Sex differences in the neural control of muscle

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Dedication

This work is dedicated to Erika Ainsley and Reese Avery

Abstract

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Sex-differences in muscle strength have been linked to differences in muscle size, involved limb, and daily activities. Early work has shown that sex-differences are greater in the upper compared to lower limb, making the upper limb an ideal model to investigate the best statistical approaches for sex comparison. Large differences in the upper limb reveals how biomechanical factors may impact neural control. Since males and females are more comparable with respect to strength in the lower limb, it allows for a determination of whether potential sex-differences in neural control exist without large differences in biomechanics. Understanding sex-differences allows for prescription of rehabilitation and training modalities, taking into account potential specificities in sex-related neuromuscular and musculoskeletal factors. The overall purpose was to examine neural and biomechanical differences that would account for sex-differences in neural control of muscle.

Manuscript 1 examined normalization versus an ANCOVA to assess sex-differences. Sex-differences were seen in elbow flexor strength and rate of force development (RFD). Normalization by either maximum strength or neural factors couldn't account for all sex-differences in RFD, resulting in an ambiguous interpretation. In contrast, both variables were able to be incorporated in an ANCOVA to determine their relative contribution.

Manuscript 2 examined the effect of task familiarization and the contribution of maximum strength, twitch contraction time, muscle fiber condition velocity, and rate of muscle activation to sex-differences in the RFD during dorsiflexion. There were no significant differences between the sexes in muscle properties, but there were differences in neural control. Additionally, across days females exhibited a neural adaptation leading to an improvement in the RFD.

Manuscript 3 directly assessed potential sex-differences in neural control during force gradation by recording motor unit activity during maximal and submaximal contractions. Females had less force steadiness (FS), which may have resulted from neural compensation for a less optimal pennation angle or a tendency towards greater joint laxity. Higher motor unit discharge rates and incidence of doublets may increase twitch force summation leading to a reduction in FS. Thus, biomechanical, not inherent sex-differences in neural drive led to neural compensation strategies manifesting as a difference in FS.

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List of Abbreviations

ANOVA	Analysis of variance
ANCOVA	Analysis of covariance
AP	Action potential
ATP	Adenosine-triphosphate
BB	Biceps brachii
$[Ca^{2+}]$	Calcium concentration
Ca^{2+}	Calcium
CMAP	Compound muscle action potential
CSA	Cross-sectional area
CT	Contraction time
$d\tau/dt_{max}$	Maximum rate of torque development
EDL	Extensor digitorum longus
EHL	Extensor hallucis longus
EMD	Electromechanical delay
EMG	Electromyography
FFT	Fast Fourier transform
IDI	Inter discharge interval
MAV	Mean-amplitude-value
MDF	Median power frequency
MFCV	Muscle fiber conduction velocity
MN	Motoneuron
MNF	Mean power frequency
MTC	Muscle tendon complex
MU	Motor unit
MUDR	Motor unit discharge rate
MVC	Maximal voluntary contraction
PCSA	Physiological cross-sectional area
PEC	Parallel elastic component
pps	Pulses per second
PT	Peroneus tertius
Q_{30}	Rate of increase in the sEMG over the first 30 ms of muscle activity
R	Intraclass correlation coefficient
RFD	Rate of force development
RMS	Root-mean-square
ROM	Range of motion
RTD	Rate of tension development
RFD	Rate of force development
SD	Standard deviation
SEC	Series elastic component
SEM	Standard error of measurement
sEMG	Surface electromyography
SD	Standard deviations
TA	Tibialis anterior
τ_{max}	Maximum torque
Vpp	Peak-to-peak amplitude

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1. INTRODUCTION

1.1 Background

Primary explorations of human muscle strength were reported at the end of the 19th century in scientific literature. These first studies involved recording muscle strength on cadavers before utilizing dynamometry on living subjects to investigate differences in strength related to sex. With these crude instruments, scientists measured multiple muscles identifying sex differences in strength where men were reported to have two-thirds greater strength. These sex differences in strength were attributed to differences in muscle strain through activities of daily living between men and women (202).

Early work by Hoffman and colleagues (207) used both bench press for upper body strength and leg press to assess lower body strength on both males and females. Their work identified that the underlying reasons for sex differences in both upper and lower body strength are uncertain, and the findings did not support sex difference in strength being solely attributed to body size. When using lean body mass and height in a multivariate analysis of covariance the male participants in the study were still stronger than the females. Interestingly, there was a greater sex difference found in upper compared to lower body strength even when accounting for body size. This led to the conclusion that additional factors must be involved. According to Wilmore (410), the dichotomy of sex differences between upper and lower body may be a result of similarities in lower body use through walking, running and other tasks of daily living. However, the results of the Wilmore (410) and Hoffman and colleagues (207) papers differed slightly. These differences may be due to specific approaches taken when

comparing different participants. To account for size differences, Wilmore (410) utilized ratio scores (normalization) while Hoffman and colleagues (207) used a more theoretically sound multivariate analysis of covariance to account for differences in stature.

Heyward and colleagues (203) investigated males and females with similar patterns of daily activities to see if strength differences in the shoulder flexors and knee extensors disappeared as a result of similar daily strain on the muscles of the upper and lower limbs. However, differences remained between these groups. This further strengthened the suggestion that sex differences in strength are not a result of differences in daily muscle strain but are largely due to differences in lean body mass and its relative distribution (i.e., upper versus lower body). Interestingly, there remained a 27% variance in strength, which was unaccounted for by normalization. Heyward and colleagues (203) suggested that there must be additional factors responsible for these variations, such as differences in muscle fibre type or perhaps motor unit recruitment patterns, which have yet to be identified, and future scientific explorations should peruse this potential explanation.

Sex differences may further be influenced by additional physiological and biomechanical factors. The factors that contribute to sex differences in muscular strength and the rate of tension development (RTD) could be a result of difference in: (1) fibre type distribution (197, 308, 370); (2) muscle and joint stiffness (22, 37–39, 143, 256, 351); (3) muscle architecture (22, 88, 356, 357); (4) potential biomechanical advantages around the joint (289, 290, 342); and/or (5) neuromuscular activation (23, 36, 74, 183, 218).

1.2. Sex Differences in Muscle Fibre Type

Maximum force is partially based on both muscle size and fibre type composition. There is disagreement within the literature regarding sex differences in fiber type proportions (Type I vs Type II) of the various muscles (197, 370). Simoneau and Bouchard (370) showed that males have a higher proportion of Type II fibres in the quadriceps compared to females. The lack of differences in knee extension speed, however, suggests that males and females have similar proportions of muscle fibre types (211, 333).

Activity patterns related to specific muscles "may" play a role in the presence or absence of sex differences in muscle contractile characteristics. For example, the tibialis anterior is mainly used for running and walking which are common activities of daily living (291, 332). It is not surprising, therefore, that sex differences in fibre type proportion are less variable in the tibialis anterior compared to the vastus lateralis (192, 198, 220), as well as its ability to produce tension (238, 239).

Conversely, in the upper limb (biceps brachii), females have been reported to have a Type I fibre area equal to (4, 284) or larger (55) compared to males. Males and females have similar type I areas but the fiber diameter for males is larger (88, 176, 197, 252, 345). If the same were true for the type II fibers, this would explain the greater maximum strength and rate of tension development (164, 296, 297, 308, 327, 355).

1.3. Sex Differences in Muscle Stiffness

The muscle tendon complex (MTC) is made up of both the series elastic (SEC) and parallel elastic (PEC) components (205, 368, 420). The PEC is often referred to as the

passive component (155, 320), while the SEC is considered active (44, 250). As the active component of the MTC, the SEC plays a major role in the stretch shortening cycle through utilization of the stored elastic energy from the muscle stretch and subsequent tension transmission (44, 250). Sex differences in the SEC are believed to be a contributing factor involved in tension development (64).

The rate of tension development depends on both the SEC stiffness and the force-velocity characteristics of the muscle (222, 327). A stiffer SEC increases maximal tension and tension development (411) by more efficiently transmitting muscle force to the bone (327). Muscle stiffness is due in large part to the number of force-generating cross-bridge attachments (129). It has been reported that males have greater muscle tendon stiffness when compared to females (21, 38–40, 143, 256, 351), which may also contribute to differences in the rate of tension development (22, 40). If this is indeed the case, sex differences in the rate of tension development should be accounted for by disparities in SEC stiffness. However, there are mixed findings on sex differences in the SEC (64, 375). It is therefore necessary to investigate other potential factors that may contribute to sex differences in both muscle strength and tension development.

1.4. Sex Differences in Muscle Architecture

The architecture of the muscle includes the physiological cross-sectional area of the muscle, length of the muscle and fibers, and the pennation angle. According to Ikai and Fukunaga (215), isometric strength is proportional to muscle cross-sectional area. A greater fibre area, and larger whole muscle, should therefore result in a significant strength advantage for males (22, 88, 356, 357). In support, muscle fibre area in females

is smaller when compared to males, in both the upper and lower limbs (4, 88, 197, 284, 356). Further, females have significantly fewer fibers in the biceps brachii (BB) (356) and tibialis anterior (197). However, the dichotomy of fiber number may be greater in the upper compared to the lower limb. Work by Levine and colleagues (272) and Heyward and associates (203) showed that men have larger, stronger muscles relative to women, but the differences were greater in the upper limbs opposed to lower limbs.

Multiple stimuli applied to muscle in short succession causes the progressive summation of twitches, resulting in greater tension than could be achieved by isolated twitches (87, 217, 293). Force output achieved by a train of stimuli is termed tetanus tension (63). Gordon and colleagues (166) investigated the length-tension relationship at various isometric tetanus tensions on sarcomeres from the single muscle fibers of frogs. In a series of papers, they suggested that as muscle length is either lengthened or shortened outside of the 'optimal' resting length, the amount of potential tension to be developed by the muscle decreases. Therefore, the length of the muscle fibre must play a significant role in potential tension development. Longer fascicle lengths have been reported in females in both the vastus lateralis (256) and triceps surae (70). These longer muscle fascicle lengths have a tendency to shift the force-frequency curve towards a lower discharge frequency necessary to produce tetanus tension (212, 301). This may be due to increased Ca^{+} sensitivity which is length dependent and is modulated by discharge frequency. Longer muscle lengths may require lower intramuscular [Ca^{+}] to produce the same forces (301). This is believed to correspond to the muscles achieving its optimal length. Therefore, specific joint positions, which can either elongate or shorten muscle fibers, may be an underlying cause of sex differences in tension development. Changing

joint angles to decrease muscle length below optimal causes a reduction in relaxation and contraction time (157, 187, 223, 346), subsequently increasing the required stimulus frequencies to achieve tetanus tension. However, due to the complex interactions between stimulus frequency, tension and joint angles further investigation has been recommended (301).

The prediction of upper body strength cannot be accomplished using the same factors in males and females. This would suggest that there are additional sex differences beyond simple muscle size. These differences may be derived from either anatomical or biomechanical differences acting at the joint (145, 218). Pennation angle is a biomechanical factor that is more subtle than muscle cross-sectional area and length. Muscle pennation angle describes the orientation of fibers compared to the distal tendons line of action (287). A greater pennation angle allows a greater amount of contractile tissue to be placed in parallel within an anatomical cross-section (163, 214, 235, 289).

There is a decrease in muscle length and an increase in pennation angle, when moving from rest to maximal tension (289). Manal and colleagues (289) reported that both male and female pennation angles increase with a rise in tension development, but to a significantly greater degree in males. The change in pennation angle is associated with a commensurate increase in cross-sectional area, which favors males for maximal strength and tension development (290). Maximal muscle tension is achieved when the muscle fibers are at an optimal length and the muscle is fully activated. Optimal muscle length also corresponds to the muscle pennation angle moving towards an optimal angle during tension development (289). The precise influence that pennation angle has on maximal tension and tension development remains to be determined, due to the multitude of

factors involved (375). Additionally, conflicting reports still suggest force generating capacity stems from sex differences in muscle size rather than muscle architecture (22, 88, 183, 209, 214, 356, 357).

1.5. Sex Differences in the Gradation of Muscle Force

It has been suggested that males have a greater tension generating capacity due to muscle recruitment patterns (76, 181, 341). Further evidence suggests females rely more on neuromuscular adaptations during strength training rather than hypertrophy, which may result in an increase in tension per muscle fibre (107). In support, it has been observed that females have a higher magnitude of electromyographic (EMG) activity, which is believed to be part of a tension development strategy (255). Further evidence based on surface electromyography (EMG) characteristics suggests that females may also recruit a larger number of motor units at various contraction intensities prior to substantial increases in motor unit discharge rates (MUDR) to increase tension (76, 403).

Christie and Kamen (74) explored short-term training adaptations in motor unit discharge rates and after-hyperpolarization duration in the tibialis anterior using intramuscular recordings of the myoelectric signal. Males had a greater motor unit discharge rate (MUDR) at maximal voluntary contraction (MVC). However, there was a weak correlation between the two variables, which suggests that other factors may contribute to sex differences in maximum strength. The lower maximal MUDR for females leads to the question of whether MUDRs are lower throughout the force gradation process? As will be detailed in the next section, Ives and colleagues (218) have observed that females also have a lower rate of rise in surface EMG.

1.6. Sex Differences in the Rate of Tension Development in the Upper Limbs

The maximum rate of tension development is as important as maximum strength in functional aspects of human performance (1, 9, 385). Ives and colleagues (218) studied the maximal speed of elbow flexion through a 90 degree range of motion, in males and females. An electromagnet resisted movement initiation at 0, 40 and 70% of an isometric maximal voluntary contraction (MVC), with a quick release once the latter two target loads were achieved. Males had a greater rate of rise in biceps surface EMG during the isometric resistance phase of the task. Furthermore, the rate of rise in the triceps surface EMG in females during the isotonic phase was unable to accommodate the breaking requirements associated with higher accelerations. The results taken together led Ives and colleagues (218) to suggest that females were "neurally constrained", independent of scaling the premovement resistance forces.

Indirect support was provided by Buchman and associates (57), who examined sex differences in muscle activity at the elbow joint while moving through varying distances and velocities. When sex differences in anthropometric measures were covaried out of the linear model, significant differences in muscle activation remained. Thus, anthropometric disparity was insufficient to account for sex differences in elbow flexor tension development, and that additional neural or biomechanical differences may exist. As will be detailed in the next section the findings by Buchman and colleagues (57) run counter to the work of Hannah and associates (183).

1.7. Sex Differences in the Rate of Tension Development in the Lower Limbs

The large disparity in strength between the upper and lower limbs may mask true underlying causes of sex differences in the rate of tension development in the upper versus lower limbs, beyond anthropometrics. Studies on sex differences in maximum strength and the rate of tension development in the lower limb, where the two groups are more comparable (145), have failed to demonstrate male superiority when initial values are accounted for.

If neural control is related to biomechanics, it makes sense that males and females would be similar with respect to the lower limb. Sex differences in the upper limb may indeed reflect "neural constraints" related to very different musculoskeletal mechanics. However, it is well-known that ratio scores to normalize the data will lead to different conclusions from the more appropriate analysis of covariance, to account for initial differences between groups (233, 343).

Hannah and associates (183) investigated sex differences in force and the rate of tension development in the knee extensors, where maximum strength between the two groups is more comparable. In their study, they reported that males were only 33% stronger and 33-57% more explosive over the first 150 ms of tension development. Hannah and associates (183) also compared twitch force and the rate of tension development during evoked contractions. The comparison between evoked and voluntary contractions allows an examination of the relative contributions of central (neural control) versus peripheral factors (muscle contractile characteristics) to maximal effort motor performance (22). It was reported that sex differences in the rate of tension development

of evoked and voluntary contractions were completely eliminated by normalization with maximum twitch force and strength, respectively. These results are in disagreement with the earlier findings of Buchman and colleagues (57) on the upper limb. It is uncertain, however, if the discrepancy is due to the fact that the sexes are more comparable with respect to maximum strength of the knee extensors, or if it is due to a statistical artifact associated with using ratio scores versus an analysis of covariance (233, 343). Thus, the relationship between strength disparity and analysis approach remains an open question.

