruited via word of mouth. Participants were recreationally active and resistance trained at least twice a week; five of the nine participants were completing a strength and conditioning Master's programme, which required them to perform Olympic lifts on a regular basis. All took part in a variety of sports, including basketball, cricket, and athletics. Participants were free from injury, and specifically had no incidence of injury to the shoulders or UQ over the past year. Suitability for participation was confirmed during a familiarisation session, in which instruction on correct technique was provided for all procedures involved. Furthermore, participants were fully informed of all protocols and potential risks, via written and verbal explanation, and subsequently provided their written consent. Approval from the Coventry University Ethics Committee was obtained prior to any procedures.

4.2 Procedures:

Participants were required for two sessions: a familiarisation session, during which their 1RM was also determined for both the External and Internal Machines, and a main testing session. For both trials, participants reported to the Strength and Conditioning suite fully hydrated, having abstained from caffeine consumption so as to avoid artificially enhancing performance (Richardson and Clarke, 2016). They were also instructed to abstain from vigorous UQ exercise for at least 24 hours prior to testing. All tests were performed at the same time of day to minimise circadian variation of the measured variables (Clarke et al., 2011).

Upon arrival for the 1RM session, the configuration specific to the participant (regarding seat height and handle length) was determined on each of the machines and noted for future trials. Correct technique was demonstrated for each of the machines, and verbal instruction was given where required. Participants were then given time to practice with low loads to enable full familiarisation, ensure good technique, and minimise the risk of injury. Preparations were then made for EMG data collection as outlined below, and participants performed an UQ warm-up consisting of low loaded repetitions through full ROM on each machine. Following this, the 1RM through full ROM was performed according to a standard protocol (NSCA, 2016). In brief, the load was incrementally increased for each of the machines, until participants could no longer perform one repetition. The 1RM was required in order to provide a reference contraction with which to normalise the EMG signals from the main testing sessions. It provided muscle activation from all muscles involved, from one dynamic movement; thus, it was a more appropriate normalisation method than a static maximal voluntary contraction (Burden, 2010; Clarys et al., 2010).

The main testing session took place at least 24 hours after the 1RM session; participants were firstly prepared for collection of EMG data and then completed a warm-up, consisting of submaximal trials at 30% of their 1RM through full ROM, on each machine. For the main protocol, three angles of abduction/adduction and four angles of rotation were selected, to incorporate a spectrum of the angles that are available throughout the full ROM that each machine provides. For example, for the Internal Machine, the angles of adduction were 15°, 45°, 75°, and angles of rotation were -30°, 0°, 30°, 60°. These were combined to give twelve settings on each machine (e.g., 15°/-30°), and three repetitions at 50% of 1RM (mean ± SD: External Machine: 57 ± 16 kg; Internal Machine 46 ± 9 kg) were performed at each setting, resulting in twelve sets of three repetitions. All settings were randomised and counterbalanced, so as to minimise any learning or order effect. Participants were instructed to keep their heads against the head rest at all times, and perform the movements as smoothly as possible, with a slight pause between each repetition. Additionally, the protocol was filmed via a digital video camera (Sony video HDR-HC9, Sony, Tokyo, Japan), positioned perpendicular to the plane of motion on the participants' dominant side. This was synchronised with the EMG data, to provide a visual and qualitative representation of the movements.

Some materials have been removed due to 3rd party copyright. The unabridged version can be viewed in Lancester Library - Coventry University.

a) **b)**

Some materials have been removed due to 3rd party copyright. The unabridged version can be viewed in Lancester Library - Coventry University.

c) **d)**

Figure 4.1. Demonstration of the movement pathway of the UQ whilst using the MuJo™ devices: a) participant sat in neutral; b) participant abducted to almost 90°; c) midway through external rotation; d) end of abduction and external rotation. Movement would then be reversed starting from the position in d).

4.3 Collection of EMG:

Muscle activation of upper trapezius, anterior deltoid, posterior deltoid, infraspinatus, pectoralis major, and latissimus dorsi were measured using surface EMG. These muscles were chosen as they are recruited at various intensities for the movements involved and are more accessible with surface electrodes than the deeper muscles of the rotator cuff, which would require indwelling electrodes (Waite *et al*., 2010). Additionally, the machines are purported to synergistically recruit many of the larger muscles alongside those from the rotator cuff, thus it was deemed appropriate to measure myoelectric activity from some of these larger muscles (MuJo™ Mechanics Ltd., 2014).

Prior to placement of electrodes, the skin was shaved to reduce skin impedance, and cleaned with alcohol wipes. Passive bipolar silver/silver chloride surface electrodes (Blue Sensor Ltd., Denmark), 30mm in diameter, were then adhered to the contracted belly of each muscle in line with the muscle fibre direction. Precise location of electrodes was determined in accordance with recommendations of SENIAM (Hermens et al., 1997) and Cram et al. (1998), and inter-electrode distance was 1.5cm. Electromyography data was recorded via an ME6000 system (Mega Electronics Ltd., Finland), with an input impedance of less than 10^{15}/0.2 ohm/pF, a common mode rejection ratio at 60 Hz of greater than 110dB, a noise level of 1.2 mV, a gain of 10 + 2% and a bandwidth range from 0 Hz – 500 Hz. Activity was sampled at 1000 Hz, via a 16-bit analogue-to-digital converter and stored on a laptop computer with MegaWin software (MegaWin PC-SW 700046 version 3.0, Mega Electronics Ltd., Finland).

4.4 Processing

Both the 1RM trials and the main session trials were processed using the same methods. Raw EMG data were bandpass filtered within the MegaWin software, with cut-off frequencies of 5 Hz and 500 Hz (Merletti, 1999), then exported to Microsoft Excel where all further analyses were conducted. Each signal was processed using the RMS, with a sliding window of width 100 ms, throughout its duration (Basmajian and De Luca, 1985; Burden, 2010). All muscles were processed for the Internal Machine; the pectoralis major was disregarded from the External Machine as the data were deemed too noisy (see appendix 2). As demonstrated in appendix 2, the recorded signal contained higher frequency components than those expected for an accurately collected EMG signal; these components are therefore considered to be noise artefacts. The presence of these artefacts makes it difficult to accurately identify thresholds and key features; this can be observed in the Figure 2 in appendix 2, and consequently, it was decided that this data could not be used accurately.

For the main session trials, the onset and offset of the active phases was determined for each muscle according to Burden (2010). In brief, the baseline EMG was treated as a stochastic variable, and the mean of this baseline was calculated over a 50 ms window. The muscle was considered active or inactive when the amplitude of the EMG rose above or dropped below two standard deviations of the mean of the baseline for 25 ms or more

(Burden, 2010; Hodges and Bui, 1996). It was necessary that both criteria were met but, importantly, each trial was verified by visual inspection as this remains the gold standard (Allison, 2003; Di Fabio, 1987; Hodges and Bui, 1996). Once the onset and offset were determined, the active phase was calculated and expressed in seconds; the non-active phase, defined as the time between two successive active phases, was also noted. This was performed individually for every muscle from each of the three repetitions in each of the twelve sets, for each machine. The peak EMG values occurring during each active phase were extracted, and averaged over the three active phases for each muscle, to provide separate values for each setting on the machines. These values were then normalised to the peak EMG calculated from the 1RM contractions ($peak_{task}$) (Ball and Scurr, 2010; Burden, 2010; Clarys et al., 2010). Muscle activity was then divided into categories of percentages, according to Kelly et al. (2002); those values below 35% of 1RM denoted low activity, between 35-70% indicated moderate activity, and above 70% indicated high activity.

4.5 Statistical Analysis

Data were analysed with IBM SPSS Statistics for Windows (Version 22.0, Armonk, NY: IBM Corp.). The effect of abduction/adduction angle and rotation angle on normalised peak EMG was analysed using a two-way analysis of variance with repeated measures for each muscle. Sphericity was assessed with the Mauchly test of Sphericity; if violated, and the level of violation was <0.75, the Greenhouse-Geisser correction was used, and if >0.75 then Huynh-Feldt correction was used (Atkinson, 2001). Where significant differences were found, pairwise comparisons with Bonferroni correction for multiple comparisons were used to identify where they occurred. Further, effect sizes were estimated using partial eta squared (η^2_p), and interpreted according to Cohen (1992): 0.10-0.24 (small), 0.25-0.39 (medium), ≥40 (large). Finally, within-subject coefficients of variation (CVs) (Atkinson and Nevill, 1998) were calculated for the peak EMG data, to demonstrate intra-individual variability, and are presented in tables 5.7 and 5.8.

5.0 Results

5.1 External Machine

A significant main effect for abduction with a large effect size in the upper trapezius was found ($F_{(1.1,\ 8.7)}$=15.34, P=0.003, η^2_p=0.66); post-hoc analysis demonstrated that EMG amplitude was significantly higher at 90° of abduction than 60° and 30°, and significantly higher at 60° than at 30° (95% CI: 90°/60° [3.97, 26.97]; 60°/30° [1.37, 22.84]); 30°/90° [-48.41, -6.77]). A significant main effect for abduction was also observed in the anterior deltoid ($F_{(2,16)}$=7.17, P=0.006), alongside a large effect size (η^2_p=0.47); post-hoc testing demonstrated muscle activation was significantly higher at 90° than at 60° (95% CI [0.29, 14.61]). A significant main effect for rotation in the latissimus dorsi was found ($F_{(3,24)}$=7.96, P=0.001), with a large effect size (η^2_p=0.50); post-hoc tests revealed EMG amplitude at 0° to be significantly lower than both 90° (95% CI [0.04, 26.80]) and 60° (95% CI [0.79,16.98]). No significant interactions or main effects were observed in the remaining muscles.

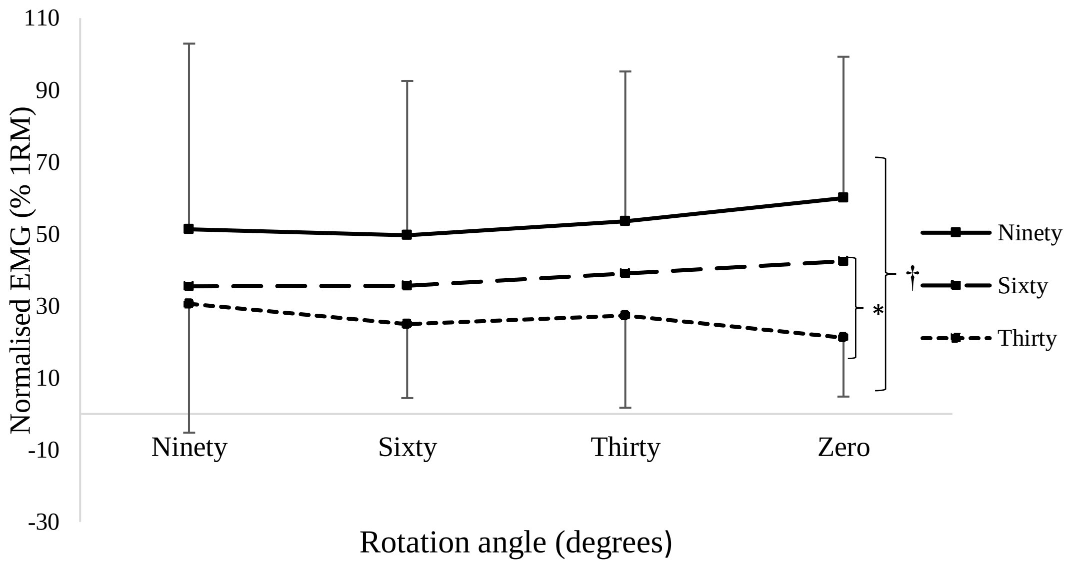

Figure 5.1: Mean ± SD of normalised peak EMG for upper trapezius across the twelve settings on the MuJo™ External Machine. Legend denotes abduction angle (degrees). *= EMG activity was significantly higher at 60° than at 30° of abduction (*P*<0.05). † = EMG activity was significantly higher at 90° than 60° and 30° of abduction (*P*<0.05).

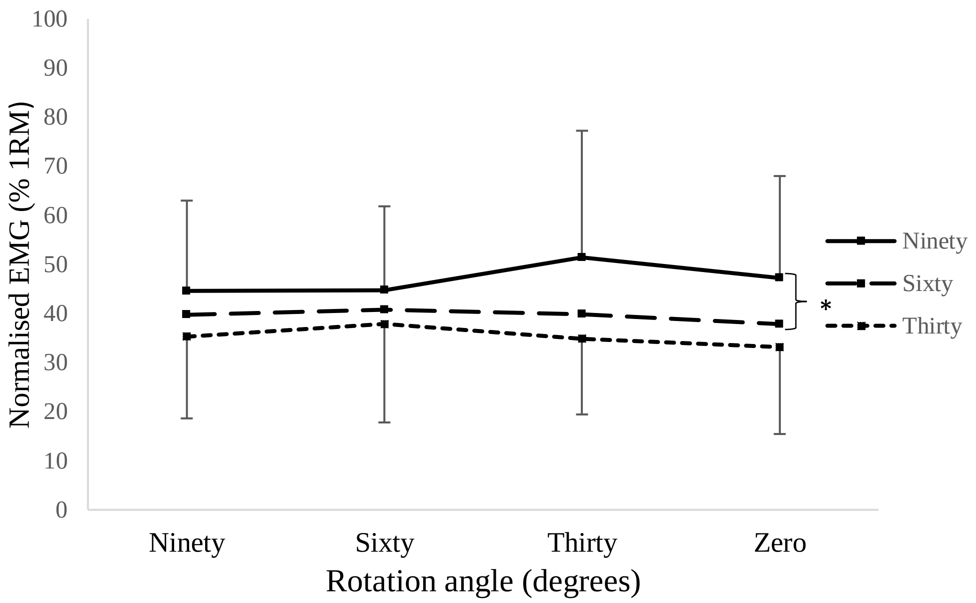

Figure 5.2: Mean ± SD of normalised peak EMG for anterior deltoid across the twelve settings on the MuJo™ External Machine. Legend denotes abduction angle (degrees). *=EMG activity was significantly higher at 90° than 60°of abduction (P<0.05).

For the upper trapezius (Figure 5.1), EMG amplitude was greatest for the abduction with the most ROM (i.e. 90°) ($F_{(1.1,\ 8.7)}$=15.34, P=0.003); all activity at both 90° and 60° of abduction was moderate, yet low at 30°. Whilst there were no significant differences between rotation angles ($F_{(3,\ 24)}$=1.32, P=0.29), for the repetitions at 90° and 60° of abduction, activation tended to increase as rotational ROM decreased, yet the opposite trend was seen in 30° of abduction. Within the anterior deltoid (Figure 5.2), EMG amplitude was highest for 90° of abduction ($F_{(2,\ 16)}$=7.17, P=0.006), yet remained fairly stable through all angles of rotation ($F_{(3,\ 24)}$=0.43, P=0.74). It was categorised as moderate through all repetitions at 90° and 60° of abduction; for 30° of abduction, it was moderate at 90° and 60° of rotation, but low at 30° and 0° (Table A1).

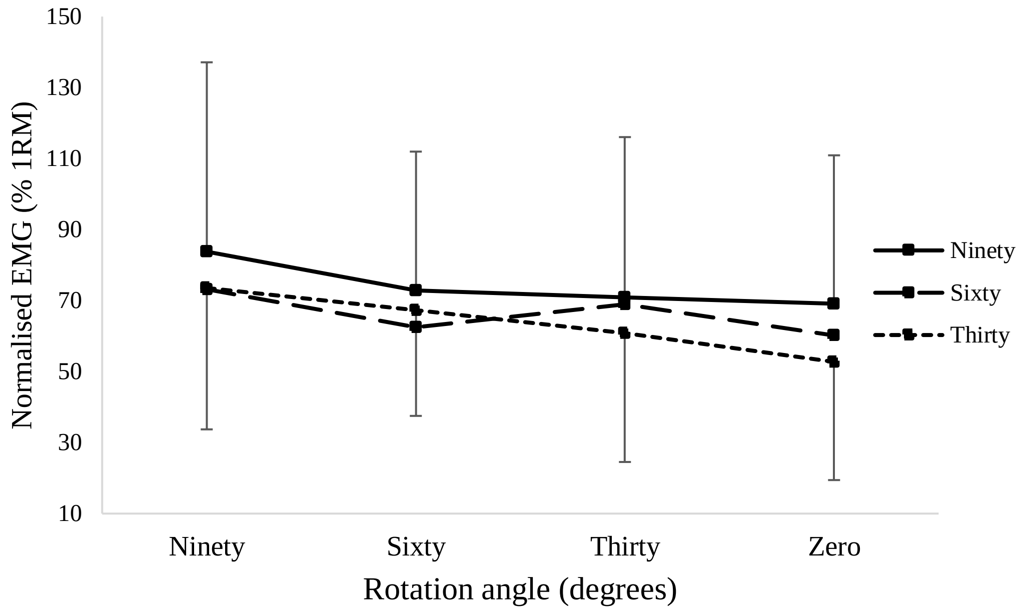

Figure 5.3: Mean ± SD of normalised peak EMG for posterior deltoid across the twelve settings on the MuJo™ External Machine. Legend denotes abduction angle (degrees).

For the posterior deltoid (Figure 5.3), again muscle activation was highest for 90° of abduction ($F_{(2, 16)}$=1.38, P=0.28), and also the greatest for the most rotational ROM (i.e. 90°); activity was high at 90° of rotation at all angles of abduction. Amplitude declined as rotation became more constrained, with the one exception of a slight increase at 60°/30° ($F_{(1.55, 12.39)}$=1.71, P=0.19); activity was categorised as moderate for 60°, 30°, and 0° of rotation for all angles of abduction, except for 30° and 0° at 90° of abduction (Table 5.1). For the infraspinatus (Figure 5.4), the highest level of muscle activity occurred during 90° of abduction ($F_{(2, 16)}$=2.55, P=0.11), yet generally during the least amount of rotation (0°), with the exception of the increase at 90°/30° ($F_{(3, 24)}$=1.53, P=0.23). Activity was mostly high throughout all repetitions, with the exceptions of moderate activity at 90° and 60° of rotation at 60° of abduction, and 60° and 30° of rotation at 30° of abduction (Table A1).

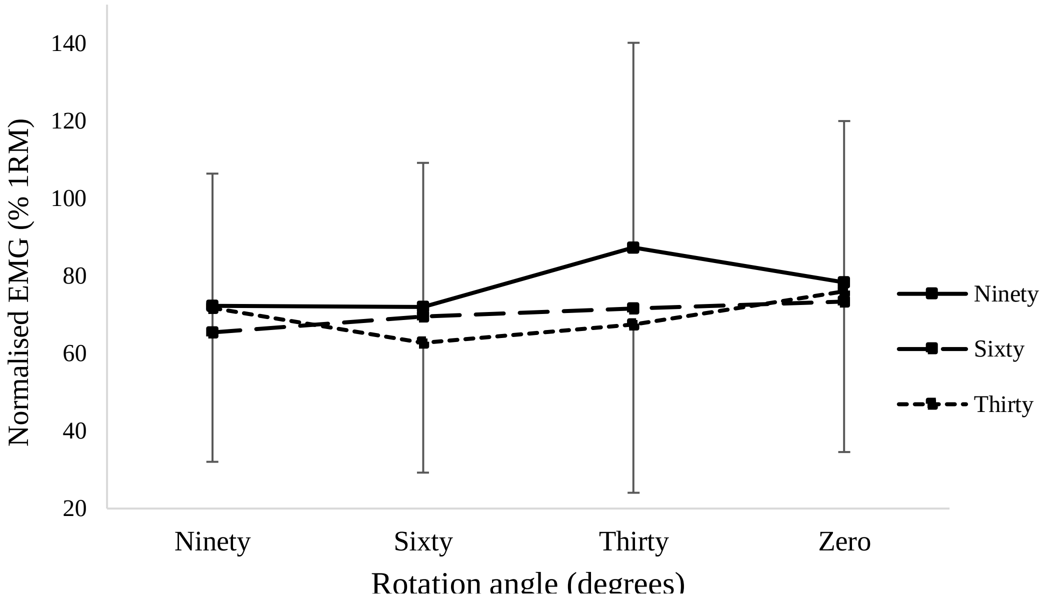

Figure 5.4: Mean ± SD of normalised peak EMG for infraspinatus across the twelve settings on the MuJo™ External Machine. Legend denotes abduction angle (degrees).