Parity in muscle strength is even greater at the ankle joint, with males being only 26% stronger during maximal isometric dorsiflexion (270). Not only are males and females more comparable with respect to maximal dorsiflexion strength, but their muscle properties are also similar. Specific strength and tension in the tibialis anterior of males and females are similar (238, 239). There are also similarities in tibialis anterior fibre type distribution (209, 375) and proportion (192, 198, 220). Given the similarities in maximal isometric dorsiflexion strength and tibialis anterior contractile characteristics, this specific action at the ankle is an ideal model to examine potential differences in the neural control of maximal efforts. Furthermore, the impact of the statistical approach on the analysis and interpretation of the results can be expected to be minimal.

1.8. Statement of the Problem and Approach

The dissertation consisted of three studies that addressed current issues in sex differences in the control and regulation of muscle force output. It has been suggested that the lower limbs in males and females are most comparable with respect to the relationship between muscle size and strength (203, 207, 308). As a result, normalization

with respect to muscle size “may” be the reason it explains most of the variance in strength and in the rate of tension development between the sexes (22, 183, 203, 410). Upper limb strength differences between males and females are of a magnitude that suggests other factors, in addition to muscle size, are involved in sex differences in maximal effort contractions (22, 37–39, 64, 143, 256, 351, 375). The normalization issue was addressed first as it affected the acceptance or rejection of the hypotheses in the first study. Maximal isometric strength and the rate of tension development in the elbow flexors were studied first because the magnitude of the sex differences are the greatest to reveal the impact of the analysis of approach.

The second study explored specific neural and muscular mechanisms that are responsible for sex differences, beyond muscle size. The experimental approach was to evaluate sex differences in maximal voluntary versus evoked muscle contractions in the dorsiflexors, to separate neural versus muscular contributions to the maximal rate of tension development. The dorsiflexors were chosen to minimize the impact of initial differences in maximum strength on the maximal rate of tension development, which would allow neural or muscular differences to manifest themselves more clearly than previously observed for the quadriceps (183). There was the added benefit that males and females are also nearly identical with respect to the generation of maximal EMG activity in the lower limb, actually eliminating this confound on the exploration of neural factors (22, 23, 183, 402).

As expected, Study 1 showed that an analysis of covariance proves more appropriate than normalization for exploring sex differences in maximal strength and the rate of tension development. As a result, Study 2 assessed the impact of initial differences

in neural and muscular variables by their contribution to the linear model as a covariate. A critical methodological control missing in other studies was the use of multiple test sessions as the maximal rate of tension development is affected by task familiarization (65, 152); the absence of which, could exacerbate sex differences and bias the results.

Study 3 directly assessed potential sex differences in neural control during force gradation of the muscle by recording motor unit activity. The preponderance of the literature on sex differences has focused on maximal voluntary contractions, for which biomechanical differences may or may not impose neural constraints during the maximal effort condition. The observation that sex differences exist in mean maximal MUDR, indirectly support this hypothesis. The question remained: are males and females similar in the neural regulation of muscle force during submaximal contractions, but differentiate at the highest levels of force output? The second study showed that the sexes are comparable with respect to evoked muscle contractile characteristics, yet, differences in motor unit variables (discharge rate, recruitment, and doublets) were present during submaximal contractions in Study 3, therefore sex differences in neural control do not reflect a limitation (or, "constraint") that is present only during the maximal effort contraction. Specifically, this means the lower strength and rate of tension development observed during maximal effort contractions are not due to any deficit in neural drive but may reflect musculoskeletal differences; the specific nature of which, leads to a new line of investigation.

1.9. Significance of the Problem

The current work represents a comprehensive investigation into potential sex differences in the neural control of maximal effort contractions. The most appropriate statistical approach for the comparison of the sexes was established, without which, hypothesis testing and correct interpretation of the results were impossible. Incorrect interpretation of the results is a formidable barrier to understanding the neuromuscular system (8, 125). The verification of isometric dorsiflexion as a model that minimizes initial differences in muscle size and strength, established the relative contribution of muscle contractile characteristic and neural drive to sex differences in maximal effort contractions. The current series of studies concluded with a more fundamental question about the control and regulation of muscle force.

There is evidence that females have lower maximal MUDRs than males. This dissertation is the first comprehensive study of motor unit activity patterns across submaximal and maximal levels of force, on a large enough scale to compare the male and female force gradation process. In terms of theoretical model development, the results provide insight into whether or not musculoskeletal differences can impact the neural regulation of force, specifically evidence of "neural accommodation". This work can now initiate a new line of investigation where sex differences are viewed as a model to explore how musculoskeletal differences impact the neural control of muscle force output. For example, since males and females exhibit the same muscle contractile characteristics and are more comparable in size and strength in the tibialis anterior during isometric dorsiflexion, then the impact of other musculoskeletal difference can manifest themselves more clearly.

An understanding of potential sex differences in force gradation, was critical for strengthening theories around training modalities and rehabilitation techniques. For example, there is indirect evidence through surface EMG that the rate of increase in activation of muscle in females is lower than that for males. Sex differences in the lower extremity musculoskeletal injuries (114) and greater falls incidence (201, 382) in females may be linked to differences in the maximal rate of tension development (26), which is influenced by neural control (152). Knowledge of the limiting motor unit variable(s) (i.e., recruitment or rate coding) is important for specificity of training and rehabilitation. For example, it is possible to increase the maximal rate of tension development without concomitant increases in strength, through proprioceptive neuromuscular facilitation techniques (152).

2. REVIEW OF LITERATURE

2.0. Ankle Joint

The ankle joint is critical during most activities of daily living (e.g. running and walking). Thirty-three individual bones comprise the ankle and foot and are subdivided into three key joints: talocalcaneal (subtalar), tibiotalar (talocrural) and, transverse-tarsal (talocalcaneonavicular) joint. The transverse-tarsal joint works in conjunction with the subtalar joint during inversion/eversion of the foot. The tibiotalar joint, commonly referred to as the ankle joint, forms the connection between the talus and the tibia and fibula (Figure 1) (52, 200).

2.1. Biomechanics

The ankle joint functions as a hinge joint due to constraints placed on the talus from the malleoli of the tibia and fibula. The ankle joint can perform multiple tasks but is primarily utilized during plantar flexion and dorsiflexion of the foot (Figure 2). Dorsiflexion movements reduce the angle between the shank and foot in the sagittal plane around the line passing through the medial and lateral malleoli. Stability of the ankle joint comes from the ligaments and is greatest during dorsiflexion where the talus is widest anteriorly. The ligaments acting on the ankle joint providing the stability are the anterior tibiofibular ligament, posterior tibiofibular ligament and the interosseous tibiofibular joint (Figure 3) (52). Ankle joint dorsiflexion is dependent upon the elasticity of the ligaments and the syndesmosis, which permits the flexibility necessary for normal ankle joint motion (Figures 4 and 5) (81).

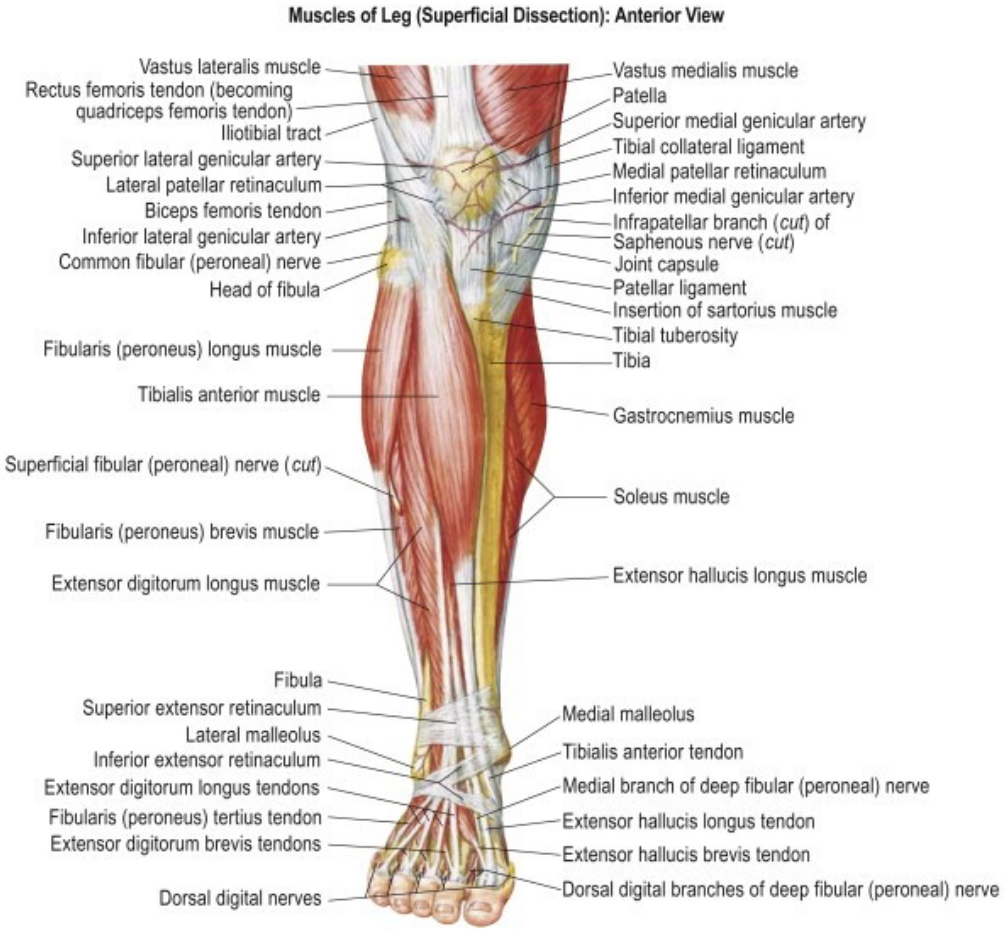

Figure 1. Muscles of the leg, anterior view (406)

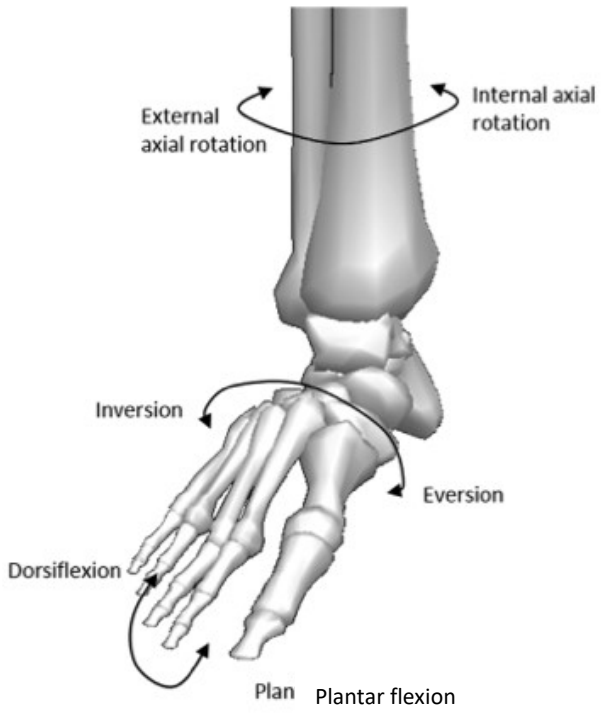

Figure 2. Illustration of the motion of the ankle joint (52).

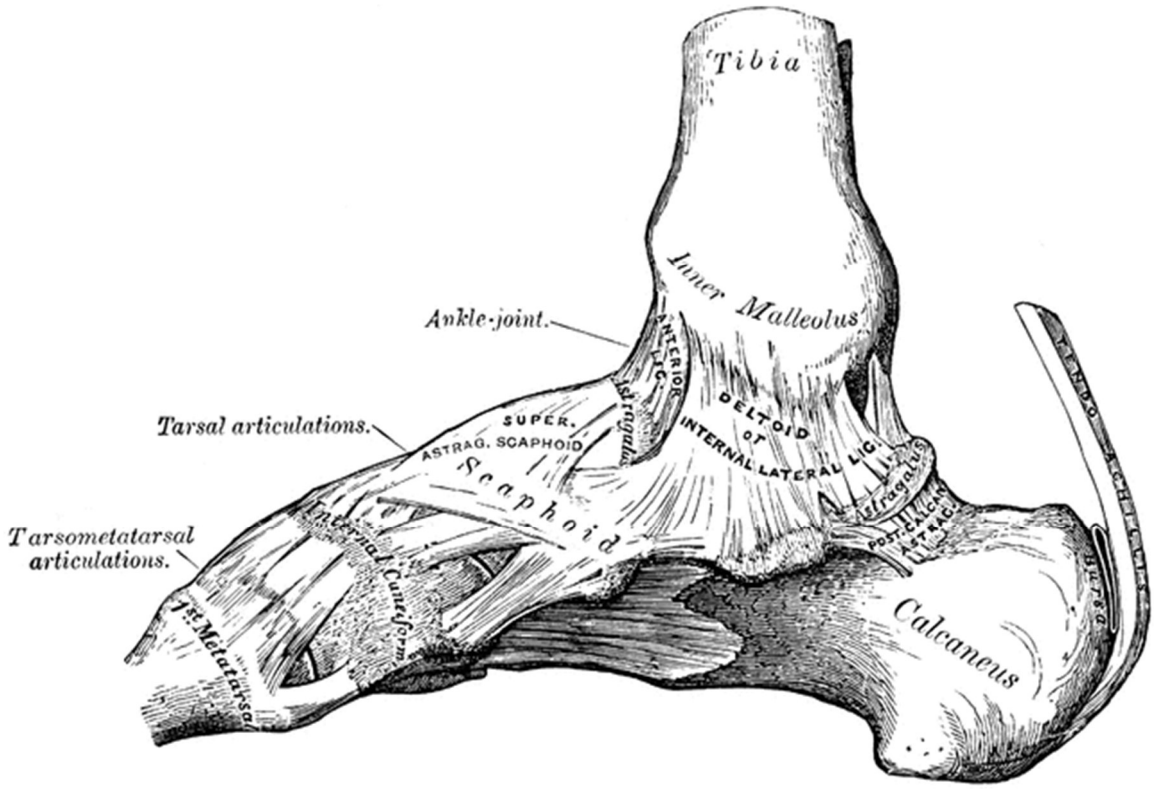

Figure 3. Ligaments of the ankle joint (409)

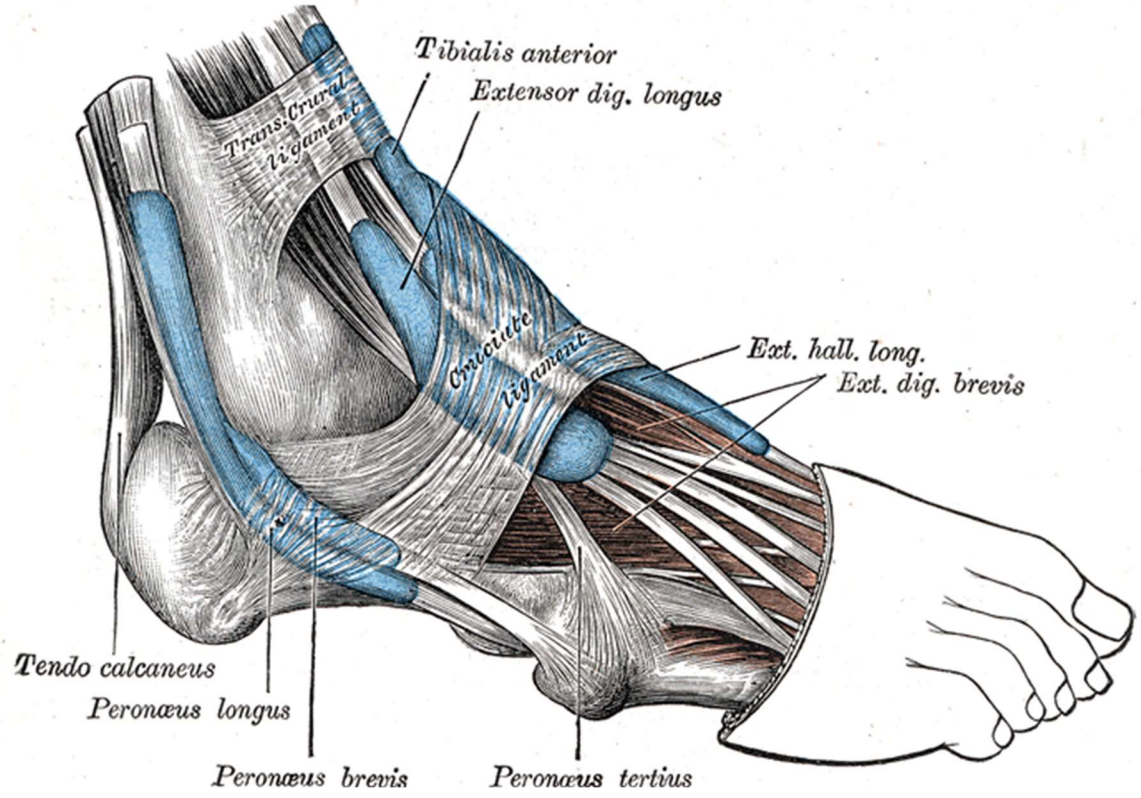

Figure 4. Anatomy of the ankle joint (409).

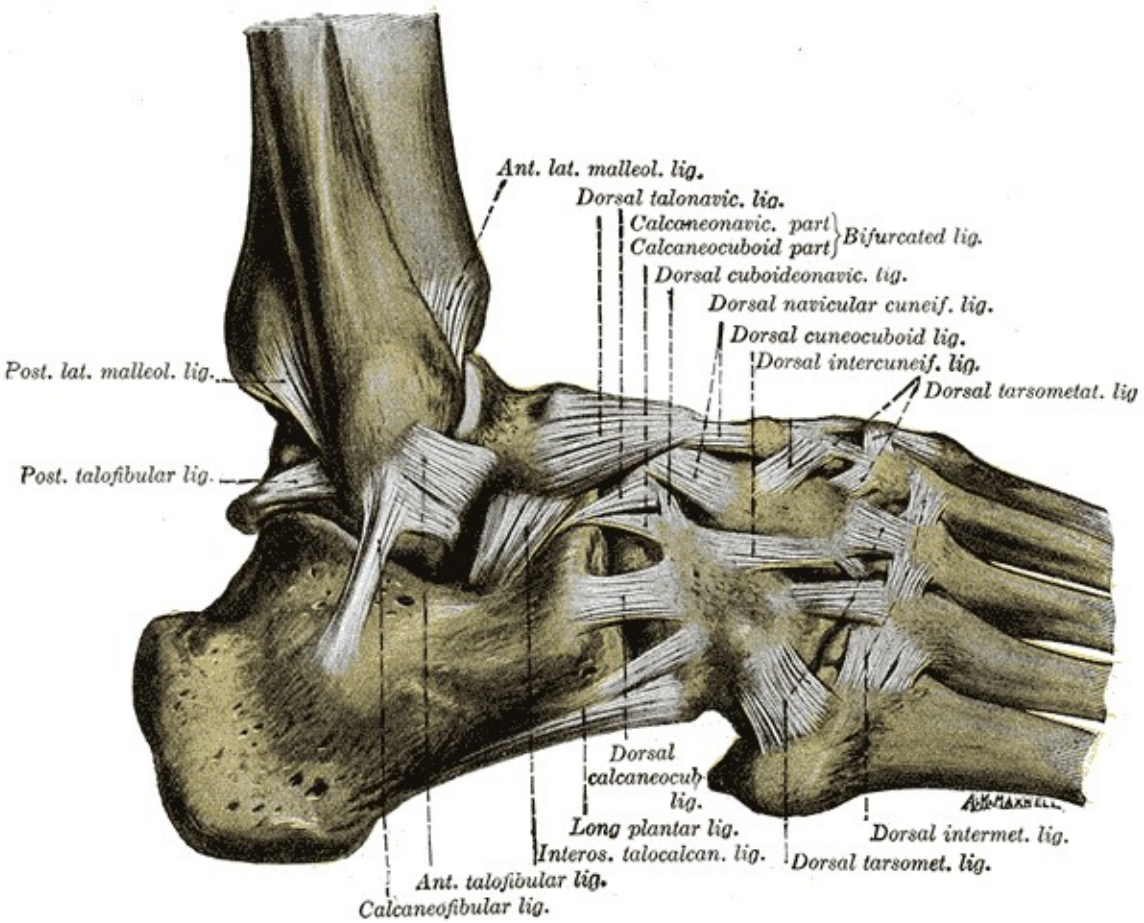

Figure 5. Additional ligaments acting on the ankle joint (409)

2.1.1. Tibialis Anterior Origins and Insertions

The group of muscles responsible for dorsiflexion movements are the extensor digitorum longus (EDL), extensor hallucis longus (EHL), peroneus tertius (PT) and the tibialis anterior (TA) (Figures 1 & 4). The EDL is the most external dorsiflexor originating from the outer tuberosity of the tibia. It spans the upper three-fourths of the anterior portion of the fibula and inserts with four tendons extending across the dorsum of the foot terminating at the distal portion (second and third) of the four lesser phalanges. The EHL is situated between the TA and the EDL originating from the anterior surface of the fibula, internal to the EDL, extending distally and inserting with a tendon at the distal phalanx of the first phalange. Part of the EDL, the PT, originates in the lower fourth of the anterior surface of the fibula and terminates distally with the insertion point of its tendon into the base of the dorsal surface of the fifth metatarsal. Situated on the outer portion of the tibia, the TA's origin is from the outer tuberosity and upper two-thirds of the tibia spanning distally and inserting with its tendon on the inner side of the internal cuneiform bone at the base of the first metatarsal (417). The dorsiflexors are activated by action potentials travelling down the common peroneal nerve (200).

2.1.2. Tibialis Anterior Moment Arms

The amount of change in a muscle fiber length is dependent upon the muscle moment arm. The moment arm is defined as the perpendicular distance between the axis of rotation and the line of action of the applied force (288). A moment arm transforms linear motion into angular joint rotation. Generally, a larger moment arm will allow a greater degree of change in muscle length for a given joint angle, resulting in a larger range of motion (ROM). Greater torque production is possible with a larger moment arm,

but the cost is lower angular velocities (325). Normal joint kinematics depends upon the moment arm and its interaction with the muscle architecture (273).

At maximum voluntary tension the TA tendon moment arm increases by approximately 0.9-1.5 cm causing the displacement of the TA tendons line of action by approximately 0.8-1.2 cm (288) (Figure 6). Although TA length should not change from rest to maximum tension during isometric contractions, the stretching of the superior and inferior retinaculum during isometric dorsiflexion results in the displacement of the TA tendon (Figures 3 & 5). Therefore, regardless of the ankle joint angle the moment arm will increase. The retinaculum are collagen sheaths that act as a mechanical stop to retain the distal tendons of the dorsiflexors in position as they curve around the ankle joint. During an isometric dorsiflexion contraction, the dorsiflexors will shorten, the retinacular collagen fibers will stretch and the orientation of the distal dorsiflexion tendons will shift away from the ankle joint. This deformation of the surrounding fascicles and retinacular bands causes the TA distal tendon to shift away from the axis of rotation when going from rest to maximum muscle tension (288).

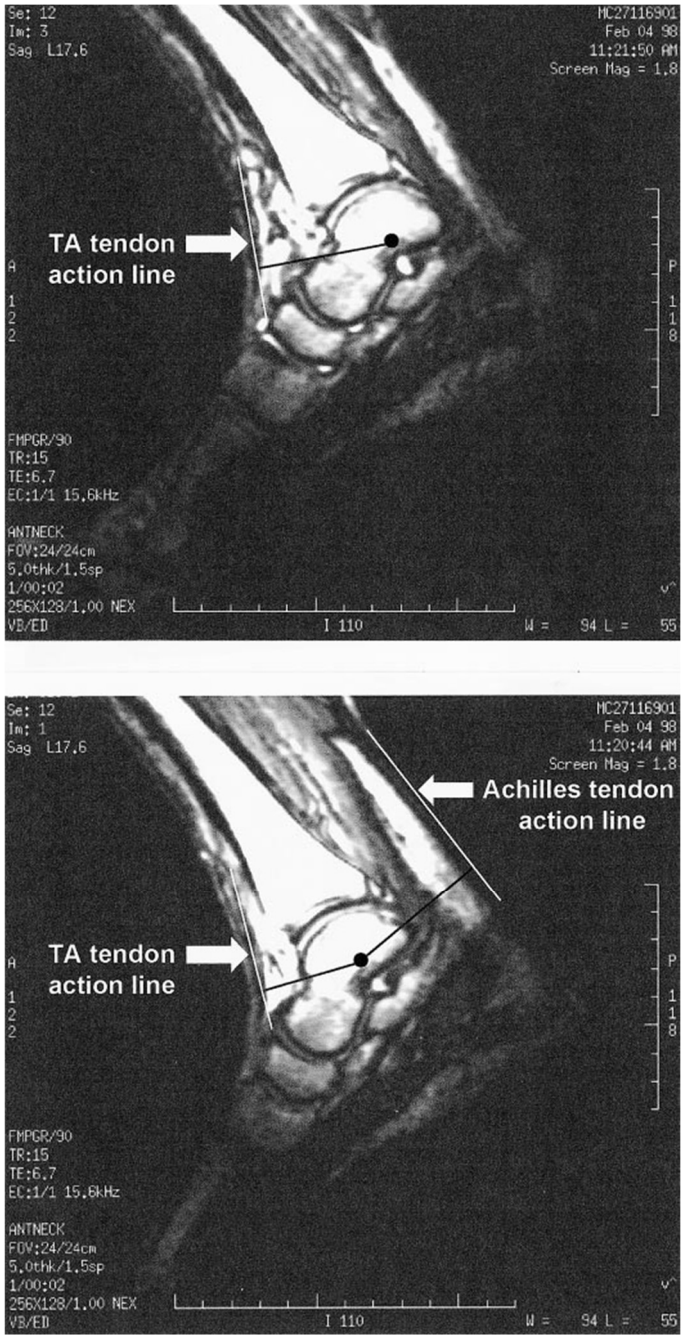

Figure 6. Sagittal magnetic resonance scans of the ankle joint at rest (top) and at MVC (bottom). The joint is in a neutral position with 0º of flexion/extension. Tibialis anterior (TA) tendon moment arm is the perpendicular distance from the black dot to the tibialis anterior tendons line of action (288).

2.1.3. Kinesiological Actions

Dorsiflexion force is produced by the actions of the four muscles mentioned above (EDL, EHL, PT, TA) on the anterior aspect of the shank acting on tendons attached to the foot. The tibialis anterior comprises half of the physiological cross-sectional area (PCSA) of the dorsiflexors and produces dorsiflexion and inversion of the ankle (408) along with the extensor digitorum longus. The peroneus tertius action produces both dorsiflexion and eversion of the foot while the activation of the extensor digitorum longus solely causes dorsiflexion. The range of motion of the ankle joint is approximately 65-75° with young females having greater values than young males (52). Muscles of the ankle joint play an important role in balance, gait and other activities of daily living (209, 265, 369, 390, 391). The tibialis anterior acts as a joint stabilizer during posture maintenance (112) and during the heel strike phase of gait, it is also the dominant dorsiflexor while lifting the forefoot during the swing phase of gait (209, 292, 352).

Joint stiffness is defined as the resistance of a tissue, joint or limb to an alteration in its shape or position (334). This definition allows investigators to consider these biological structures as ‘elastic springs’, the spring are representative of the tissue or joints behaviour as it is bent or deformed. Davis and DeLuca (93) further defined joint stiffness as the manifestation of resistance during inter-segmental displacement resulting from muscles and other joint structures. Stiffness is a mechanical property that determines the absorption and/or transmission of the delivery of external forces to the soft tissue (41, 412). Total joint stiffness involves the contribution of multiple structures that are located in and around the joint (191, 351, 415). For example, there are passive elements (viscoelastic components), but neural drive to the muscle alters the number of

actomyosin cross-bridges (motor unit recruitment) that are formed (169, 224). Superimposed on cross-bridge formation, is the observation that motor unit discharge rate can affect the natural frequency and damping properties of the joint (160). Despite the slight variation in the definition of joint stiffness, it can be agreed upon that active muscle stiffness, which is the stiffness properties exhibited by the musculotendonal tissue, is essential for the maintenance of joint stability (404).

There is evidence that females may exhibit greater ankle joint laxity than males, which would impact the dynamics of joint torque production (154, 170, 171, 251). Riemann and associates (351) studied ankle joint stiffness by passive resistance to dorsiflexion at 5° per second and demonstrated that females have lower passive ankle joint stiffness than males. Extending this work, Granata and colleagues (170) investigated sex differences in dynamic leg stiffness during a hopping task. Since females were shown to also have lower dynamic stiffness values, it was argued that that the hopping task revealed potential differences in active muscle stiffness.

Sex differences in joint laxity have been reported across all ages (221, 334, 416) and may be a result of sex differences in collagen content or the tissues elastic components (250, 251). It has been suggested that disparities in estrogen concentrations between males and females may be responsible for differences in joint laxity (105, 120, 190). Park and colleagues (336) investigated shifts in estrogen concentrations between the follicular and luteal phases of the menstrual cycle and its effect on joint laxity. It was identified that with increases in estrogen concentration there is a concomitant increase in the knee joint laxity measurement by approximately 1-3 mm (336, 337). This increase in laxity may be the result of fluctuations in estrogen concentrations during the menstrual

cycle. These fluctuations may result in structural changes to the ligaments stabilizing the joint. Several studies have investigated the effect of estrogen on the human anterior cruciate ligament (277, 278, 372, 373, 419). These studies suggested the synthesis of collagen is significantly reduced in the presence of higher estrogen concentrations (277, 278, 372, 419), which may result in greater knee joint laxity (373).

Komi and Karlsson (251) suggested that the sex differences in the elastic tissue within the female muscle might be a result of greater rates of elastic energy storage, resulting in a lower rate of force development. This is one way in which sex differences in joint stiffness can affect contractile dynamics. Additionally, increases in joint laxity have contributed to the delayed onset of force production following muscle activation (electromechanical delay) in females (413). Electromechanical delay is believed to result from a certain amount of 'slack' within the musculotendinous tissue and structural components of the joint prior to tension development (24, 413). As tension is developed the elastic elements of the muscles and tendons take up the slack (168), allowing the tension to travel down the tendon's line of action to produce joint torque. More slack in the system (passive, active or structural) causes a longer electromechanical delay. This has been reported during tension development, which is significantly longer in females (24).