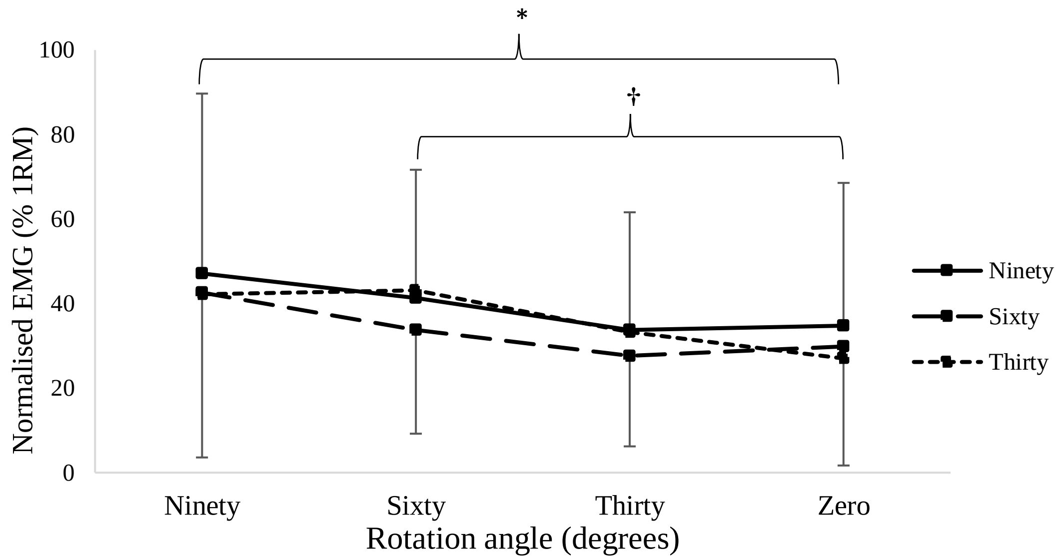

Figure 5.5: Mean ± SD of normalised peak EMG for latissimus dorsi across the twelve settings on the MuJo™ External Machine. Legend denotes abduction angle (degrees).
***= EMG significantly lower at 0° than at 90° of rotation (*P*<0.05).**
† = EMG activity significantly lower at 0° than at 60° of rotation (*P*<0.05).

Within the latissimus dorsi, EMG amplitude was higher at 90° of abduction than 60° ($F_{(2, 16)}$=1.01, P=0.39), and predominantly greater than at 30° with the exception of those repetitions at 60° of rotation (Figure 5.5). Additionally, for 90° and 60° of abduction, the highest peak EMG occurred with 90° of rotation, with the lowest activation at 30° of rotation; yet for 30° of abduction, the highest peak EMG occurred at 60° of rotation and the lowest at 0° of rotation ($F_{(3, 24)}$=7.96, P=0.001). Activity was low at 30° and 0° of rotation for all angles of abduction, and predominantly moderate at 90° and 60° of rotation at all angles of abduction, except for low activity at 60°/60°.

For all muscles, the longest active phase duration occurred during the movements at 90°/90° (representing the least constrained movement), with the shortest duration occurring at 30°/0° (the most constrained movement), and a mean difference of 1.57 s. Generally, the duration of active phase decreased as ROM decreased, with the shortest times mostly occurring for 30° of abduction (table 5.1). However, for each muscle, the duration of the active phase at 30°/30° was longer than that for 30°/60°, even though it comprises 30° less rotation. For the non-active phases, the longest duration occurred at 60°/60° for trapezius (2.25 s), anterior deltoid (2.12 s), and posterior deltoid (2.37 s). However, the longest duration for the infraspinatus was noted at 90°/90° (2.82 s), and for latissimus dorsi at 30°/30° (2.35 s) (Table 5.2).

5.2 Internal Machine

A significant main effect for rotation in the pectoralis major was found ($F_{(3, 24)}$=6.98, P=0.002), with a large effect size (η^2_p=0.47). Post-hoc tests demonstrated that EMG amplitude was significantly lower at 60° than 30° (95% CI [0.87, 21.35]). No significant interactions or main effects were observed in the remaining muscles.

For the upper trapezius (Figure 5.6), EMG amplitude was highest at 75° of adduction ($F_{(1.22, 9.76)}$=1.69, P=0.23), as activity was high for all angles of rotation; amplitude also tended to increase as rotational ROM decreased, with the exception of 15° of adduction, in which activation decreased as ROM decreased ($F_{(1.32, 10.57)}$=0.44, P=0.58). The highest peak EMG value in the trapezius occurred during the most constrained repetitions (i.e. 75°/60°). Within the anterior deltoid (Figure 5.7), EMG amplitude was mostly highest during repetitions at 45° of adduction, and lowest at 75° of adduction ($F_{(1.20,\ 9.64)}$=3.69, P=0.08); the one exception was those repetitions at 15°/60°, which were higher than those at 45°/60°. Activity was categorised as high for all angles of rotation at both 15° and 45° of adduction; it was moderate at 75° of adduction, for all rotation angles (Table A2). Peak EMG was also fairly high at 60° of rotation for all angles of adduction; however, the highest value occurred for 45° of adduction occurred at 0° of rotation, with similar values observed for -30° and 60° of rotation at 75° of adduction ($F_{(1.55,\ 12.41)}$=2.15, P=0.16) (figure 5.7).

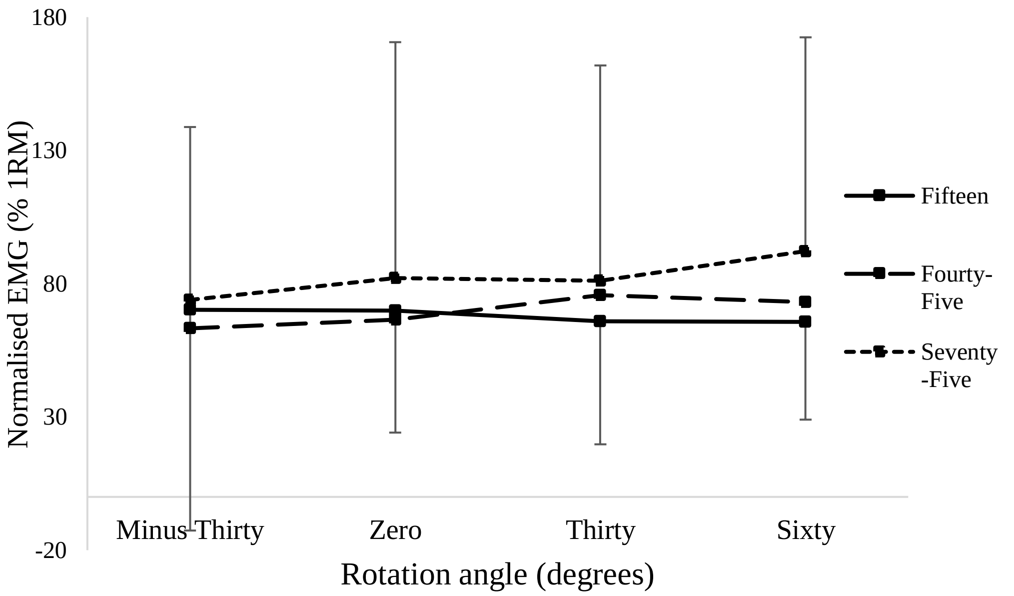

Figure 5.6: Mean ± SD of normalised peak EMG for upper trapezius across the twelve settings on the MuJo™ Internal Machine. Legend denotes adduction angle (degrees).

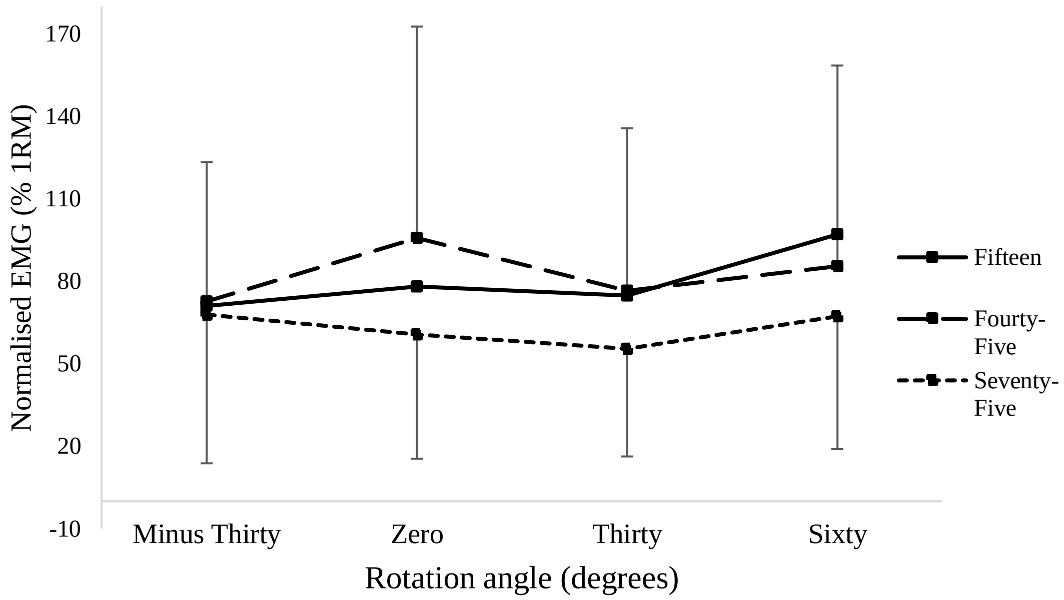

Figure 5.7: Mean ± SD of normalised peak EMG for anterior deltoid across the twelve settings on the MuJo™ Internal Machine. Legend denotes adduction angle (degrees).

For the posterior deltoid (Figure 5.8), EMG amplitude was predominantly lowest at 15° of adduction, and initially highest at 75° until surpassed by 45° as rotational ROM became more restricted ($F_{(2,\ 16)}$=0.57, P=0.58). Activity was mostly high throughout, with the exceptions of moderate activity at -30° of rotation at 15° and 45° of adduction, and also at 30° of rotation at 15° and 75° of adduction (Table A2). No significant differences were observed for rotation ($F_{(1.29,\ 10.29)}$=0.19, P=0.74); activation was fairly consistent throughout the rotational ROM for 15° of adduction, and increased steadily for 45° of adduction. However, at 75° of adduction it mostly decreased as rotational ROM becomes more constrained. The highest peak EMG values in the posterior deltoid occurred during the repetitions with the least adduction ROM and greatest rotational ROM (i.e. 75°/-30°). For the infraspinatus, EMG amplitude was mostly highest at 15° of adduction, with the exception of the repetitions at 45°/-30° and 75°/0° ($F_{(2,\ 16)}$=0.12, P=0.89) (Figure 5.9). Additionally, aside from the aforementioned repetitions at 75°/0°, there was a general trend for activation to decrease as rotational ROM decreased, for all angles of adduction ($F_{(3,\ 24)}$=1.07, P=0.38). Activity was classed as moderate throughout all repetitions in the infraspinatus (Table A2).

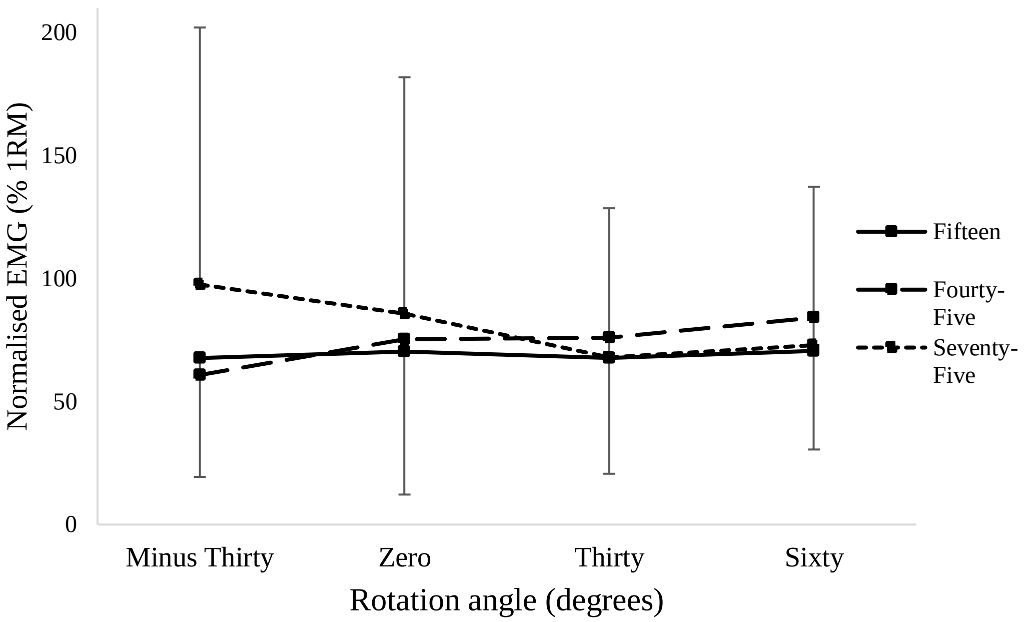

Figure 5.8: Mean ± SD of normalised peak EMG for posterior deltoid across the twelve settings on the MuJo™ Internal Machine. Legend denotes adduction angle (degrees).

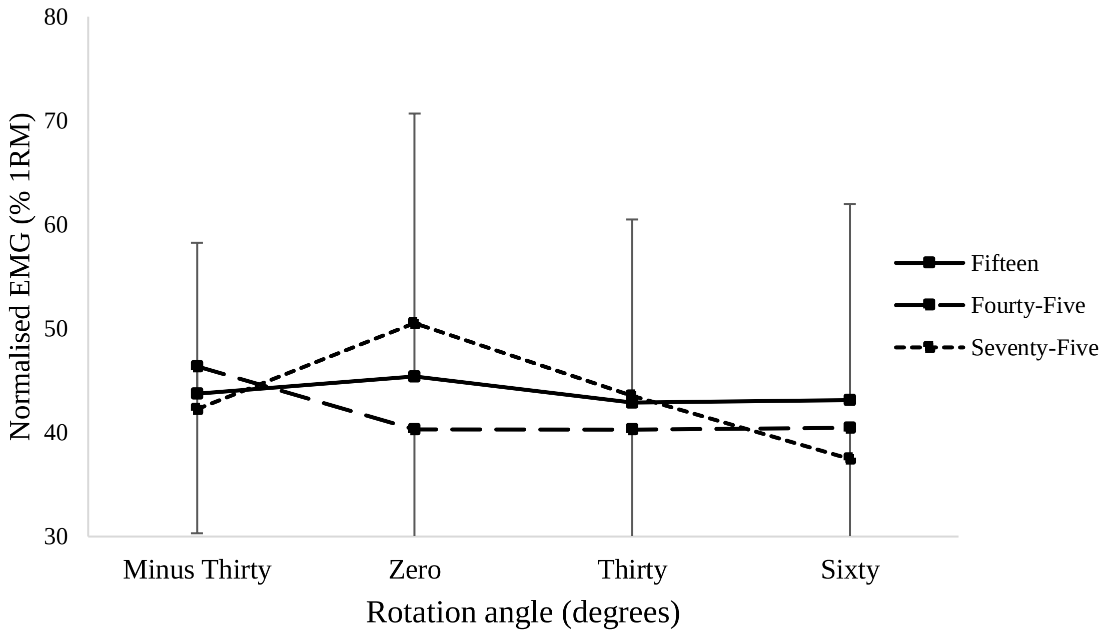

Figure 5.9: Mean ± SD of normalised peak EMG for infraspinatus across the twelve settings on the MuJo™ Internal Machine. Legend denotes adduction angle (degrees).

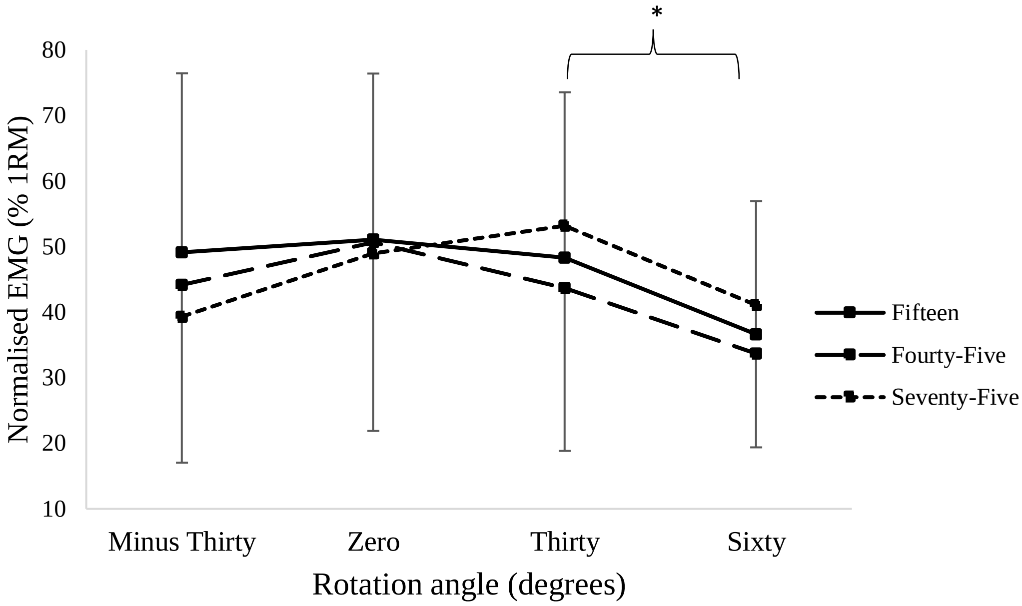

Figure 5.10: Mean ± SD of normalised peak EMG for pectoralis major across the twelve settings on the MuJo™ Internal Machine. Legend denotes adduction angle (degrees). *= EMG significantly lower at 60° than at 30° of rotation (*P*<0.05).

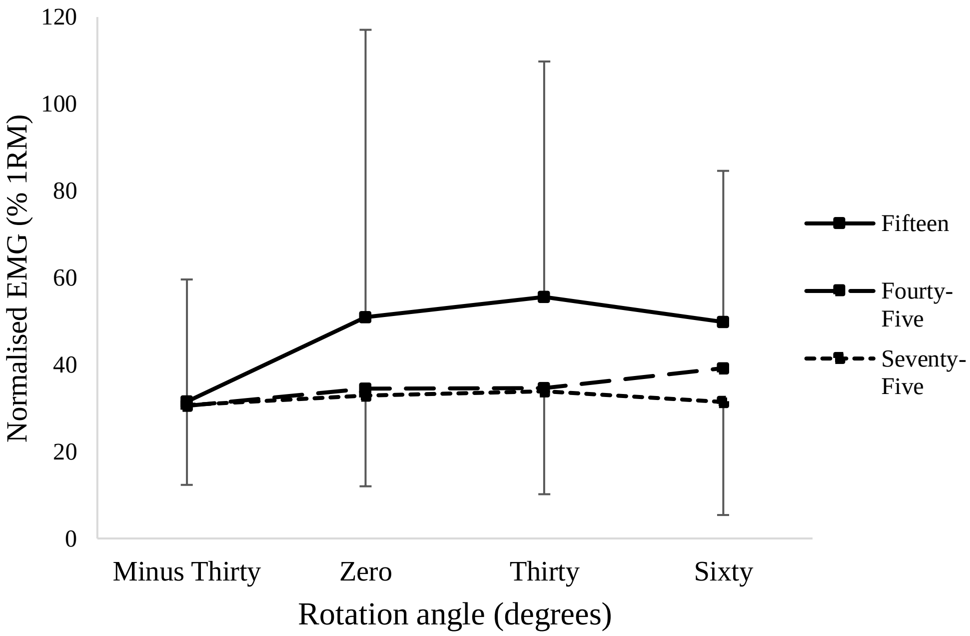

Figure 5.11: Mean ± SD of normalised peak EMG for latissimus dorsi across the twelve settings on the MuJo™ Internal Machine. Legend denotes adduction angle (degrees).

Within the pectoralis major, activity was moderate throughout all repetitions, except for low activity at 45°/60° (Table A2). EMG amplitude at 15° of adduction was higher than 45° at all angles of rotation, and higher than 75° of adduction at -30° and 0° of rotation ($F_{(1.07, 8.52)}$=0.35, P=0.71) (Figure 5.10). Activation was greatest at 0° of rotation for 15° and 45° of adduction; yet for 75° of adduction, activation was greatest at 30° of rotation. Additionally, EMG amplitude was lowest at 60° of rotation for 15° and 45° of adduction, but at -30° of rotation at 75° of adduction ($F_{(3, 24)}$=6.98, P=0.002). Finally, for the latissimus dorsi, activation was greatest at 15°, and lowest at 75°, of adduction ($F_{(1.22, 9.74)}$=2.29, P=0.16) (Figure 5.11). For 15° of adduction, activity was categorised as moderate at all rotation angles, except for -30°, yet the opposite effect was found at 45° of adduction, and activity was low for all repetitions at 75° of adduction (Table A2). Further, at 15° and 75° of adduction, EMG amplitude was highest at 30° of rotation, yet was highest at 60° of rotation for 45° of adduction ($F_{(1.64, 13.08)}$=1.38, P=0.28).