2.1.4. Tibialis Anterior Fiber Architectures

The geometric arrangement of muscle fibers in relation to the tendon's line of action affects the manner in which force is transmitted to the joint (149). Muscles are further categorized by their architecture, which is composed of the following variables: (1) muscle length; (2) muscle fiber length; (3) muscle physiological cross sectional area

(PCSA); and (4) muscle fiber pennation angle (273). Muscle length is the distance from the origin of the most proximal fibers inserting into the tendon to the insertion of the most distal fibers inserting into the distal tendon. Muscle fiber length is more difficult to obtain and is often used synonymously with muscle length. The PCSA is calculated from the muscle volume, multiplied by the cosine of the muscles pennation angle and divided by the muscle fibers optimal length (150). Essentially, PCSA is the sum of the number of sarcomeres in parallel within a muscle and is directly proportional to the muscles maximal capacity to generate tension. Muscle pennation angle describes the orientation of the muscle fibers compared to the distal tendons line of action (axis of tension generation) (287). Pennation can either be on one (uni-pennate) or multiple (multi-pennate) angles in relation to the tendons line of action (273).

Differences in pennation angle allow for either a greater or lesser transmission of tension to the tension-generating axis. To this end, a larger pennation angle will allow greater force output compared to a smaller pennation angle due to an increased number of muscle fibers arranged in parallel within a given cross-sectional area. A greater number of fibers in parallel (due to muscle hypertrophy) will increase the PCSA of the muscle and its tension generating potential (150). Pennation angle is not static throughout movement and will vary through the range of motion in proportion to muscle length and tension (290). As pennation angle changes the muscle fibers ‘rotate’ allowing tensile force to be transmitted more efficiently down the tendon, even though the pennation angle may not be optimal for force output (Figure 7) (150).

In the tibialis anterior the muscle architecture consists of two uni-pennated muscle angles converging to make one bi-pennate muscle. It is believed that both portions of the

muscle will contribute approximately half of the force being produced by the tibialis anterior (287). The pennation angles for the tibialis anterior range from 10° to 20° (206, 287) resulting from various ankle joint angles and muscle contraction intensities (206, 287). Furthermore, Manal and associates (289) reported changes in pennation angles when going from rest to maximal muscle tension and found that they were 5° and 3° for males and females, respectively. This small difference in pennation angle results in males having a significantly larger optimal pennation angle in the tibialis anterior (14.7° ± 2.2°) when compared to females (12.1° ± 1.4°).

During the shortening phase of a muscle fiber under isometric conditions, not all of the tensile force is directly transferred to the joint to produce torque. A portion of the force is transmitted to the non-contractile musculotendonous tissue as it stretches under load. The force output from the contractile tissue must overcome the elasticity or slack of both the tendons (aponeurosis) and the actomyosin cross-bridge known as the series elastic component (SEC) and parallel elastic component (PEC) (234). The SEC is considered to be the stiffer of the two elastic components, which allows a greater transfer of tension directly to the muscle and tendons allowing greater torque output (20).

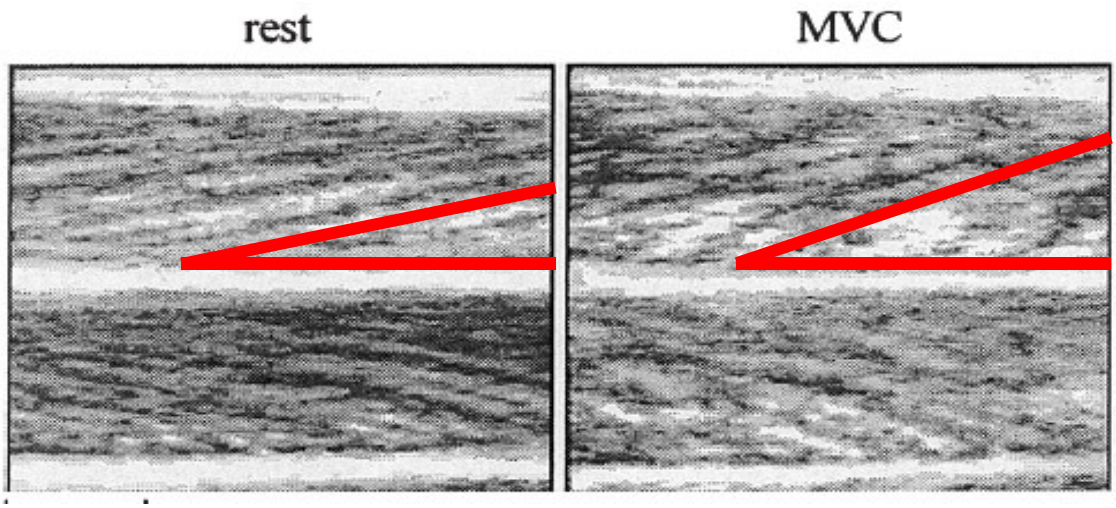

Figure 7. Changes in pennation angle in the tibialis anterior from rest (left) to maximum (right). Modified from (287).

2.2. Physiology

The development of muscle tension involves many factors contributing to the contractile components. These factors relate to the length-tension relationship of the contractile (muscle) and non-contractile (tendons, aponeurosis) fibers, the force-velocity relationship and neuromuscular activation (motor unit recruitment and rate coding) (3, 7, 82, 119, 166, 234). In general, the development of maximal tension in the muscle-tendon complex is achieved while actin and myosin (contractile muscle proteins) are at an optimal distance from one another to maximize cross-bridge attachment/cycling.

2.2.1. Nerve Supplies

In the lower limb, the deep fibular nerve descends from the common fibular bifurcation towards the ankle where it is divided into medial and lateral branches. Prior to reaching the ankle, the deep fibular or common peroneal nerve innervates the anterior compartments of the leg musculature, namely the tibialis anterior. There are approximately 50 innervations of the motor nerve branches in the tibialis anterior, of which approximately 90% innervate the proximal one-third (348). As the primary muscle for ankle dorsiflexion, the tibialis anterior has many innervation sites that can become clustered. These clusters of dense populations of motoneuron synapses are referred to as muscle motor points (Figure 8). The motor point can be identified from superficial stimulation of the musculature where the lowest stimulus intensity evokes a response (47). Although all muscles in the human body contain at least one motor point, the tibialis anterior has had up to five identified (104, 247, 304, 305, 349) with the strongest being located in the proximal one-third of the muscle, near the cluster of nerve innervations.

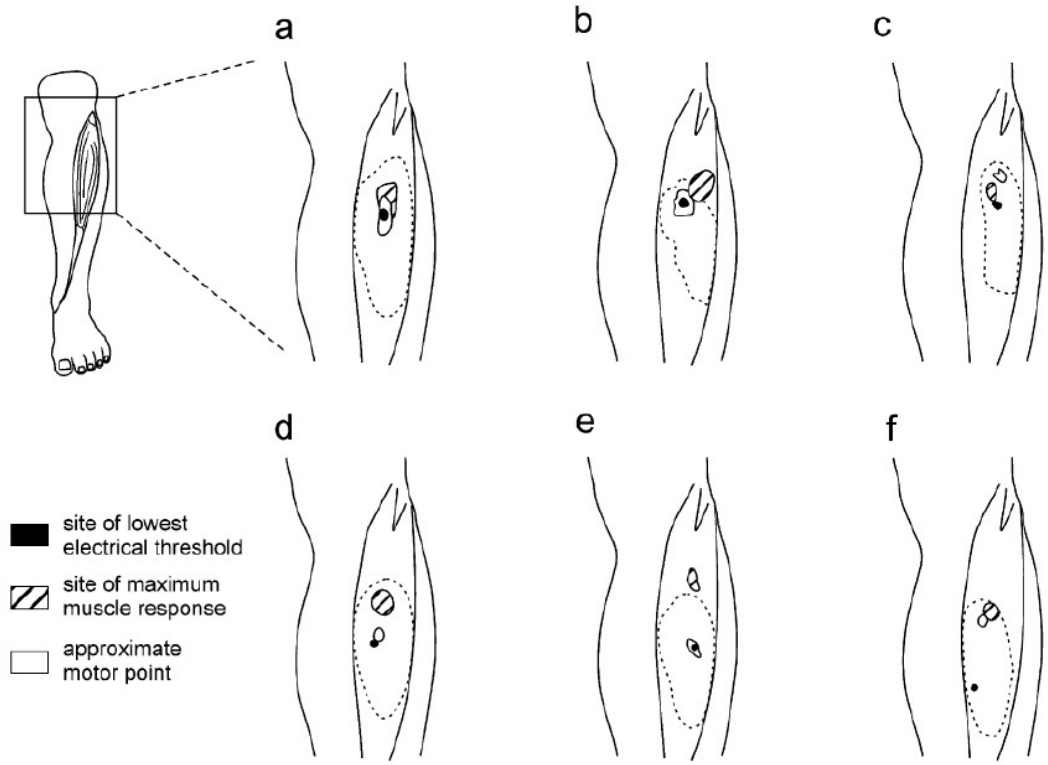

Figure 8. Tibialis anterior motor point locations. This illustration symbolizes the relationship between the regions of lowest electrical discharge, maximum muscle response, and the approximate motor point. The predominant motor point was generally fopunbd in the upper third of the tibialis anterior (47).

2.2.2. Tibialis Anterior Muscle Fiber Type

Skeletal muscle fibers are subdivided into two basic sub-units based on their metabolic and contractile characteristics: Type I and Type II fibers. Type I fibers are also known as the slow twitch fibers due to their relatively slower contraction and relaxation times compared to Type II. Type I fibers can utilize oxidative means of adenosine-triphosphate (ATP) production, thus being more fatigue resistant and can be recruited at lower force levels. Type II fibers perform in quite the opposite manner. Type II have a faster cross-bridge cycling (contraction and relaxation speed), rely more on anaerobic ATP production which makes them more fatigable and have a higher threshold for recruitment (46, 82, 208, 234, 370).

The tibialis anterior is composed of 79.0 ± 1.6% Type I fibres, while Type II fibres make up 28% of the total cross-sectional area (326). According to reports by Feinstein and colleagues (137) and Van Cutsem and associates (398), the tibialis anterior houses approximately 445-550 motor units. In contrast, McNeil and colleagues (300) reported in young men the motor unit number to be closer to 150. The discrepancy in motor unit number may be a result of measurement techniques (statistical vs. motor unit number estimation).

2.3. Elbow Joint

The elbow joint is made up of articulations between the radius, ulna and humerus (269). At the elbow, flexion and extension takes place at the ulnahumeral hinge joint (5, 142, 417), while the pivoting joints, proximal radioulnar and radiohumeral joints, allow supination and pronation (380) (Figure 9). Stability of the elbow joint is a result of both

musculature and passive soft tissue stabilizers. Two key stabilizers are the medial and lateral collateral ligaments (142).

2.3.1. Elbow Joint Biomechanics

The primary function of the elbow is to position the hand in space through flexion or extension while acting as a fulcrum for the forearm (5, 142). The range of motion of the elbow joint required to complete activities of daily living is between 0° and 145° from extension to flexion in the average male, while females may have up to 20° of hyperextension. Rotation of the forearm at the elbow joint ranges from 90° supination to 80° pronation (6, 42, 318, 319).

2.3.2. Biceps Brachii Origin and Insertion

Moving from distal to proximal, the biceps brachii is bifurcated into a long and short head (28, 409). The long head originates from a tendon attached to the supraglenoid tubercle of the scapula, inserting with a distal tendon into the bicipital tuberosity of the radius (142, 172). The short head of the biceps originates from the tendon at the coracoid process of the scapula and inserts via aponeurosis into both the forearm fascia and ulna (142, 362). In some cases the two heads will fuse and both insert on the radial tuberosity by the biceps tendon and by the aponeurosis into the forearm fascia and ulna (Figure 10) (14, 89, 362).

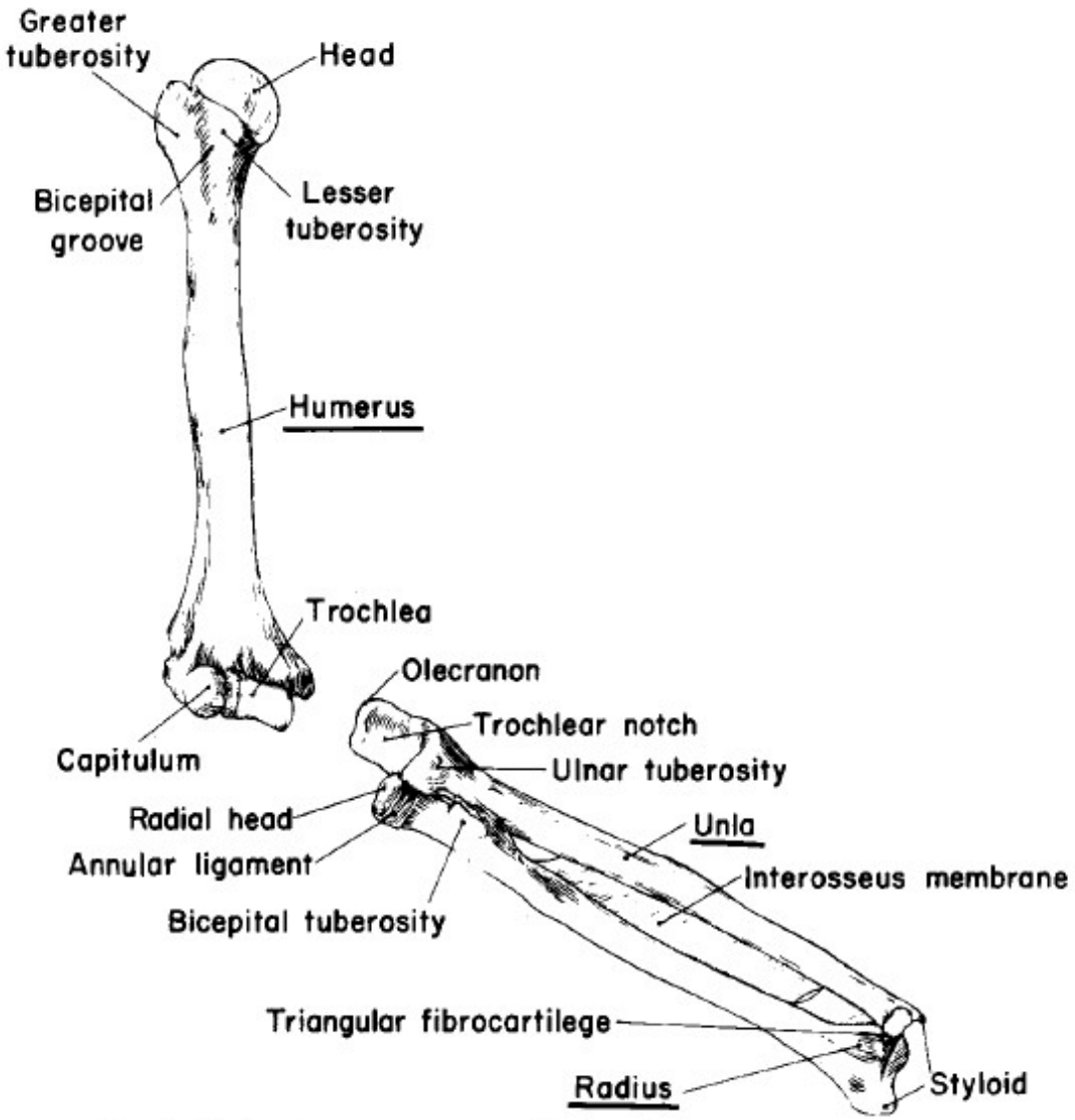

Fig. 1 Skeletal components constituting the human elbow joint with important bony landmarks indicated

Figure 9 Skeletal anatomy of the arm (68)

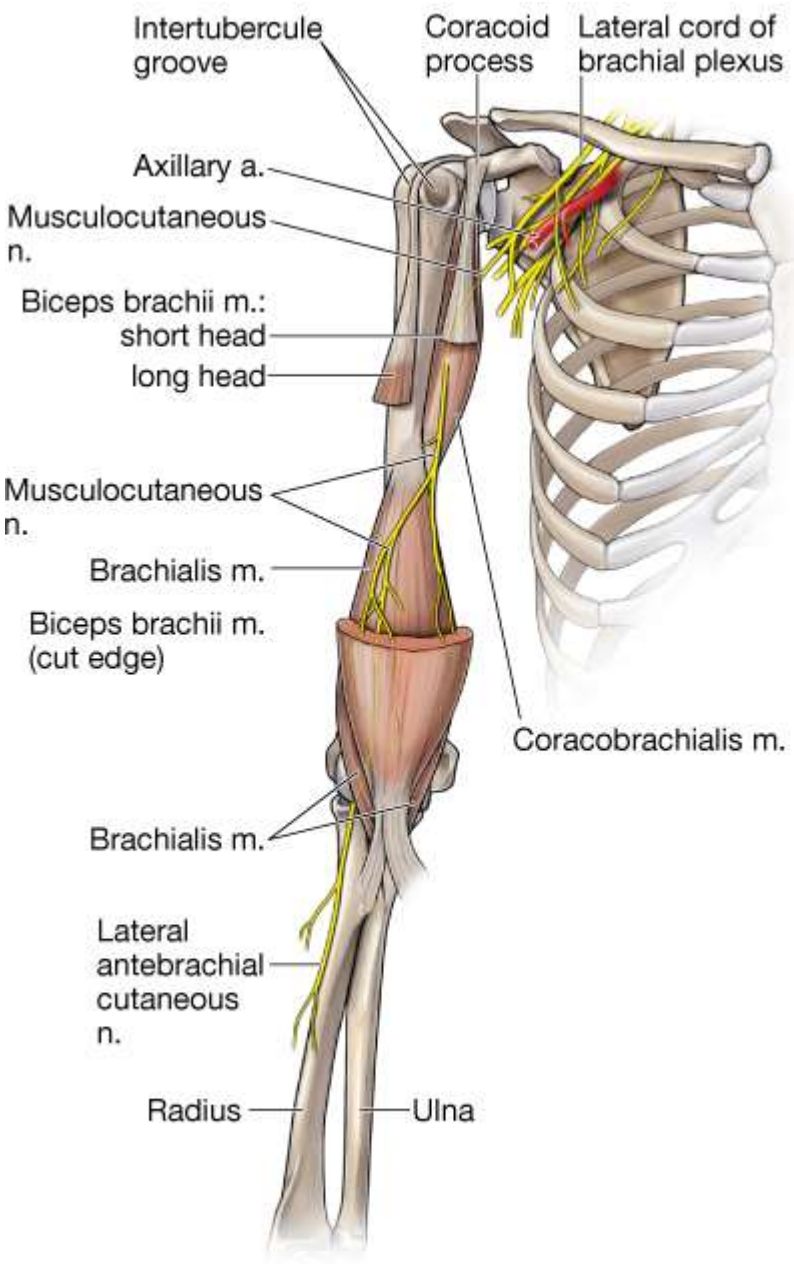

Figure 10. Shoulder and elbow joint with the origins and insertions of the biceps brachii (118).

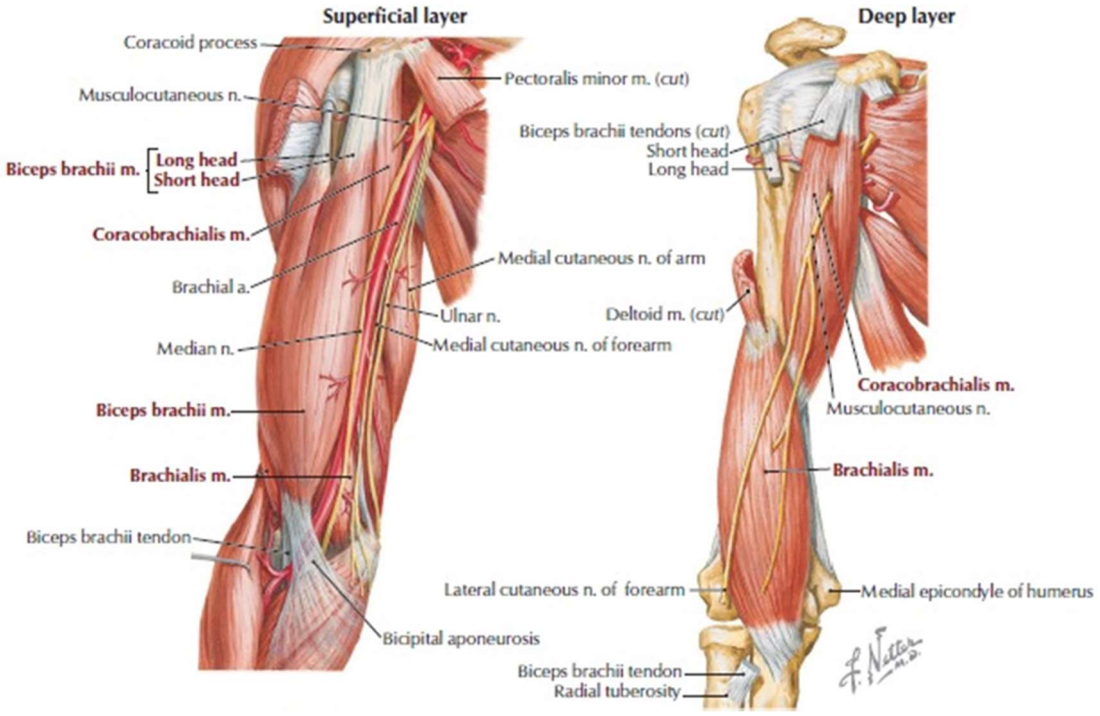

Figure 11. Anterior compartment of the arm muscles and nerves (184)

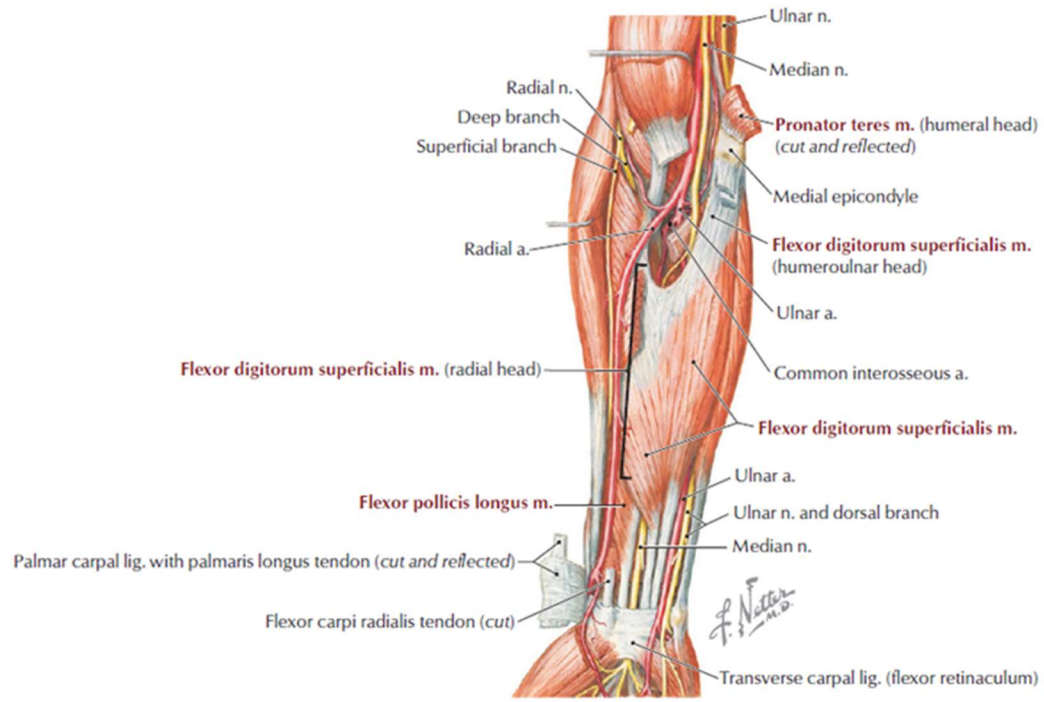

Figure 12. Anterior compartment forearm muscles and nerves (184)

2.3.3. Biceps Brachii Moment Arm and Architecture

The biceps brachii is a fusiform muscle and runs at or nearly parallel to the tendons line of action (92). The moment arm, measured from a line passing through the inferior medial epicondyle and center of the lateral epicondyle (260), and its insertion on the radial tuberosity (321), is between 4.2-5.4 cm (121, 236). As elbow flexion moves towards the angle where peak tension occurs (90° – 100°) the moment arm can increase by 5-7 mm when the forearm is supinated (322). The muscle fibers of the biceps brachii are arranged in parallel to the axis of the upper arm (217) and therefore have a pennation angle, in most cases, equal to zero (236).

2.3.4. Elbow Joint Kinesiological Actions

There are 26 muscles crossing the elbow joint and are divided into four main groups: (1) elbow extensors; (2) wrist and finger extensors and supinators; (3) flexor-pronator group; and (4) elbow flexors (142). Extension of the elbow joint is primarily accomplished by contracting the triceps, while flexion is a result of the contraction of the biceps brachii, brachialis, brachioradialis and the pronator teres (5). As a primary flexor of the elbow joint the biceps brachii accounts for approximately 50% of elbow flexion force (236), especially when the forearm is supinated (141). Rotation of the forearm during supination, is controlled by both the biceps brachii and other more minor supinators (17) (Figures 10-12).

2.3.5 Biceps Brachii Nerve Supply

The biceps brachii is innervated by a single branch of the musculocutaneous nerve (Figure 11), which may bifurcate into both the long and short heads prior to further

branching to supply each head's compartments (69, 362). The motor points for the biceps brachii can be found in a line running from medial to lateral midway between the muscle origin and insertion (10, 12, 295).

2.3.6 Biceps Brachii Muscle Fiber Type and Motor Unit Estimates

The composition of fiber type of the biceps brachii has been reported to be between 50–60% Type II fibers in males (246, 308) and approximately 57% Type II fibers in females compared to Type I fibers (308). Males are estimated to have approximately 126 motor units and females have approximately 110 motor units (308), however, conflicting reports have estimated the numbers in either sex to be closer to 300 with a range of 191-538 (25, 115). This discrepancy may be due to methodological differences.

2.4 Electromyography

2.4.1 Origin of the Electromyographic Signal

Electromyography (EMG) represents the physiological processes that cause muscles to generate force and complete movements that allow us to interact with the world around us (99). The initiation of a muscle contraction is the result of an initiation of an action potential (AP) from the grey matter in the spinal cord being propagated down a motoneuron (MN). This AP travels down the axon of the MN, activating all of its branches synapsing with their associated muscle fibers within a given motor unit (338). A motor unit is defined as an alpha-motoneuron and all of the muscle fibers it innervates (77). The neurotransmitter acetylcholine is released at the synapse (neuromuscular junction) to initiate depolarization in the post-synaptic muscle membrane. The

depolarization then propagates bi-directionally away from the neuromuscular junction (innervation site) (303, 304). These innervation sites on the muscle have a tendency to cluster in a region known as the innervation zone near the muscle belly (353). Electrochemical events associated with the generation of muscle action potentials and their propagation along the membrane result in a voltage change that is detected by the recording electrode, whether it is placed on the skin surface or inside the muscle via a needle or wire (Figures 13 & 14) (17, 94, 226). The summation of the myoelectric potentials are the building blocks of the EMG interference pattern (182). The interference pattern generated from the myoelectric signal can be influenced by both extrinsic and intrinsic factors.

2.4.1.1 Frequency

Fourier analysis of the myoelectric signal is one of the main analysis methods. In its most basic form, it determines the combination of sinusoids with specific amplitudes and frequencies that comprise the myoelectric signal. The result is an amplitude-frequency distribution curve that reflects the underlying generation of the myoelectric signal and the factors that affect it (Figure 15). The fast Fourier transform (FFT) is the most widely used digital implantation (117, 232), from which the mean power frequency (MNF) and median power frequency (MDF) may be calculated (275, 306, 331). These measures characterize the amount of signal energy at a particular frequency, and they are used in the same way as traditional measures of the frequency-distribution curve in statistics (185, 323).

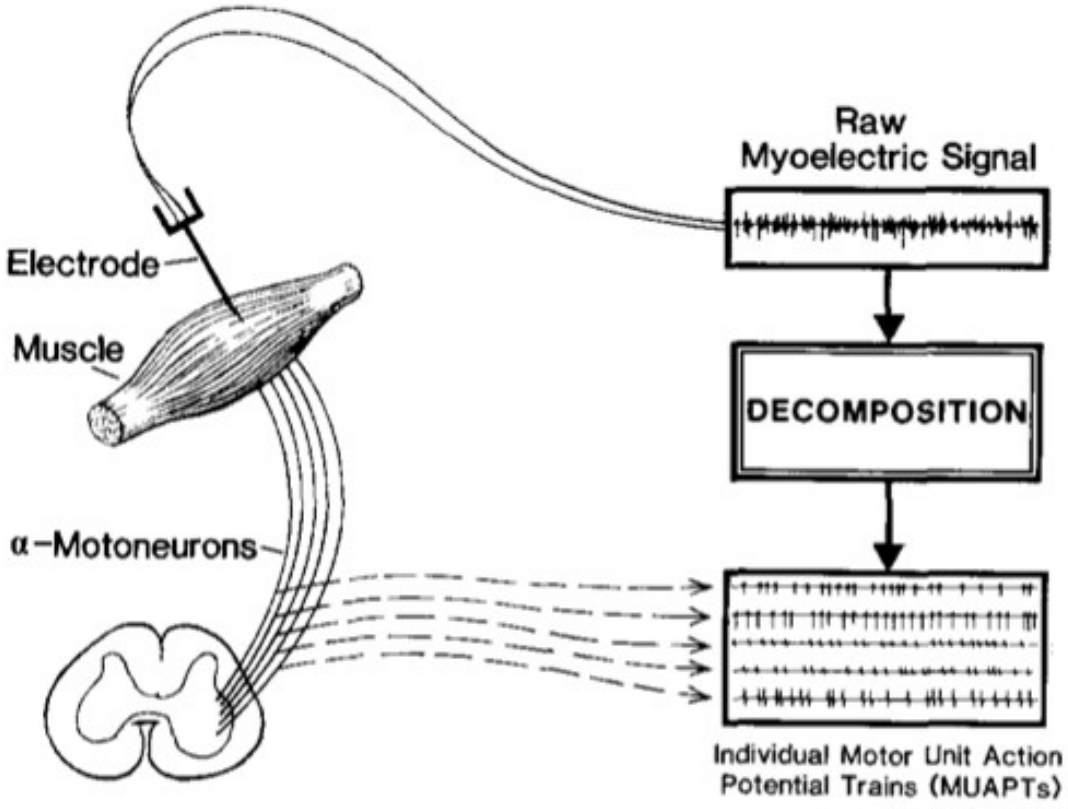

Figure 13. Schematic of the myoelectric signal from the ventral horn through the propagation from the alpha-motoneuron to the muscle. The intramuscular electrode is inserted into the muscle where it records the myoelectric interference pattern from inside the muscle. Decomposition of the interference pattern produces individual motor unit action potential trains (101).