Whilst some muscles did follow a trend that active phase duration decreased as ROM decreased, an overall trend for all muscles was not observed. The longest duration occurred at 15°/60° for trapezius (3.77 s) and pectoralis major (4.54 s), at 45°/0° for anterior deltoid (4.75 s), and 45°/-30° for posterior deltoid (4.10 s), infraspinatus (4.32 s), and latissimus dorsi (3.40 s). A similar trend was observed for the non-active phase duration, with the exception of the infraspinatus, which showed a slight increase for 75°/60° (2.11 s) compared to 75°/30° (1.96 s). Additionally, the non-active phase duration generally declined as ROM decreased. As shown in Table 5.3, many of the muscles remained active for the duration of the set, having not dropped below the calculated threshold until the end of the third contraction.

Table 5.1. Mean active phase duration (s) for all muscles across abduction and rotation angles for the MuJo™ External Shoulder Machine.

Angle	Muscle				
	Trapezius	**Anterior Deltoid**	**Posterior Deltoid**	**Infraspinatus**	**Latissimus Dorsi**
90°/90°	5.53	5.48	5.30	5.10	5.58
90°/30°	4.66	5.18	4.77	4.68	4.37
90°/60°	5.55	5.18	5.04	4.65	5.16
90°/0°	4.64	4.43	4.46	4.30	4.26
60°/90°	5.06	4.89	4.88	4.47	4.73
60°/30°	4.64	4.74	4.62	4.71	4.02
60°/60°	4.32	4.63	4.96	4.17	4.12
60°/0°	4.37	4.39	4.46	4.19	3.49
30°/90°	4.87	4.68	4.85	4.80	4.59
30°/30°	3.89	3.98	4.41	3.97	3.68
30°/60°	4.59	4.25	4.59	4.43	4.27
30°/0°	3.90	3.94	4.04	3.85	3.43

Table 5.2. Mean non-active phase duration (s) for all muscles across abduction and rotation angles for the MuJo™ External Shoulder Machine.

Angle	Muscle				
	Trapezius	**Anterior Deltoid**	**Posterior Deltoid**	**Infraspinatus**	**Latissimus Dorsi**
90°/90°	2.23	1.94	2.14	2.82	1.92
90°/30°	2.01	1.63	2.08	1.99	2.09
90°/60°	1.68	2.09	2.15	2.37	2.01
90°/0°	1.74	2.01	1.96	2.04	2.16
60°/90°	1.91	2.11	2.00	2.23	2.20
60°/30°	1.84	1.75	1.80	1.89	2.05
60°/60°	2.25	2.12	2.37	2.29	2.22
60°/0°	1.77	1.85	1.71	2.03	2.28
30°/90°	1.87	1.95	1.84	1.96	2.29
30°/30°	1.75	1.81	1.47	1.80	2.35
30°/60°	1.66	1.86	1.82	1.69	1.89
30°/0°	1.65	1.64	1.85	1.75	2.11

Table 5.3 Mean active phase duration (s) for all muscles across adduction and rotation angles for the MuJo™ Internal Shoulder Machine.

Angle	Muscle					
	Trapezius	**Anterior Deltoid**	**Posterior Deltoid**	**Infraspinatus**	**Pectoralis Major**	**Latissimus Dorsi**
15°/-30°	3.38	4.40	3.59	3.94	4.40	3.19
15°/0°	3.20	4.35	4.06	4.06	4.00	3.30
15°/30°	3.25	4.49	3.42	3.99	4.35	3.14
15°/60°	3.77	4.46	3.73	3.56	4.54	3.01
45°/-30°	3.15	4.59	4.10	4.32	4.05	3.40
45°/0°	3.04	4.75	3.89	3.95	4.06	3.30
45°/30°	3.54	3.94	3.06	3.97	3.94	2.83
45°/60°	3.54	4.05	3.62	4.28	3.80	2.56
75°/-30°	3.62	3.91	3.21	3.74	3.86	2.85
75°/0°	3.33	3.63	3.47	3.38	3.73	2.93
75°/30°	2.92	3.68	3.45	3.53	3.44	2.88
75°/60°	2.63	3.52	3.94	2.90	2.91	2.88

Table 5.4. Mean non-active phase duration (s) for all muscles across adduction and rotation angles for the MuJo™ Internal Shoulder Machine.

Angle	Muscle					
	Trapezius	Anterior Deltoid	Posterior Deltoid	Infraspinatus	Pectoralis Major	Latissimus Dorsi
15°/-30°	3.84	3.18	3.71	2.98	2.66	4.01
15°/0°	3.49	2.82	2.86	2.66	2.90	3.47
15°/30°	3.43	2.42	3.25	2.24	2.16	3.16
15°/60°	3.01	2.52	2.83	2.68	1.82	3.09
45°/-30°	3.33	2.56	2.49	2.44	2.94	3.11
45°/0°	3.68	2.38	3.16	2.65	2.57	3.25
45°/30°	3.11	2.63	3.20	2.36	2.31	3.22
45°/60°	3.20	2.96	2.85	2.19	2.27	3.21
75°/-30°	3.19	2.63	2.95	2.60	2.41	2.94
75°/0°	2.79	2.58	2.22	2.52	2.14	2.79
75°/30°	2.42	2.02	2.35	1.96	2.00	2.40
75°/60°	2.12	1.81	1.68	2.11	1.88	2.02

Table 5.5. Within-subject CV (%) for all muscles across abduction and rotation angles for the MuJo™ External Shoulder Machine.

Angle	Muscle				
	Trapezius	Anterior Deltoid	Posterior Deltoid	Infraspinatus	Pectoralis Major
15°/-30°	12.5	16.8	20.9	15.7	17.0
15°/0°	17.0	17.4	15.9	10.4	19.5
15°/30°	21.6	11.1	16.8	13.3	15.6
15°/60°	13.4	22.4	22.6	18.6	18.7
45°/-30°	18.0	20.5	14.6	10.8	14.0
45°/0°	25.5	14.5	15.3	12.7	22.1
45°/30°	26.3	11.6	18.7	10.3	14.9
45°/60°	17.2	11.0	17.4	11.2	13.2
75°/-30°	15.1	25.4	23.2	15.7	22.0
75°/0°	20.3	22.6	16.8	12.0	28.3
75°/30°	20.4	17.1	20.7	13.6	16.8
75°/60°	14.8	14.7	19.3	15.5	16.6

Table 5.6. Within-subject CV (%) for all muscles across adduction and rotation angles for the MuJo™ Internal Shoulder Machine.

Angle	Muscle					
	Trapezius	**Anterior Deltoid**	**Posterior Deltoid**	**Infraspinatus**	**Pectoralis Major**	**Latissimus Dorsi**
15°/-30°	27.7	18.1	39.1	23.2	15.7	33.7
15°/0°	27.0	18.5	23.0	14.4	15.1	17.2
15°/30°	18.3	19.8	21.1	19.9	15.1	23.2
15°/60°	25.3	10.8	24.1	17.5	21.7	19.0
45°/-30°	28.0	24.2	30.2	18.4	22.0	39.7
45°/0°	18.9	22.5	25.8	19.1	16.4	32.5
45°/30°	36.9	14.2	29.3	24.3	19.7	23.8
45°/60°	29.0	18.7	31.6	24.1	19.3	31.9
75°/-30°	36.8	17.1	29.6	23.1	14.5	25.9
75°/0°	29.8	21.1	38.1	29.4	20.2	22.5
75°/30°	28.6	17.2	41.2	22.6	17.8	24.0
75°/60°	25.9	18.4	53.8	29.4	17.6	26.5

6.0 Discussion

Strength and power within the UQ is often considered a performance determinant for many sports, and also may prevent injury to its structures, thus is an important consideration for athletic and general populations alike. Resistance machines are often used by these populations, yet many types of machine are restricted to one plane of movement. However, exercising in multiple directions may be more effective for developing intermuscular coordination, as well as strength. The MuJo™ machines enable such movement over varying ROM, yet little was known about how changing the ROM affects the activity of the muscles used when exercising on these machines. Consequently, the aim of this study was to investigate the effect of different ranges of motion on muscle activation whilst exercising on the MuJo™ equipment. The main findings were that for the MuJo™ External Shoulder machine, EMG amplitude was significantly affected by abduction in the trapezius and anterior deltoid, and by rotation in the latissimus dorsi. Additionally, for the MuJo™ Internal Shoulder machine, EMG amplitude was affected by rotation in the pectoralis major. Furthermore, as ROM decreased, generally the active phase duration also decreased within the External Machine, yet no real trend was observed for the Internal Machine. A consequential key conclusion was that there were different activation levels depending on what role a muscle was performing at a specific time, under different constraints. With a smaller ROM, primarily the agonist musculature was required for the movement; whereas, with a larger ROM, the stabilising musculature was also required, to control the movement in absence of support from the machine. Consequently, these results may have implications for exercise prescription with the MuJo™ equipment.

For the External Machine, EMG amplitude within the trapezius muscle was found to be significantly greater with more abduction ROM; as aforementioned, the trapezius is a scapular muscle, acting to rotate and stabilise the scapula and is thus essential for normal scapulothoracic rhythm (Ekstrom et al., 2003; Jordan et al., 2012). In order for abduction and elevation at the shoulder to occur, the scapula must be positioned correctly so as to ensure adequate glenohumeral stability and allow the prime movers to perform their role (Reed et al., 2013; Reinold et al., 2009, Youdas et al., 2012). Furthermore, the trapezius has been noted to also assist with the initiation of abduction, by working collectively with the other axioscapular and stabilising musculature in a coordinated manner (Reed et al., 2013). Its early recruitment here is considered to be another indication of its scapular stabilisation role.

Within the present study, the upper trapezius demonstrated a higher level of activation at 90° of abduction, and decreased as ROM decreased down to 30° of abduction. This is potentially due to the requirement to fully rotate the scapula in order to allow abduction to 90°, as without scapular movement the deltoid becomes actively insufficient (Schenkman and Rugo De Cartaya, 1987). At 30° and 60° of abduction, however, there is less upward rotation of the scapula, with it instead residing in a state of relative stability (Inman et al., 1944), resulting in less recruitment. An increase in upper trapezius activity with increasing shoulder elevation angle has been noted previously (Lim et al., 2015; Uga et al., 2016), with Lim et al. (2015) also observing significant differences between each abduction angle used. However, Lim et al. (2015) investigated horizontal abduction, and Uga et al. (2016) did not observe any significant differences between elevation angles; this was attributed to the arm being fully supported through ROM, thus minimising the activation required for keeping the arm elevated. Consequently, the divergences in results obtained are likely to be due to methodological differences.