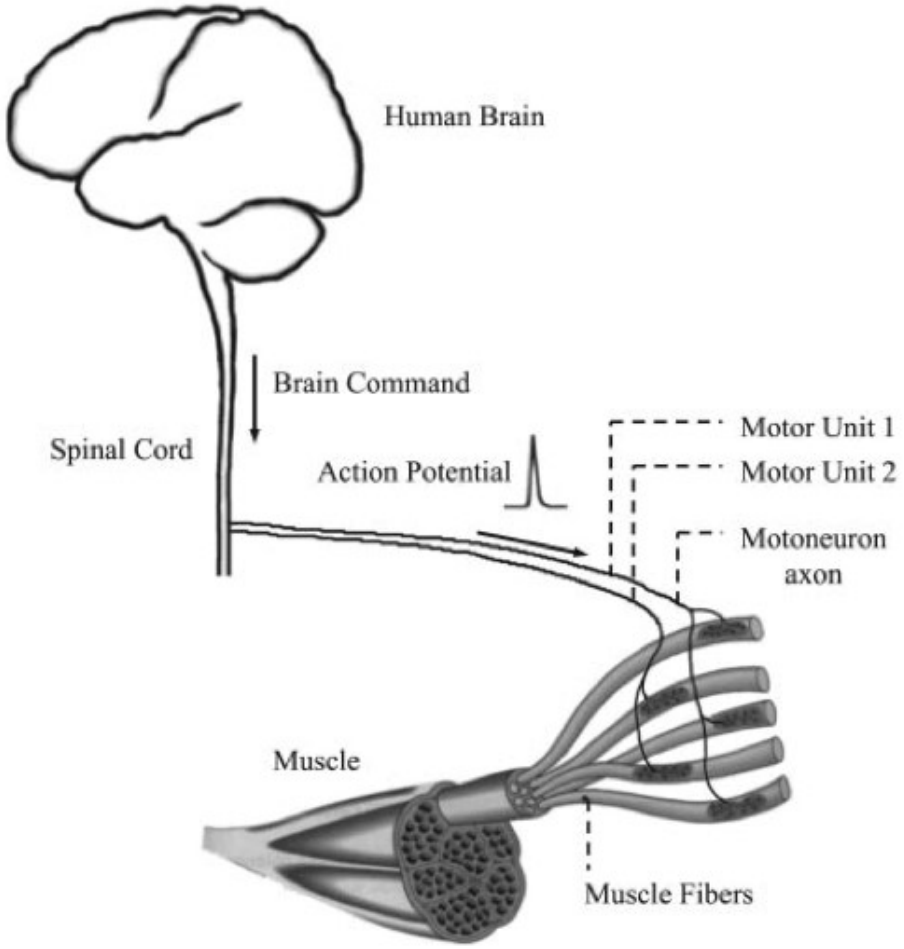

Figure 14. Schematic of the neuromuscular system. Starting from the voluntary neural command from the brain, where the action potential will propagate from the spinal cord along the motoneuron to the muscle. This motoneuron and all the muscle fibers it innervates represents one motor unit (276).

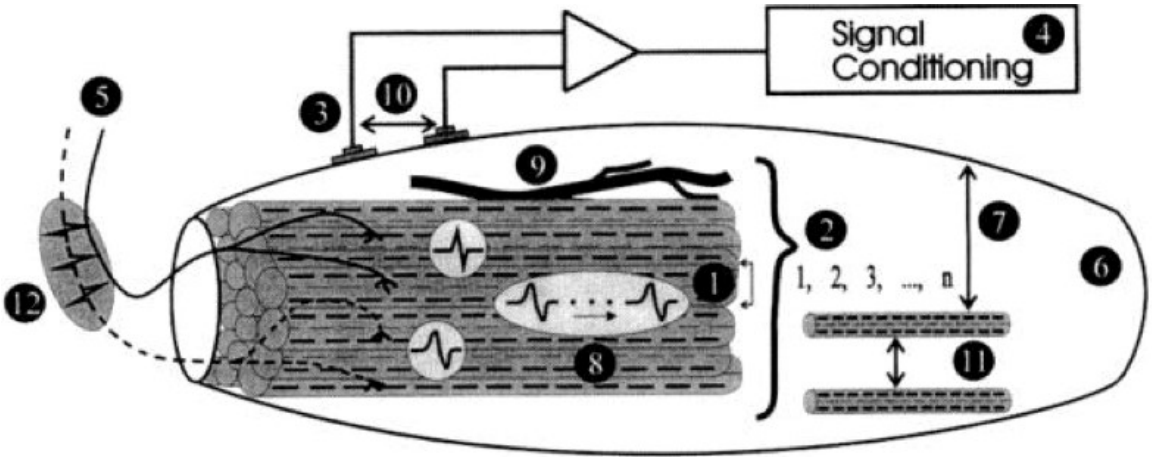

Figure 15. Some factors that influence the myoelectric signal: (1) muscle fiber diameter; (2) number of muscle fibers; (3) skin impedance; (4) signal conditioning; (5) number of active motor units; (6) tissue; (7) distance from the electrode to the myoelectric signal; (8) muscle fiber conduction velocity; (9) muscle blood flow; (10) inter-electrode distance; (11) muscle fiber type location; (12) motor unit discharge rate (226).

The amplitude-frequency distribution curve (Figures 16 & 17) is also called the power spectral density function, analogous to the probability density function in statistics. The shape of the power spectrum of the myoelectric signal reflects the overall shape of the power spectrum of a single motor unit action potential (97). Thus, anything that affects the shape of the motor unit action potential also affects the shape of the power spectrum for the entire myoelectric signal (383). Physiological factors include muscle fiber diameter, conduction velocity and muscle length, to name a few (29, 53, 216, 264, 274). Non-physiological factors relate to the specific type of electrode that is used and its configuration (mono- or multipolar) (54, 99, 104, 132, 274, 339, 347). Motor unit recruitment and rate-coding has been shown to affect the shape of the myoelectric power spectrum and its statistics (13, 106, 210, 232, 254, 267, 307, 405).

As higher threshold motor units are recruited, since the shape of the motor unit action potential (amplitude and duration) differs markedly from low threshold motor units, there will be a narrowing and shift to the right of the entire power-spectrum (32, 33, 72, 316). The same is true for muscle fiber conduction velocity. While related to muscle fiber diameter, muscle fiber conduction velocity can change independently due to fatigue (29, 165, 324). Muscle fatigue is associated with a decrease in velocity of an action potential propagation, increasing the length of time an action potential is under the electrode, widening the waveforms, and the result is spectral compression (53, 305, 377). In contrast, increases in motor unit discharge rate or synchronization are isolated to the 10 to 40 Hz frequency range (225). While these findings have been demonstrated for indwelling EMG, similar changes in the surface EMG signal are more equivocal (54, 58,

275, 282). In either case, it is impossible to link specific changes in MNF and MDF to only one probable mechanism (232, 305).

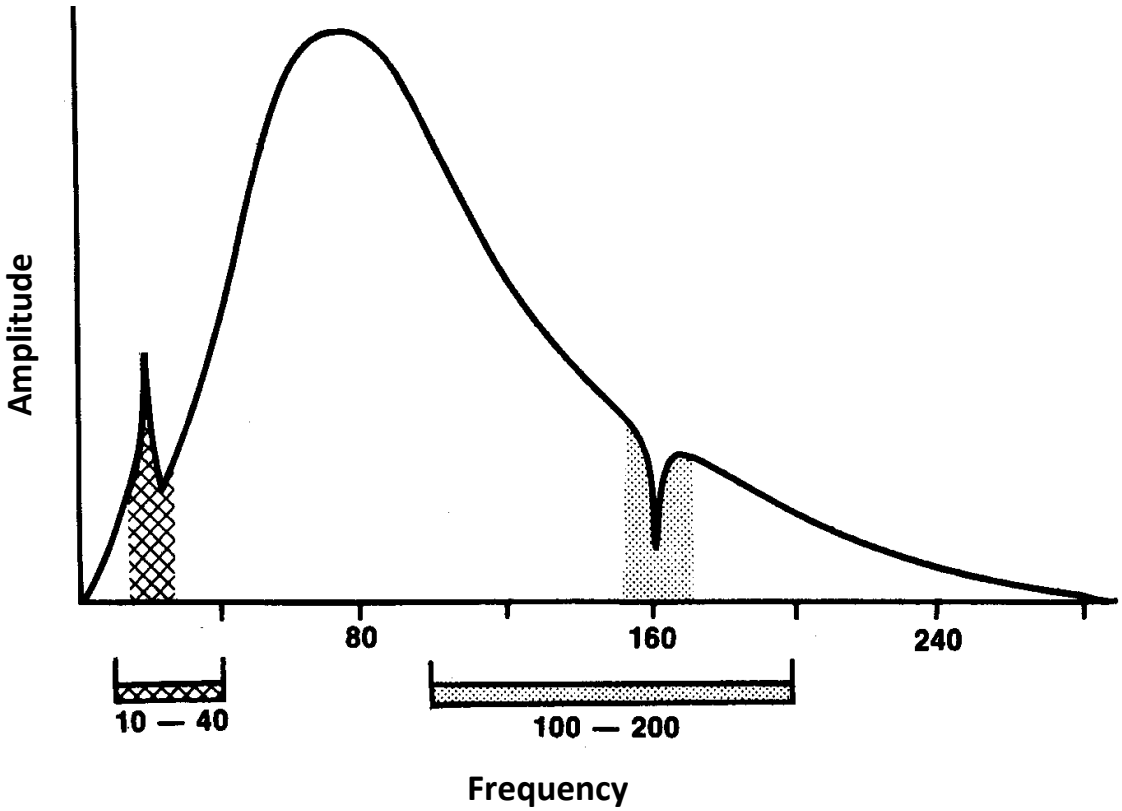

Figure 16: Diagrammatic representation of the frequency spectrum of the differentially detected EMG signal with a graphic representation of the modifications which occur as a function of time and force contraction. The shape has been purposely exaggerated so as to accentuate interesting segments. The peak in the low frequency components is associated with the discharge rates of motor units; the dip in the high frequency components is associated with the conduction velocity along the muscle fibers. The bars on the frequency axis indicate the range over which the peaks and dips may occur. Modified from (17).

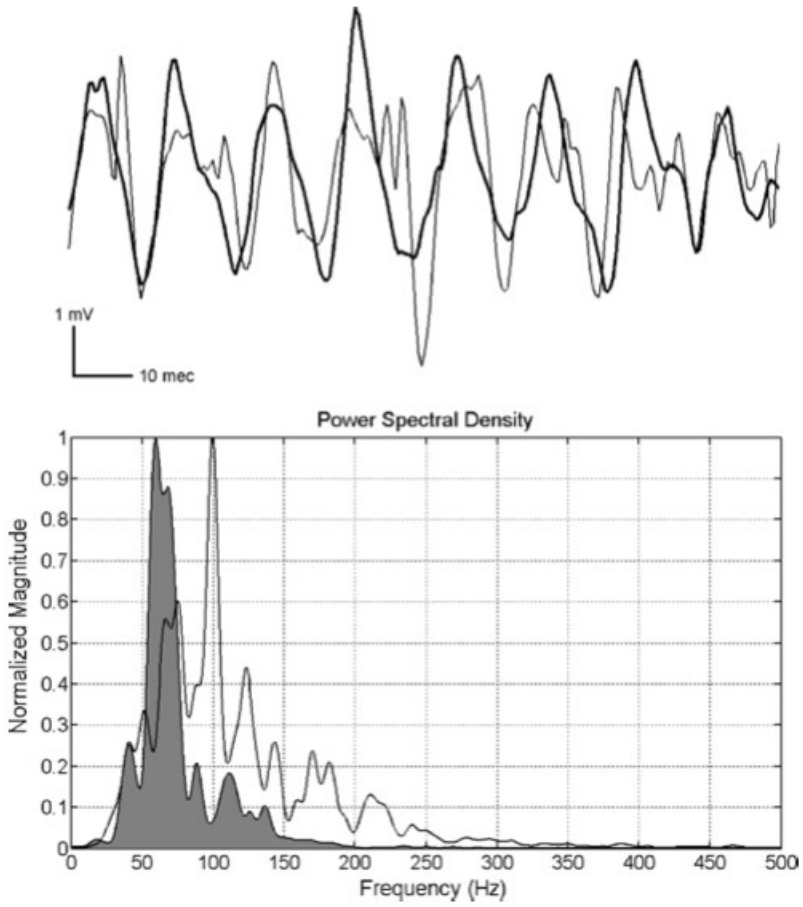

Figure 17. The top portion is the amplitude-frequency distribution curve for both monopolar (thick line) and bipolar (thin line) electrode configuration. The bottom is the power spectra for both the monopolar (shaded) and bipolar (un-shaded) sEMG recordings (151).

2.4.1.2 Amplitude

The amplitude of the myoelectric signal is affected by the same physiological and non-physiological factors as observed for the frequency content for the signal. Much attention has been given to the relationship between the amplitude of the myoelectric signal and force, which has applications in biomechanics (94, 131, 199, 232, 237, 266, 414). The amplitude of EMG activity has also been used as a measure of neural drive to the muscle, but this is subject to debate (131). The root-mean-square (RMS) amplitude and mean-amplitude-value (MAV) are the primary measures of the amplitude of the myoelectric signal (305).

Both linear and non-linear relationships have been observed between isometric force and the amplitude of the EMG signal (414). Waveform cancellation is cited as one of the main non-physiological factors affecting the relationship between EMG amplitude and force, as well as motor unit variables (94, 113, 237, 418). Cancellation poses a problem because additional motor units may be recruited during the gradation of force, yet the superposition of positive and negative phases of muscle action potentials could result in little to no increase in EMG amplitude (94, 237). Signal cancellation affects both intramuscular and surface recordings of the myoelectric signal, and can potentially reduce the true signal amplitude by more than 50% during voluntary contractions (94).

Recent modeling and simulation work by Keenan and colleagues (237) demonstrated a potential loss in EMG signal amplitude of almost 62%. This means that amplitude measures underestimate changes in peripheral neural drive, which affects both recruitment and rate-coding (94, 113, 132, 237). During fatigue, the slowing of muscle

fiber conduction velocity causes an increased motor unit action potential duration (377). As a result, there is an increased probability of overlap between motor unit action potentials, which could lead to greater cancellation or increased summation (147). The controversy is due to the superposition of positive and negative waveforms, across motor unit action potential trains. For example, when two positive or negative phases overlap, signal amplitude increases. Signal reduction occurs when a positive and negative phase overlap. In general, the amount of overlap is random, so signal increases do not double, and decreases do not cancel to zero. However, in the case of synchronization, motor units discharge closer in time, so the positive or negative phases overlap to a greater degree, resulting in large increases in amplitude (94, 237, 418).

2.4.2.3 Compartmentalization

Neuromuscular compartments are the subunits within a muscle group that are delineated by both muscle architecture and nerve branching patterns (362). According to English (123) neuromuscular compartments exhibit task specific behaviours. Segal (362) studied compartmentalization by measuring the branching of musculocutaneous nerves in the perfused human biceps brachii. The goal of the study was to follow the nerves in cadavers until they could no longer be seen from the nerve origin prior to branching to its insertion into the muscle compartments. The observations from the analysis showed that a single branch may bifurcate (2/3 of specimens) to supply the long and short heads separately or may originate from two separate musculocutaneous nerves (1/3 of specimens). It is believed that different compartments within the active muscle will have different functional tasks. Additionally, within the different compartments, further branching into compartmentalized subunits were found more often either proximal or

distal to the muscle belly. Separate portions of muscles will have unique muscle fiber type distributions that are innervated by distinctly different motor nerves. Subsequently, the motors nerves can have different activation patterns leading to different usage, which is task specific, despite being located within the same muscle (273).

Compartmentalization of the musculature during specific tasks can lead to phenomena where within the motor unit there is a silent area. If this area falls outside of the pick-up volume of an intramuscular electrode no desirable myoelectric energy may be recorded (378). This was also observed by Lynn and associates (283) who looked at different radial distances away from the recording electrode. Due to factors such as volume conduction and electrode pick-up volume it was discovered that not all active motor units are represented within the interference pattern. Due to the existence of the silent area and issues with volume conduction and electrode pick-up volume Christensen and Fuglsang-Frederiksen (71) recommend approximately 10 recording sites in order to ensure the recording of all the active motor units within a muscle. Furthermore, recording at different depths may be beneficial when using the intramuscular electrode because different threshold motor units have been reported to be deeper (lower threshold MUs) compared to more superficially (higher threshold MUs) (248). This could also lead to a recording bias when using surface electrodes as the average pick-up volume is between 15-30 mm below the recording electrode which could lead to an interference pattern dominated by high threshold (short duration, high amplitude), larger fiber diameter, fast twitch motor units (378).

Previous literature has indicated that compartmentalization of motor unit activation may be a result of different factors, namely joint angle or a given task (34, 109, 162, 180,

395, 399). Loeb (279) suggested that the central nervous system activates task groups, as opposed to individual muscles, consisting of different regions of involved muscles to linearize the motor output. Investigations of the flexors and extensors of the elbow joint has revealed a distribution of subpopulations of motor units in different muscle regions that are activated based on task (179, 180).

ter Haar Romeny and colleagues (180) reported that a given task dictates what extent of the motor unit territory of the biceps brachii is either activated, to support the task, or deactivated, if it is not mechanically advantageous. This task specific activation may originate from multifunctional muscle afferents of synergist muscle groups or cortico-spinal motoneurons. Multifunctional motoneurons innervate multiple involved muscles or motor unit pools, thus maximizing task efficiency (34, 108, 130, 146, 367). Fetz and associates (138) referred to this co-activation of more than one muscle group as being a 'motor field'. Therefore, in the case of multifunctional muscles, such as the biceps brachii, the recruitment of the motor unit pool may be based on the changes in task, joint angle or direction of the forces being exerted. When controlling for variations in task, joint angle and force vector, as is the case during isometric contractions, muscle group and motor unit territory activation become reproducible (180).

2.5 Motor Unit Control of Force Gradation

Motor control of force gradation begins with an alpha motoneuron originating in the grey matter at the anterior horn of the spinal cord. This motoneuron will innervate a number of local muscle fibers from terminal axonal branches, which comprise motor units (99). The motor unit is the smallest element of the neuromuscular system (17, 77,

364). All of the motoneurons will generate some degree of excitation in almost all of the innervated muscle fibers within the respective motor unit (302). The excitation is propagated as a motor unit action potential, which travels the length of the cell and initiates the excitation of all of the muscle fibers. Individual cell cross-bridge coupling and subsequent sarcomere shortening is the result, and is commonly referred to as a muscle fiber contraction (17). The gradation of muscle force occurs by recruiting more motor units and increasing the discharge frequency of the already active motor units (2).

2.5.1 Henneman's Size Principle

At the onset of force gradation, the motoneurons that are smallest are activated first and as force gradation continues to increase, progressively larger motoneurons are activated/recruited (194). This orderly recruitment of motoneurons was defined by Henneman and colleagues (196) as the 'size principle'. This principle is based on the order of motoneuron recruitment during a stretch reflex (196), which is an extension of the observation that motoneuron excitability is inversely proportional to the size of the motoneuron (193). When referring to motoneuron size, it has been reported that the motoneuron axonal diameter is proportional to the surface area of both its soma and dendrite (16, 60, 79, 241). Therefore, a larger motoneuron with a larger axonal diameter will have a greater excitability threshold (196). Subsequently, this defines the order of motoneuron recruitment by size (18, 79, 196). Enoka and Stuart (128), demonstrated that motorneuron size is the basis for the orderly recruitment of motor units during ramped increases in muscle tension. They suggested that there is a strong relationship between the probability of a motoneuron to discharge and its size. Force gradation is controlled by the orderly recruitment of motoneurons and their associated motor units based on size.

Increases in tension are accomplished by the orderly recruitment of new, progressively larger motor units that have larger tension outputs and motoneuron excitability thresholds (127, 177, 195, 311, 335). As a result, during force gradation smaller motoneurons are more likely to discharge before the larger motoneurons (98).

2.5.1.1 Motor Unit Recruitment

Muscles are not controlled by a single motoneuron. To achieve force gradation a population of motoneurons must be recruited throughout a range of excitation. The control over whether or not a motor unit will become active will depend largely on the membranes level of excitability to achieve the appropriate level of force. The recruitment of a motoneuron depends on its membrane threshold, which is related to its size (193). The level of excitability in the motoneuron pool is determined by its descending (central nervous system) drive and proprioceptive feedback (11, 281, 294, 366, 389). The excitability level in the motoneuron pool will bring motoneurons closer to or farther away from their threshold for discharging (196). Neural drive increases or decreases commensurately with the force requirement for the task. Recruitment threshold is expressed as a percentage of a maximal voluntary contraction (MVC) and monitored during ramp contractions because of this commensurate output of neural drive with respect to force (78, 109, 144, 312, 313).

The motor unit pool is completely recruited (saturated) well before maximum force output of the muscle has been achieved. The percentage of maximum force at which motor unit recruitment is saturated has been termed “the recruitment range” (261). The percentage of maximum force at which new motor unit recruitment ceases, differs

between large and small muscles. Kukulka and Clamann (261) demonstrated that muscles (FDI) cease recruiting new motor units at 50% MVC while larger muscles (biceps) continue recruitment until approximately 80 to 90% MVC. Our laboratory and others have established that the recruitment range for the tibialis anterior is up to 90% MVC (109, 398). The biceps brachii is known to have a similar recruitment range (261).

In accordance with the size principle, smaller axonal diameter motoneurons with lower tension output and with proportionately lower excitatory thresholds are recruited at the onset of tension development and in short succession to initiate an increase in tension relative to the excitation of the motorneuron. It is not until later tension development that larger axonal diameter motoneurons, with higher twitch tension and with proportionately higher thresholds are recruited (311). These intrinsic excitatory thresholds are dependent on the amount of membrane depolarization required (243).

2.5.1.2 Rate Coding

After motor unit recruitment has saturated, rate coding is required to continue muscle tension increases to achieve maximum force (228). Therefore, rate coding is responsible for increasing force beyond the muscles recruitment range (102, 189, 313). Rate coding is the modulation of the motor unit discharge rate as part of force gradation (45). The motor unit discharge rate is the number of 'twitches' per second an individual muscle fiber contributes to force output (99). Muscle fiber twitches are initiated by the propagation of the action potential along the motoneuron to the neuromuscular junction. This excitatory potential is transported across the synapse via acetylcholine where the resulting action potential will travel along the muscle fiber and activate the contractile

proteins. As the rate of action potential discharges increase so does the number of muscle fiber twitches resulting in a summation as they merge upon one another. Discharge rate increases muscle tension due to the summation of twitches, until they fuse to achieve maximal force output (393).

The modulation of the motor unit discharge rate is more complex than the recruitment order of motor units (196, 228). Motor units are discharged within a range based on a number of complex factors that may be a function of: (1) synaptic drive; (2) membrane threshold; (3) motoneuron activation history; (4) muscle size; (5) function; or (6) population of motor units (45, 139, 228). Motor unit discharge rate is also a function of the diameter of the motoneuron's motor axon. Differences in the diameter of the motoneuron will influence the amount of input resistance and subsequently the conduction velocity of the motoneuron action potential (79, 139).

Burke and colleagues (62) differentiated motor units as being either 'S' or 'slow type' (slow twitch, fatigue resistant) and 'F' or a 'fast type'. The fast type was further differentiated based on sensitivity to fatigue and are referred to as: 'FF', fast twitch fatigue-sensitive; 'FI', fast twitch intermediate fatigue resistance and 'FR', fast twitch fatigue-resistant. Greater input resistance is observed in smaller diameter motoneurons (S), which subsequently slows the propagation of the motoneuron action potential towards the active muscle. In contrast, the larger the diameter of a motoneuron axon (F), input resistance is lessened, resulting in a greater conduction velocity. A greater conduction velocity increases the rate of motoneuron action potential transmission, allowing for higher motor unit discharge rates (139, 213, 243).

The range in which a motor unit can be discharged is referred to as its discharge range. Minimum discharge rates of motor units during voluntary contractions have been recorded between 5 and 7 Hz (258, 376, 398). As force gradation increases to moderate force levels (~35% MVC), discharge rates for low threshold, first recruited, motor units have been reported as greater than higher threshold, later recruited, motor units (102, 315, 386, 387). Interestingly, during maximal effort contractions there is discrepancy. Some have shown that the low-threshold motoneurons continue to discharge at higher rates (102, 230), while contradictory research suggests at maximal contraction levels the later-activated, higher threshold motoneurons discharge at higher rates (178, 315). Variations in recruitment thresholds and discharge rates may be a result of the function and anatomy of the muscles being investigated (102). It is possible that smaller muscles, such as the first dorsal interosseous, require greater rate coding to produce a smooth increase in tension development as they are used in fine motor control (313). Whereas a larger muscle, such as the deltoid, is used in powerful movements. As the deltoid has a greater number of motor units, it would be beneficial during force gradation to activate as many motor units as possible to achieve maximal force. Subsequently, utilizing rate coding only as a secondary means of achieving maximal tension (102). In both the biceps brachii and the tibialis anterior the maximal discharge rates, regardless of motoneuron recruitment threshold, are approximately 31 Hz and 33 Hz respectively (25, 398). The dichotomy in the research regarding the discharge rate of low- versus high-threshold motor units is still up for debate and will be highlighted in the next section when discussing the ‘onion skin’ phenomenon.

As for the affect motor unit discharge rate has on force gradation, it has been extensively reported that there is a high correlation between motor unit discharge rates and force output during isometric contractions in multiple muscle groups (102, 268, 313–315, 340). The achievement of maximal muscle tension outside of the motor unit recruitment range is because of increased motor unit discharge rates, whether in small (102) or large (231) muscle groups. Although a single motor unit action potential may have a minimal effect on force gradation (80, 122), if the time between discharges lessen and demonstrate a uniform inter-pulse interval, the motor unit will have demonstrated an action potential sequence. The sequence is referred to as a motor unit action potential train (99). It is the summation of the twitch forces with short interpulse intervals from the motor unit action potential trains that leads to greater force gradation, eventually resulting in maximal force production during fused tetanus (61, 63).

2.5.2 Onion Skin

It has been reported that the discharge rates of earlier recruited motor units are higher than those that are recruited later, specifically during isometric contractions (99, 102, 315, 340, 386). De Luca and Erim (98) made the analogy that this hierarchical, layered response is like that of the ‘onion skin’ and is the result of the common drive of the motoneuron pool (95). The ‘onion skin’ phenomenon suggests that at any given level of muscle tension, lower threshold motor units will have a higher discharge rate than those of higher threshold motor units (99, 102, 315, 340, 364, 387, 407) (See figures below).

Figures 18 through 20 illustrate the ‘onion skin’ phenomenon. As net excitation increases, more motor units are recruited that have different discharge rates (Figure 18).

Increasing excitation then results in the recruitment of new motor units at slower discharge rates than those previously active (Figure 19). It is not until maximal excitation, that motor unit discharge rates converge closer to one another (Figure 20). However, the first recruited, low threshold motor units continue to discharge at faster rates, compared to the higher threshold, later recruited motor units (98). Interestingly, De Luca and Erim (98) portray the net excitation applied to all active motor units, regardless of fiber type, from the same common excitability or common drive, which will be discussed further.

2.5.1.3 Synchronization

Motor unit synchronization is the simultaneous or near simultaneous (± 10 ms) discharge of two motoneurons that occur more often than by chance (309, 313). The origin of motoneuron discharge synchronization is believed to be from common supraspinal descending pairs of branched cortical presynaptic neurons that synapse on two or more lower motoneurons, thus increasing the probability of simultaneous discharges (134, 244, 298, 361). The magnitude of the synchronization of motor unit pairs is influenced by: (1) the task being performed; (2) the involved muscle group; (3) associated motor units; and (4) previous physical activity (49–51, 134, 359, 360, 364). Synchronization occurs during tasks that require precision (15) and skill training (358), in both small and large muscle groups (140) and following training in different populations (229, 309, 310).

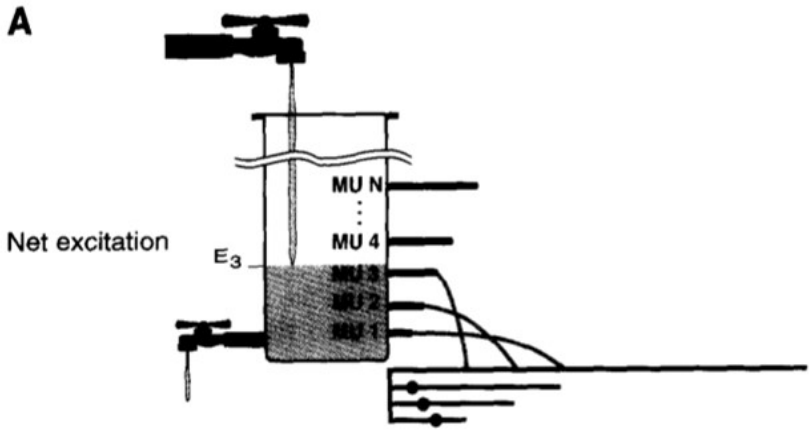

Figure 18. Represents the regulation of motor unit excitation and subsequent recruitment and discharge rate in the modulation of force gradation. The large tap represents the net excitation from the nervous system, while the small tap is the inhibition. 'E_3' represents the excitability required to recruit the first three motor units. The length of the spout represents the initial discharge rate of the motor unit and the circle on the bottom lines represent the discharge rate below which the motor unit cannot discharge. As excitation increases so does the number of active motor units and their discharge rates (98).

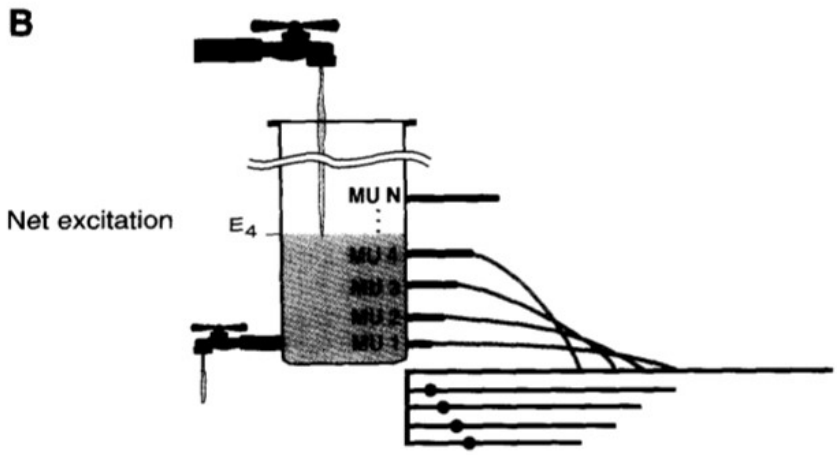

Figure 19. In this, the second figure in the series, as the net excitation increases, so do the number of active motor units and their discharge rates. 'E_4' represents the excitability required to recruit the first four motor units. The length of the spout represents the initial discharge rate of the motor unit and the circle on the bottom lines represent the discharge rate below which the motor unit cannot discharge. Note, the newly recruited motor units, although requiring a greater excitation to discharge, will discharge at lower rates than those previously recruited (98).