Abduction is often used as a movement to assess shoulder health and adequate function (Reed et al., 2016), particularly with overhead throwing athletes who may demonstrate changes in shoulder ROM over time (Wilk et al., 2002). However, methods vary with the plane in which the abduction is performed, i.e. either the frontal or scapular plane. The MuJo™ equipment enables abduction to be performed in the frontal plane, which has been argued to be less effective than the scapular plane, due to the latter's facilitation of full scapular rotation and optimal musculoskeletal alignment (Alpert et al., 2000; Reed et al., 2016). However, when investigating whether the muscle activation patterns are different between planes, it was observed that abduction may be performed in either the frontal, scapular, or 30° anterior to the scapular plane, with no meaningful difference in muscle activation patterns (Reed et al., 2016). The authors did note a slightly lower upper trapezius activity in the scapular + 30° plane, but concluded that it was not clinically meaningful due to the minimal level of activation it represented (6.2% MVC). Therefore, and particularly as there was no difference between the scapular and frontal planes, assessing shoulder function with abduction on the MuJo™ equipment could be appropriate.

A significant main effect for abduction was also noted in the anterior deltoid, with a significantly higher EMG amplitude observed at 90° of abduction than at 60°. The anterior head of the deltoid is primarily a shoulder flexor, but also an agonist of internal rotation, together with the pectoralis major, latissimus dorsi, and subscapularis (Jordan et al., 2012). However, it is also a synergist of abduction when all heads of the deltoid contract simultaneously, and the deltoids provide an important stabilising force via the formation of a force couple with the rotator cuff (Schenkman and Rugo De Cartaya, 1987). In the present study, with increasing shoulder abduction, the anterior deltoid demonstrated increased activity, similarly to results in the literature (Alpert et al., 2000; de Witte et al., 2014); this may be representative of the fact that the anterior deltoid's moment arm increases as abduction angle increases (Jordan et al., 2012; Reinold et al., 2009). Between 0° and 40° of abduction, the moment arms for the middle and anterior deltoids are less than those for the rotator cuff musculature, indicating perhaps that they are not effective abductors at low abduction angles (Reinold et al., 2009). However, whilst the anterior deltoid may not be optimally positioned to produce abduction torque at lower angles, it can still generate force to provide stabilisation through the aforementioned force couple. This may be reflected by the present results, in which the activity was lesser at 30°, yet not considerably less than at the potentially more effective 60° of abduction. At 90° of abduction, however, the moment arm of the anterior deltoid becomes more favourable so as to generate torque, shifting its role to become an effectual abductor.

The significantly higher activity at 90° of abduction may also provide further support for the concept that the deltoid as a whole will provide stabilisation at this angle due to the advantageous alignment of their fibres (Boettcher et al., 2010; Kido et al., 2003). That the posterior deltoid similarly demonstrated higher muscle activity at 90° of abduction than at 60° and 30°, even though no significant effect was noted, is perhaps an indication of this role. Further, the higher activation of the deltoids observed in the repetitions with less restriction may reflect a motor strategy employed to control the movement, in accordance with hypothesis 2. During those repetitions through a larger ROM, less support is provided by the machine, resulting in an increased requirement for stability from the shoulder musculature itself; thus, greater activity from the stabilising musculature and co-activation from agonists and antagonists is necessitated (Cacchio et al., 2008). As the activity of the posterior deltoid muscle appears to decline almost linearly as both abduction and rotational ROM decreases, with the exception of the slight increase at 60°/30°, this could be considered support for its stabilising function in these movements.

The only significant effect for rotation with the External Machine was observed in the latissimus dorsi. As aforementioned, the latissimus dorsi is a strong adductor, and also assists with internal rotation, thus is logically performing as an antagonist during external rotation. The present results indicate that muscle activation was predominantly highest at 90° of rotation and generally demonstrated a decline as rotational ROM declined; specifically, EMG amplitude was significantly lower at 0° of rotation than at 90° and 60°. In a similar way to the deltoid, with a greater level of ROM the latissimus dorsi perhaps demonstrated more recruitment in order to help support and control the movement; this provides further support for hypothesis 2. Interestingly, the repetitions at 90° and 60° of abduction demonstrated a slight increase in EMG amplitude at 0° of rotation; this coincides with the general increase within the infraspinatus as rotational ROM decreased. It is possible that as the movement became more dominated by the infraspinatus as rotational ROM declined, the antagonist latissimus dorsi was also required, particularly as larger angles of abduction result in a less stable movement.

Significant differences in the latissimus dorsi have also been observed previously, yet this was as a result of abduction angle rather than rotation angle (Park and Yoo, 2013). Interestingly, and in contrast to the present results, EMG amplitude was found to be higher for a lower abduction angle, with the highest values noted at 60° of abduction. This may be due to differences in methods and exercise used, however, as that study focused on differences between variations of isometric pull down exercise. Nevertheless, the authors postulated that the decline in EMG amplitude with increase in elevation angle was due to the length-tension relationship, in that the latissimus dorsi moved away from its optimum length as elevation angle increased (Park and Yoo, 2013). This may not be the case within the present study; isometric exercises were not used, meaning the length of the muscle will have changed throughout the exercise, particularly due to multiple movements in both the frontal and transverse planes.

It is also important to note that each repetition on the External Machine involved both external and then internal rotation, in order to return the upper limb to the starting position. The latissimus dorsi is known to be important in the production of medially and laterally directed forces, with its activity having an impact upon the glenohumeral, sternoclavicular, and acromioclavicular joints (Arwert et al., 1997). Further, it is also considered to be one of the most efficient internal rotators, even with the arm abducted (Itoi et al., 1996). Therefore, the higher activation observed with larger ROM may also be a result of the greater amount of internal rotation required to return the upper limb to the start to begin another repetition. The movement on the External Machine also requires adduction once the limb has reached the top of the movement, wherever that has been defined with the mechanical stops. When investigating the role of six segments of the latissimus dorsi, Brown et al. (2007) observed that all six segments behaved as prime mover segments for adduction, with large adductor moment arms and effective mechanical lines of action. Therefore, a greater amount of adduction due to a larger ROM will logically result in greater recruitment of the latissimus dorsi.

Within the Internal Machine, the only statistically significant effect noted was for internal rotation in the pectoralis major; EMG amplitude was lower in the more constrained movements at 60° than at 30°. The pectoralis major is considered a prime mover for internal rotation (Brown et al., 2007; Itoi et al., 1996; Jordan et al., 2012); thus, superficially, the results appears to refute hypothesis 1. However, it may be that 60° of internal rotation was too restricted, leading to an almost isometric movement that resulted in a lesser recruitment of the pectoralis major. Instead, as values were significantly higher at 30° of rotation, this may have been a more advantageous angle for the pectoralis major to perform its role. Indeed, the highest value was observed for 75°/30°, representing the most constrained adduction angle but with slightly more available rotational ROM; this may have been enough of an increase in ROM to further recruit the pectoralis major. Interestingly, at 75° of adduction, the activation of the anterior deltoid was lower at each angle of rotation compared to 45° of adduction, and at 75°/30° it approached that of the pectoralis major (anterior deltoid: 55.49% 1RM, pectoralis major: 53.18% 1RM). This may perhaps indicate that as rotational ROM declined, the pectoralis major may have been acting as more of an agonist than the anterior deltoid, which instead predominated during repetitions with greater ROM.

When investigating the difference in muscle activation when using plate machines versus less restricted cable machines, EMG amplitude of the pectoralis major and anterior deltoid were significantly higher for the cable machine than those for the plate (Signorile et al., 2017). However, it appears that normalisation was not undertaken, or it was simply not reported. It is recommended this information should always be detailed (Burden, 2008) and as such, the results may be questioned. Nevertheless, the authors postulated the higher muscle activation was a result of the greater ROM available when using the cable machine, and due to the differences in starting and ending angle of the shoulder between machines. Additionally, Cacchio et al. (2008) observed a very high activity level for the pectoralis major using the Freemotion chest press, which affords the user more freedom when moving the limbs to achieve the task. Activity for the antagonist and stabilising musculature was similarly high or moderate during the less constrained task (Cacchio et al., 2008), postulated to be a result of a quantity of force being used for joint stabilisation instead of just torque generation during the less constrained task (Anderson and Behm, 2004; Cacchio et al., 2008). Within the present study, whilst EMG levels for the greatest amount of rotational ROM (-30°) were not the highest, EMG amplitude was consistently high at 0° of rotation across all adduction angles. It may be that at this angle of rotation, the pectoralis major had its most advantageous moment arm, thus enabling it to produce more torque and hence be preferentially recruited. Conversely, it may be that at -30° of rotation, the muscle was lengthened slightly too much, thus disallowing the development of optimal active tension.

During external rotation and abduction, for all angles, the posterior deltoid and the infraspinatus demonstrated the highest EMG activity, and were the only muscles to demonstrate activity above 70% of 1RM. This was expected, as both muscles act as agonists for these movements. At 90° of rotation, the posterior deltoid was more active than the infraspinatus at every abduction angle, but the infraspinatus demonstrated greater activation at almost all other rotation angles. The infraspinatus also had the lowest CVs for the External Machine, particularly at 60° of abduction; equally, the lowest CVs for the Internal Machine were observed for the pectoralis major. Consequently, it appears that the prime movers demonstrated the least variability between trials. Additionally, the infraspinatus activity was highest for the greatest ROM, yet still mostly high for less ROM; whereas the antagonist activity from the latissimus dorsi and anterior deltoid, and the stabilising activity from the trapezius, declined as rotational ROM declined. This trend for different activation patterns during a motor task is similar to the results of Cacchio et al. (2008), who identified a re-organisation of motor strategy to optimise muscular effort.

Level and timing of muscle activity is considered to be dependent upon movement direction and speed, and so is a function of what role the muscle is fulfilling at the time (i.e., agonist, antagonist, or synergist) (Macpherson, 1991). During less constrained repetitions, the agonist activity is maximised to produced force against the external load in the desired movement direction, whereas activity of the stabilising musculature is increased so as to oppose the load in all other directions, thus controlling the movement (Cacchio et al., 2008). This reflects the differing roles for monoarticular and biarticular muscles already discussed, and, indeed, may complement the already stated benefits of training biarticular muscles. Whilst this warrants future research, a more efficient distribution of moments across joints, together with the optimised motor strategy developed by training with this equipment, may lead to considerable improvements in intermuscular coordination.

During internal rotation and adduction, both deltoids consistently demonstrated the highest muscle activation of all muscles measured, with some repetitions resulting in activation of over 90% 1RM. Again, this was expected as the anterior deltoid and posterior deltoid act as agonists and antagonists respectively during internal rotation and adduction. The posterior deltoid had the highest CVs for both machines, and CV values increased to the maximum observed of 54% as ROM decreased on the Internal Machine. Indeed, the variability observed throughout was relatively high, particularly for the Internal machine. This may be a result of many trials only consisting of one repetition instead of three on the Internal Machine, but it also may reflect a tendency of the stabilising/antagonist musculature to be more variable between conditions. Interestingly, the trapezius EMG activity is also high or moderate throughout the internal movements; the participants were instructed to keep their heads against the head rest at all times, so as to remove unwanted trapezius activity from supporting the head. However, the starting position for the Internal Machine may have caused some extraneous trapezius activity; the arms begin abducted and externally rotated, so the trapezius will be actively maintaining the rotation of the scapula to allow for the abduction. It is perhaps surprising that the activity of the pectoralis major was not categorised as high for any repetitions, considering its agonist role; the highest value was 53.18% at 75°/30°. However, the loads lifted during the protocol were relatively low, being just 50% of 1RM; moreover, the present values are not too dissimilar from those in previous literature (Cacchio et al., 2008; Signorile et al., 2017).

In accordance with this, a potential limitation of the thesis is that activity of the subscapularis was not measured, as this muscle is considered an important internal rotator (Jordan et al., 2012). Additionally, together with the inferior and middle glenohumeral ligaments, the subscapularis works as an important stabilising structure of the glenohumeral joint, especially with the arm abducted to 45° (Jordan et al., 2012). It may have also been pertinent to investigate the activity of the supraspinatus, considering its vital role in abduction (Reed et al., 2013); including these muscles may have provided further information into the activity of the UQ during the movements involved. However, measurement of subscapularis and supraspinatus activity via surface electrodes can be fraught with complications, including accidental identification of overlying muscles such as the pectoralis major and deltoids. Indeed, whilst relationships between surface and indwelling electrodes have been observed when examining the subscapularis and supraspinatus, large overestimations and bias in predictive equations indicate that surface electrodes may not be appropriate to validly measure their activity (Waite et al., 2010). Consequently, if these muscles are to be included in future work, indwelling electrodes are recommended.

Alongside this, a further limitation may have been present in the form of crosstalk; whilst a double differentiator and a bipolar electrode configuration were used, it is possible that the signal contained energy from surrounding musculature. This may be particularly pertinent regarding the infraspinatus and latissimus dorsi. When comparing the infraspinatus activity recorded from surface and intramuscular electrodes during isometric and dynamic tasks, Johnson et al. (2011) found that activity levels recorded by the electrodes varied under different conditions. Both types of electrodes recorded similar patterns of recruitment whilst the muscle was activated to a high and moderate degree, but different patterns were observed during exercises that should have resulted in low levels of activity. High activity was still observed with the surface electrodes, which the authors concluded was likely due to crosstalk (Johnson et al., 2011). Consequently, the infraspinatus EMG data is likely to be accurate from the External Machine, but possibly may have been overestimated during the repetitions on the Internal Machine. This effect has also been observed in the latissimus dorsi, in which surface electrodes overestimated activity from the muscle during tasks where it would be expected to be minimally activated (i.e., flexion and abduction) (Ginn and Halaki, 2015). Accordingly, techniques to confirm the presence or absence of crosstalk would perhaps have been advantageous; such techniques include functional tests or cross-correlation (Burden, 2010; Winter et al., 1994). Functional tests involve the participant contracting surrounding muscles whilst keeping the muscle of interest silent; however, the participant may not be capable of preferentially activating adjacent muscles, and/or also not simultaneously activating the muscle of interest (Burden, 2010; De Luca, 1997). Alternatively, the frequency spectrum of the signal can be examined, to determine the extent to which it is composed of lower frequencies, as expected with a muscle distant from the electrodes (De Luca, 1997). As such, either this technique or cross-correlation should be included in future investigations.

It was hypothesised that those repetitions that were less restricted would comprise longer active phase durations than those with more constraint, as the limb travels further over greater ROM, leading to a longer contraction. For the External Machine, this was mostly observed; there was certainly an overall trend that active phase duration decreased as ROM decreased, yet there were a few exceptions. As aforementioned, for all muscles, active phase duration at 30°/30° was slightly longer than that at 30°/60°, even though the former comprises less rotational ROM. Additionally, for many muscles this oscillating pattern of alternatively higher and lower active phase durations throughout the angles was quite common, particularly with the latissimus dorsi. It is not completely clear why this is the case; however, it may be due to inter-individual variation in approach to the motor task. Electromyography is considered quite variable (Kamen, 2014), an issue possibly exacerbated by the novelty of both the Machines and task to the participants. Whilst training status was mostly consistent between participants, individual variations in motor control may have caused the unexpected discrepancies.

No consistent pattern could be observed for the active phase duration on the Internal Machine, even when omitting those contractions where the signal did not drop below the threshold until the end of the third contraction. Indeed, this tendency to remain active occurred quite often during the internal repetitions, perhaps an indication that the movements on the Internal Machine are more variable than those on the External Machine. Further, it is interesting to note that this occurred most often in the posterior deltoid, followed by the trapezius and infraspinatus, and less so in the anterior deltoid, pectoralis major, and latissimus dorsi. Consequently, it appears the antagonist and stabilising musculature had a greater propensity to remain active for the motor task, perhaps to provide further support and control to a relatively unfamiliar task on novel equipment. Conversely, on the rare occasions that the pectoralis major and anterior deltoid did remain active for all three contractions of one set, they did so during the more constrained task, potentially reflecting their agonist role during internal rotation and adduction.

The raw EMG data were often considerably noisy, an additional limitation of the present study. Whilst efforts were made to adequately filter the signals, and maximise the signal-to-noise ratio, it would seem this was not uniformly achieved. This was predominantly an issue when considering that the pectoralis major data for the External Machine had to be discarded due to too much noise (see appendix 2). As a further consequence, with the data that were processed and analysed, the excess noise makes accurate identification of onset and offset times difficult, and may have inflated the peak EMG values obtained from each trial. Consequently, in future work, filtering should be optimised. The EMG signals were filtered according to the guidelines from the British Associate for Sport and Exercise Sciences, and the International Society of Electrophysiology and Kinesiology (Burden, 2010; Merletti, 1999); however, it may have been more appropriate to analyse the frequency spectrum of both the signal and noise, via Fourier transformation (Winter, 2004). In doing so, specific cut-off frequencies could have been set for the bandpass filter.

6.1 Practical Implications

Cognisant to the present results, it may be beneficial to consider which settings would be most appropriate for targeting specific musculature. If aiming to target the prime movers of a movement, it could be recommended that ROM is restricted to that range which will best reflect their length-tension and joint-torque relationships. For example, at the more restricted angles of abduction, the infraspinatus (considered to be prime mover for external rotation) demonstrated the highest activation for the least amount of ROM in rotation. Alternatively, if attempting to develop strength and motor control for a movement pattern, involving much of the UQ's musculature, it may be more beneficial to allow a bigger ROM through which the limb can travel. As an example, on the External Machine, the latissimus dorsi and posterior deltoid were recruited most highly during the repetitions with greatest ROM in abduction and rotation (90°/90°). However, it is important to note that whilst muscle activity is correlated to force production, the two do not equate to each other as the relationship is not strictly linear (Disselhorst-Klug et al., 2009). Nevertheless, provided it has been measured and analysed appropriately and meticulously, EMG can provide an adequate estimate of the level of excitation of the respective muscles.—Importantly, however, the length-tension and joint-torque relationships will vary due to training status, sex, and age (Brughelli and Cronin, 2007; Narici et al., 2003; Winegard et al., 1997); thus, recommendations on settings for the machines must be made according to the specific goals of the participant, or at the discretion of their healthcare provider.

MuJo™ aims to provide improved musculoskeletal care, the facilitation of targeted athletic training, and structured rehabilitation following or as an alternative to surgery to the shoulder. Based upon the results of this study, it could be an option to use these machines as part of a staged rehabilitation programme. After the shoulder has been passively moved through various ranges following surgery, active movements through ROM are recommended (Funk, 2016). As such, the participant can gradually move the supported upper limb through increasing ROM using the machines with no load added; the consequential changes in muscle length and tension will stimulate muscle spindles and the Golgi-tendon organ, thus gradually restoring neuromuscular coordination (Purves et al., 2001). Controlled isometrics are also often recommended as part of the initial stages of such a programme; these exercises aim to strengthen key structures of the shoulder and prevent atrophy, whilst developing proprioception and neuromuscular coordination (Manske, 2016; Neumann, 2013). These can be performed on the equipment by setting the mechanical stops to a small range that is safe and specific to the injury, and that will develop strength in the desired musculature. Additionally, as strength gains are joint-angle specific (Godfrey and Whyte, 2006), this small range can be moved to different angles to develop strength throughout ROM. Often, the rotator cuff must be targeted and developed, to ensure future stability and avoid re-injury; the MuJo™ machines provide an optional modality with which to accomplish this. The arm is supported and the range can be pre-set before the participant begin; therefore, recently repaired structures will not be subjected to excess stress, and safe but challenging exercise can be performed.

Scapular stability is also an important aspect of rehabilitation for the shoulder, as smooth and efficient interaction between proximal stabilising and distal mobilising structures is essential for effectual shoulder function (Neumann, 2013). Therefore, exercises to target the trapezius and serratus anterior should be included, such as controlled abduction/adduction exercises performed on each machine. The present results seem to indicate that the upper trapezius is more preferentially activated with more abduction ROM on the External Machine, yet less ROM on the Internal Machine, but that less rotational ROM appears to be optimal for both machines. As aforementioned, however, the precise settings ought to be recommended by the participant's healthcare provider, to reduce the risk of re-injury. Regardless of the range chosen, provided they are performed in a safe and controlled manner, exercises performed on these machines should target the axioscapular musculature and promote normal scapulohumeral rhythm. Additionally, once strength and control have been developed, load can be added to the exercises to further improve stability in the rotator cuff, and strengthen prime mover musculature. Following this, loaded exercises may be performed over full ROM, to target stabilising muscles and improve motor control. That many stages of the rehabilitation process can be performed on one machine demonstrates the MuJo™'s potential for use within a therapeutic environment. However, the benefits must be weighed against the limitations discussed throughout the thesis; for example, the relatively high within-subject variability and the potential for crosstalk.

7.0 Conclusions

This thesis aimed to investigate the effect of altering ROM on the activation of muscles used whilst exercising on the MuJo™ External and Internal Shoulder Machines. The findings provided some support for hypotheses 1 and 2, but not comprehensively, and many of the active phase durations refuted hypothesis 3. Consequently, the hypotheses cannot be accepted.

For the External Machine, significant effects were observed for abduction within the upper trapezius and anterior deltoid, and for external rotation within the latissimus dorsi. It was suggested that the upper trapezius was highly activated during more abduction ROM in order to provide stabilisation, and to rotate the scapula to allow for full abduction. The anterior deltoid demonstrated increased activation with more ROM to assist with abduction and provide an important stabilising force together with the rotator cuff. It was also proposed that the latissimus dorsi was activated to a higher degree with more ROM as it was acting as an antagonist to external rotation, and also assisting with internal rotation to return the limb to the start position for the next repetition. Additionally, the posterior deltoid and infraspinatus demonstrated the highest EMG amplitude throughout all angles, which was expected as they are agonists for the movements involved on the External Machine.

Regarding the Internal Machine, the only significant main effect found was for internal rotation in the pectoralis major muscle. It was postulated that 0° of rotation was the most advantageous angle, potentially due to it resulting in the optimal moment arm length; however, without further data this is speculation at the present time. In addition, both deltoids demonstrated the highest levels of activation, and the upper trapezius also showed high to moderate activation levels consistently throughout; this was suggested to be due to the starting position of the arms in abduction on the Internal Machine, requiring the scapula to be rotated.

The active phase duration was mostly longer for the repetitions consisting of more ROM, but with a few exceptions; it was considered these were likely due to inter-individual differences in coordination and technique. Further, generally it appeared that differences existed in the level of activation exhibited by certain muscles, depending upon the role they were performing at the time. This was concluded to be due to changes in motor strategy, in which high force generating capacity was required from agonists during more restricted tasks, but large forces were required from all musculature during less restricted tasks. This was due to a quantity of force being used for joint stabilisation and control of the movement, as opposed to strictly acting against the external load in the preferred direction during the more restricted tasks; therefore, with larger ROM there may be a greater potential for developing motor control.

Future investigations should perhaps focus upon differences in kinematics between ranges of motion, and also on the forces and moments occurring in the UQ. Electromyography provides an indication of the activation of different muscles, but not information on the forces that may result; consequently, questions on how force is applied may be answered with the use of inverse dynamics. This may help further elucidate some of the changes in activation observed, and lend support to or refute the suggestions made regarding the differences found. Additionally, now that an understanding has been obtained of how an asymptomatic population responds to the equipment, studies examining injured or other populations of interest would further contribute to the body of knowledge. Ultimately, however, a randomised control trial is required to obtain a true and valid understanding of the full impact the MuJo™ equipment may have in the treatment and prevention of musculoskeletal disorders.

To conclude, the patterns observed in muscle activity as a consequence of altering ROM may reflect the greater requirement for stabilisation from the shoulder during repetitions over a greater range. In order to utilise these devices effectively, users must consider if the goal is to target and recruit the agonists of a movement, or if it is to improve intermuscular coordination over a larger ROM.

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9.0 Appendices

9.1 Appendix 1

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Figure A.1 MuJo™ External Shoulder Machine

Some materials have been removed due to 3rd party copyright. The unabridged version can be viewed in Lancester Library - Coventry University.

Figure A.2 MuJo™ Internal Shoulder Machine

9.2 Appendix 2

To illustrate the noisy characteristics of the data and further elucidate why the pectoralis major data for the External Machine was discarded from the analysis, below are some examples of signals from two muscles. Both figures display data from one set of three repetitions from the same participant; the blue lines represent the bandpass filtered data, with the overlying orange line showing the RMS data. Figure 1 shows activity from the anterior deltoid, demonstrating the active and non-active phases which were easily identified via the threshold analysis. Figure 2 shows activity from the pectoralis major from the same participant and set; due to the excess noise, threshold analysis could not take place, and manual identification proved impossible to obtain an accurate and reliable value. Consequently, it was decided that the data ought to be discarded.

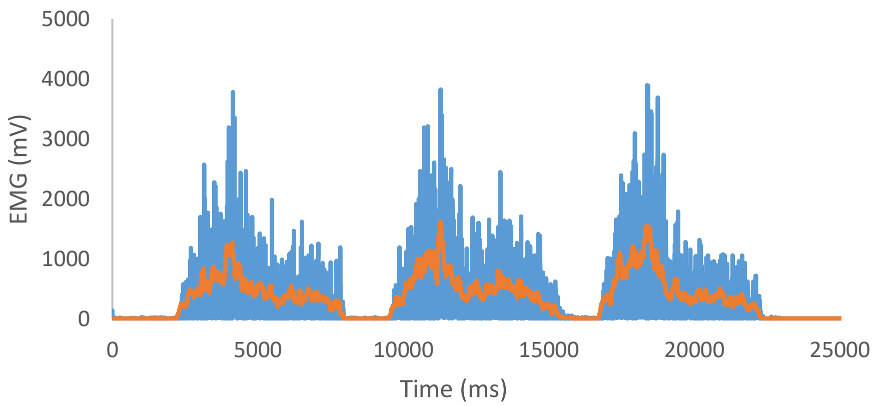

Figure A.3. Filtered and RMS data from the anterior deltoid.

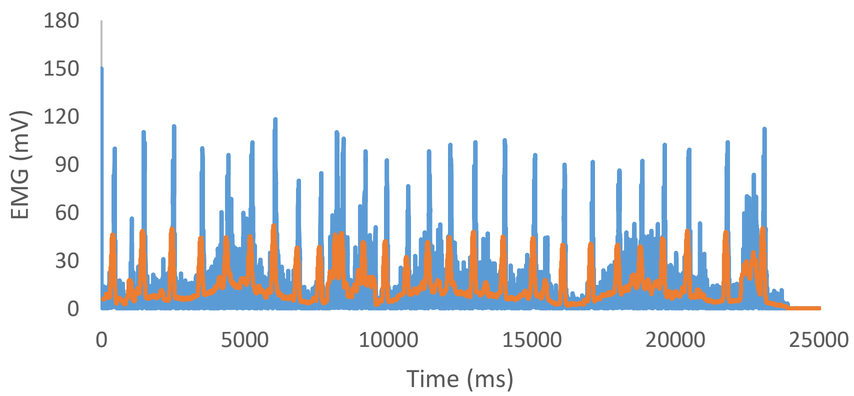

Figure A.4. Filtered and RMS data from the pectoralis major.

9.3 Appendix 3

Table A.1. Mean ± SD of normalised peak EMG (% 1RM) values for all muscles across abduction and rotation angles for the MuJo™ External Shoulder Machine.

Angle	Muscle				
	Trapezius	**Anterior Deltoid**	**Posterior Deltoid**	**Infraspinatus**	**Latissimus Dorsi**
90°/90°	51.3 ± 51.6	44.6 ± 18.4	83.8 ± 53.3	72.3 ± 34.1	47.1 ± 42.6
90°/30°	49.6 ± 42.9	44.7 ± 17.1	72.9 ± 39.1	72.0 ± 37.2	41.4 ± 30.3
90°/60°	53.6 ± 41.6	51.4 ± 25.8	70.9 45.2	87.3 ± 52.8	33.8 ± 27.8
90°/0°	60.0 ± 39.3	47.3 ± 20.7	69.1 ± 41.8	78.3 ± 41.6	34.8 ± 33.7
60°/90°	35.5 ± 32.7	39.8 ± 18.8	73.2 ± 32.4	65.4 ± 33.6	42.6 ± 39.0
60°/30°	35.6 ± 32.1	40.8 ± 17.8	62.5 ± 28.5	69.5 ± 36.0	33.8 ± 24.6
60°/60°	39.0 ± 32.4	39.9 ± 18.4	69.0 ± 32.3	71.7 ± 41.9	27.6 ± 21.4
60°/0°	42.4 ± 40.1	37.8 ± 15.9	60.2 ± 31.7	73.4 ± 45.8	29.9 ± 28.2
30°/90°	30.6 ± 35.8	35.3 ± 16.7	73.5 ± 39.8	71.8 ± 39.7	42.2 ± 36.7
30°/30°	25.0 ± 20.6	37.9 ± 20.1	67.3 ± 29.8	62.8 ± 33.5	43.1 ± 46.2
30°/60°	27.4 ± 25.7	34.8 ± 15.4	60.8 ± 36.2	67.5 ± 43.4	33.3 ± 32.5
30°/0°	21.2 ± 16.4	33.1 ± 17.7	52.7 ± 33.3	76.0 ± 41.5	27.0 ± 26.3

Table A.2. Mean ± SD of normalised peak EMG (% 1RM) values for all muscles across adduction and rotation angles for the MuJo™ Internal Shoulder Machine.

Angle	Muscle					
	Trapezius	**Anterior Deltoid**	**Posterior Deltoid**	**Infraspinatus**	**Pectoralis Major**	**Latissimus Dorsi**
15°/-30°	70.2 ± 82.8	71.1 ± 54.1	67.7 ± 48.3	43.7 ± 14.5	49.1 ± 27.3	31.5 ± 28.1
15°/0°	69.9 ± 45.8	78.1 ± 55.3	70.2 ± 58.0	45.4 ± 25.3	51.1 ± 25.3	51.0 ± 66.1
15°/30°	66.0 ± 46.3	74.9 ± 56.7	67.7 ± 47.0	42.9 ± 17.6	48.3 ± 25.3	55.6 ± 54.2
15°/60°	65.7 ± 36.7	97.1 ± 84.9	70.6 ± 40.1	43.1 ±18.9	36.6 ± 20.3	49.8 ± 34.8
45°/-30°	63.2 ± 73.3	72.7 ± 50.7	60.8 ± 52.4	46.4 ± 16.1	44.2 ± 27.1	30.5 ± 22.9
45°/0°	66.4 ± 75.0	95.8 ± 76.9	75.4 ± 70.5	40.3 ± 13.2	50.6 ± 28.8	34.5 ± 37.6
45°/30°	75.7 ± 68.3	76.6 ± 59.1	75.9 ± 74.5	40.3 ± 16.2	43.7 ± 24.8	34.5 ± 28.7
45°/60°	73.1 ± 44.9	85.5 ± 73.1	84.1 ± 49.1	40.4 ± 15.9	33.7 ± 14.3	39.2 ± 24.3
75°/-30°	73.9 ± 64.6	67.9 ± 54.1	97.5 ± 104.5	42.3 ± 22.3	39.3 ± 20.7	30.7 ± 18.4
75°/0°	82.1 ± 88.5	60.8 ± 45.3	85.7 ± 96.1	50.5 ± 19.8	48.9 ± 28.5	32.9 ± 20.9
75°/30°	81.1 ± 80.8	55.5 ± 39.2	67.9 ± 60.6	43.5 ± 12.6	53.2 ± 35.7	33.9 ± 23.7
75°/60°	92.2 ± 80.2	67.2 ± 48.3	72.9 ± 64.4	37.5 ±16.1	41.1 ± 25.4	31.4 ± 26.0