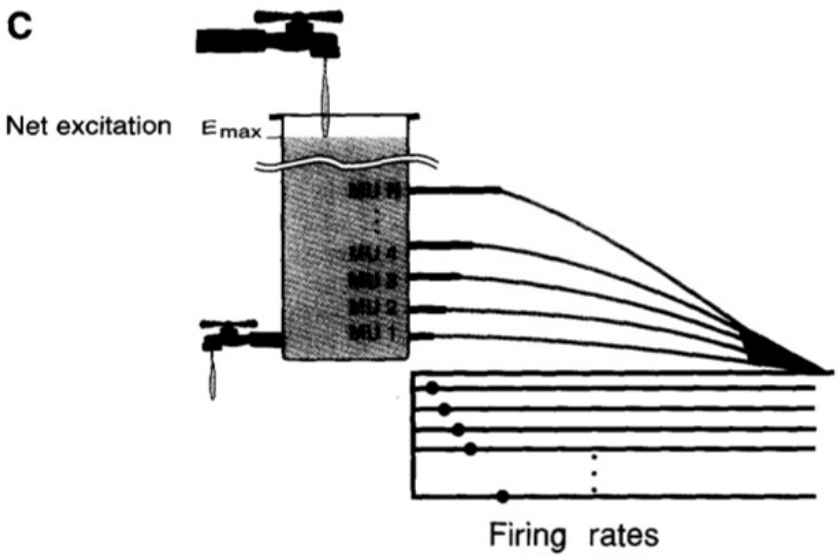

Figure 20. In the final figure in the series, the maximal net excitation initiates the activation of a greater number of motor units. 'E_{max}' represents the excitability required to recruit the remainder of the motor units. The length of the spout represents the initial discharge rate of the motor unit and the circle on the bottom lines represent the discharge rate below which the motor unit cannot discharge. This 'extreme' neural drive leads to all active motor units discharging at nearly the same rate once recruitment is no longer utilized for force gradation (98).

It would be simplistic to assume that synchronous pairs of motor units discharging nearly simultaneously would increase force output. Interestingly, there is discrepancy in the literature regarding this outcome. Some suggested that the synchronization of motor unit discharges is a mechanism to enhance force output (140, 309, 364). Kamen and Roy (229) reported that higher motor unit synchronization at high force levels (>80% MVC) was believed to contribute to the production of larger forces, more so than at lower force levels. Indirect support through modeling and simulation work, suggest that changes in the sEMG signal above 80% MVC is most likely due to synchronization (153). Smaller muscles with fewer motor units would benefit greatly from the activation of pairs of synchronous motor units to efficiently produce force and to enhance tension development at larger levels of force (80, 140). In contrast, there is only limited evidence that larger muscles may utilize motor unit synchronization. Fling and colleagues (140) showed a tendency towards motor unit synchronization in the biceps brachii during isometric contractions, but limited the intensity to 80% MVC.

Contradictory findings have suggested that motor unit synchronization does not enhance force output (80, 346, 418). In a simulation study, Yao and colleagues (418) indicated that an increase in motor unit synchronization resulted in an expected increase in surface EMG amplitude, but did not observe an increase in the average force output. One theory on why motor unit synchronization does not increase force output is its effect on the SEC (series elastic component). The understanding is that the SEC would be able to slacken in-between each synchronous discharge and subsequently need to be taken up by the following discharge (401). This would then negate the effect of the synchronous discharges and leave the average force output unaffected (80, 344). Furthermore, this

would also contribute to the greater force fluctuations observed with synchronization (80, 344, 346, 418). Although motor unit synchronization may not enhance average sustained force output, it may influence higher initial motor unit discharge rates to enhance the rate of tension development during ballistic contractions (397).

Motor unit synchronization may impose its greatest influence on force output by altering force stability. Numerous research groups have indicated that synchronization has an influence on the force profile and its stability (80, 156, 174, 299, 344, 346, 418). Increase in force fluctuations may be a result of either the SEC or an increase in the variability of the motor unit discharge rate associated with synchronization (80, 344, 346). The force fluctuations resulting from the SEC come from the slack in the musculoskeletal complex that inevitably needs to be taken up prior to force production. Motor unit discharges stretch the SEC and subsequently shorten the muscle fiber length. If synchronization results in increasing the interval between motor unit discharges, the elastic components in the musculotendinous complex would allow the muscle and muscle fibers to lengthen. Lengthening of muscle fibers has been shown to reduce the force output. Therefore, successive motor unit discharges, or synchronous pairs of discharges, would be necessary to shorten the slack musculotendous complex, returning the muscle and its fibers to their previous length and force output (344). The addition of either more active motor units or variability in discharge rate could contribute to fluctuation in force seen during synchronization (253). Previous literature has shown that distributing motor unit discharges, opposed to synchronizing them, produces a smoother muscle contraction (346). Motor unit discharge rate variability will be discussed in greater detail below.

2.5.1.4 Doublets

Modulation of the motor unit discharge rate to increase tension development can be accomplished through increased post-synaptic excitation to the muscle fiber, but this is usually limited by maximum discharge capacity (~35-65 Hz) (153, 172). It has been reported that the incidence of doublet discharges can briefly increase the motor unit discharge rate to almost four times (50-200 Hz). A doublet discharge is identified as a short inter discharge interval of two consecutive MU discharges (2.5-20 ms) (204, 257, 371) followed by a longer than normal inter discharge interval (100 ms) (66) They can occur either independently or in a series of recurring events. The incidence of a doublet discharge can lead to a significant increase in twitch force, however it also increases motor unit discharge rate variability (158).

A number of research groups have studied maximal motor unit discharge rates in both small and large muscles. Characteristics that may alter maximal discharge rates seem to be based on six factors: (1) individuals age (73, 83, 90, 245);; (2) muscle shortening velocity (188); (3) muscle fiber and motor unit type (43, 176, 354); (4) the muscles contractile speed (25, 397); (5) size of the muscle (48, 261, 317, 393); and (6) the number of active motor units at maximum muscle tension (72, 230, 313). Larger postural muscles have been reported as having lower maximal motor unit discharge rates, while smaller intrinsic hand muscle have higher rates (48). Large muscles such as the following: (1) biceps brachii; (2) triceps brachii; (3) soleus; (4) tibialis anterior; (5) vastus lateralus and medialus; and (6) rectus femoris have been reported to have average motor unit discharge rates of approximately 26 Hz (25, 83, 188, 219, 271, 285, 350, 354, 394, 398). Smaller muscles such as the following: (1) adductor pollicis; (2) first dorsal

interosseous; (3) extensor digitorum communis; (4) thenar; and (5) big toe extensors have average discharge rates closer to approximately 34 Hz (25, 116, 175, 230, 315, 392).

Bellemare and colleagues (25) recorded motor unit discharge rates and contraction speed from evoked supramaximal stimulations in three human muscles. Recording from 300 motor units in the biceps brachii resulted in an average discharge rate of 31.1 ± 10.1 Hz. Motor unit discharge rates were found to vary greatly between muscle groups and subjects. In general, smaller intrinsic hand muscles and proximal muscles discharge at higher rates, compared to distally located postural muscles (25). Similarly, discharge rates may also be related to the recruitment range of the muscle. Smaller muscles recruit all their motor units by approximately 50% MVC and then rely on rate coding to achieve maximal force (313, 381), while larger muscle recruitment of new motor units "almost" throughout the entire range of its force output (109, 261). It was also determined that the speed of contraction, determined by contraction time and half relaxation time, is proportional to the excitation frequency required for tetanic force fusion (31, 60). Therefore, once maximal force is achieved further excitation would not result in greater force output (25).

Dalton and associates (90) measured the non-dominant elbow flexors in both old and young subjects during dynamic onset contractions. The data were extracted from a stable five-second portion of the maximum contraction. Discharge rates were close to 42 pps. Using fine wire electrodes inserted into the biceps brachii, Kukulka and Clamann (261) studied motor unit discharge rates during maximal isometric contractions. The biceps brachii was observed to recruit new motor units up to at least 88% of an MVC,

and the maximal motor unit discharge rate 32 pps. However, they did not specify how they reached MVC, whether it was a ramp or dynamic contraction.

Connelly and colleagues (83) compared contractile properties of the tibialis anterior at 30^{o} of plantar flexion during brief, sustained voluntary and evoked dorsiflexion contractions at various force levels in young and old recreationally active individuals. They reported age-related differences at each force level (10%, 25%, 75% and 100% of MVC). The maximal motor unit discharge rates reported in the tibialis anterior of the younger adults was 41.9 ± 8.2 Hz. However, these values are much higher than those reported by others. Research by Christie and Kamen (74) and Van Cutsem and colleagues (398), while studying the tibialis anterior, had participants perform isometric ramped contractions at 20^{o} of plantar flexion over six to ten seconds, while measuring both motor unit recruitment and rate coding. Christie and Kamen (74) reported a maximal motor unit discharge rates of 32.7 ± 6.8 pps while Van Cutsem and associates (398) observed 32.1 ± 10.7 pps. Bigland-Ritchie and associates (31) also studied the tibialis anterior during maximal isometric contractions and reported very similar values. The authors measured motor unit discharge rates at two muscle lengths by manipulating the ankle joint angle (control – 90^{o}; shortened – 75^{o}). Measurement of the motor unit discharge rate was over a ten second voluntary contraction with intramuscular tungsten microelectrodes. No significant difference in motor unit discharge rate between the two muscle lengths (control: 28.2 ± 9.9 Hz; shortened: 27.7 ± 9.8 Hz) was observed. The lack of significant differences despite changes in muscle length, may be explained by the maximal contraction requiring the physiological limits for maximal discharge frequency. It was suggested that later recruited motor units might discharge at lower rates, causing motor

unit pooled averages to be lower, hence skewing the data (31, 401). The differences in discharge rates between those reported by Connelly and colleagues (83) during the brief sustained voluntary contractions, and those of Christie and Kamen (74), Van Cutsem and associates (398) and Bigland-Ritchie and associates (31) may result from differences in population, sample size, decomposition techniques, joint angle or task.

The underlying mechanisms behind the incidence of doublet discharges is dependent upon the intrinsic properties of the motoneuron membrane. This may arise from either an increased period of depolarization or a delay in the depolarization of the action potential. Delayed depolarization may result from an impulse travelling antidromically, following an action potential or increased motoneuronal excitability. It is also believed that doublets may be part of the secondary discharge range of motor units to rapidly adjust for changes in force. Other theories suggest doublets may just be a random variability of the inter discharge interval (IDI) as a result of the malleable properties of the motoneurons (as seen in training) and synaptic branching and not a motor cortex strategy (240, 258, 259).

There is an increased incidence of doublets at the onset of force generation (19, 173, 257, 259, 397). The increased incidence of doublet discharges at the onset of muscle activation can take advantage of the muscles catch-like properties to increase both the peak force achieved and the rate at which the force develops (35). Short IDIs of between 5-10 ms have been shown to be optimal for tension development during the onset of muscle activation (19, 63, 109). The increase in force may arise from increased [Ca^{2+}] in the myoplasm after the first discharge (110). The increase in [Ca^{2+}] is due to the release of more Ca^{2+} from the second discharge, prior to the reuptake of free Ca^{2+} from the

sarcoplasmic reticulum as a result of the first discharge (19, 249). Increased [Ca^{2+}] would activate further conformational changes to the tropomyosin allowing more attachments of cross bridges (167). The second discharge in the doublet pair usually produces a lower twitch force, which suggests that, the sarcolemma is still in relative refractory (135). Biomechanically, the benefits of doublets may be more effective during concentric isometric force gradation as the doublets may be more efficient in dealing with changes in muscle stiffness as it removes any slack in the muscle tendon unit prior to the production of torque about the joint.

Doublets have been recorded in 10-30% of motor units, and represent only a small portion of the total number of discharges (158). Despite the fact that the doublets occur in many motor units during isometric contractions they seem to be more prevalent in high threshold motor units during maximal contractions as force fluctuates or as fatigue sets in. During submaximal isometric contractions, they have been identified as occurring closer to the minimum firing range of small, slow twitch fibers and repetitively during slow rising ramped contractions (19, 66, 173, 257).

2.5.3 Common Drive

De Luca and Mambrito (103) have observed that the firing rates of motor units are not controlled individually. Rather, motor unit discharge rates tend to increase and decrease in unison, even though the different motor units in the pool may be discharging at different rates. The phenomenon has been termed “common drive” and one way in which the nervous system simplified the control of muscle (98, 102, 365). Kamen and De Luca (227) demonstrated that common drive can be assessed through the cross-correlogram of instantaneous motor unit discharge rates with correlation coefficients

greater than 0.77. The absolute differences in discharge rate between motor units are due to the organization of the central and peripheral inputs to the given motor unit pool as well as differences in the intrinsic properties of the motoneuron (98). If during a maximum contraction, there is an extreme amount of central drive, the motor unit discharge rates will eventually converge to a near common value irrespective of motor unit threshold (98).

It has been suggested that common drive reduces the burden placed on the central nervous system to regulate motor unit discharges independently (98), however, the central versus the peripheral origins of the common drive is still the subject of investigation (194). Theoretically, isometric contractions minimize muscle spindle activity, which allows the central drive to dominate motoneuron pool excitability (63). De Luca and associates (100) suggested that the common drive is central in origin but is modulated by proprioceptive feedback of either the muscle spindles or the Golgi tendon organs.

One theory is that the central nervous system controls all the activations of the motor units through a central command of the excitability of the motoneuron pool (86, 98, 102). The central and peripheral nervous systems regulate motor output through the modulation of excitatory inputs to the motoneuron pool. The excitation is in response to the modulation of twitch forces to maintain a desired level of force, which subsequently leads to expected modifications in motor unit discharge rate (85, 102). Thus, there is an intricate relationship between motor unit twitch force characteristics, motor unit discharge rates, and eventual force output (45, 85, 228, 230, 242).

The interrelationship between motor unit twitch force characteristics, motor unit discharge rates, and force output is thought to be linked with recruitment threshold. Later recruited, higher threshold motor units, discharge at lower firing rates and have higher force output twitches (86, 96). When recruited during a voluntary contraction to sustain target force output, it may be necessary to decrease the discharge rate to minimize force fluctuations (84, 85, 317, 418). Force modulation through common drive is then evident as common oscillations in active motoneurons discharge rate (1-2Hz), due to changes in excitatory neural drive from common inputs (86, 98, 133).

The variations in motor unit discharge rates may also be affected by the feedback from the muscle spindles (34, 59, 86). The spindles can either facilitate or inhibit motoneurons, causing a subsequent variability in motor unit discharge rates and a decrease in the amplitude of the cross-correlation used to measure common drive (84). Therefore, as force output is increased and more motor units are recruited, force fluctuations may also increase (84, 101, 317, 418). Motor unit variability will be discussed in the following section.

2.6 Sex Differences in Force Steadiness.

According to Brown and colleagues (56) isometric force steadiness is the ability to maintain a constant contraction around a given force output. Fluctuation in force lead to decreased steadiness in the contracting muscle. The level of force output (maximal or submaximal) affects the degree of force fluctuations. The greatest fluctuations present in a parabolic shape with very high and very low force outputs having greater fluctuations compared to moderate force outputs (56, 91, 124, 186, 263, 388, 396). The pattern of

motor unit behaviour (MUDR and recruitment strategies) along with the capacity of the muscle to generate force will affect the amplitude of force fluctuations (75, 124, 388, 418). In the elbow flexors sex differences have been reported to be minimal at intermediate force outputs when force gradation is being controlled through both the recruitment of motor units and rate coding (374). Brown and colleagues (56) reported, when performing isometric elbow flexion, females did not exhibit differences in force steadiness between 10 and 75% MVC.

Oscillations in the common synaptic input to the motor unit can explain a large portion of fluctuations in force that are influenced by descending commands (328, 329). Synaptic noise is the result of motoneurons receiving synaptic inputs from thousands of connections. These numerous inputs can lead to variability in action potential discharge times as the motoneuron membrane potential fluctuates (27, 111). Random fluctuation in membrane potential or 'synaptic noise' can be observed in the coefficient of variation of the motoneuron inter-spike interval (CV ISI) (111). However, the variability in the ISI can also be influenced by oscillations in common synaptic inputs to the motoneuron population (102). It has been suggested that force steadiness is associated with either the ISI of motor unit discharge times (124, 317) or 'motoneuron noise', which is the common low-frequency oscillations in neural drive (motor unit action potential trains, MUAPT) (330). This can be detected by the first common component (FCC). The FCC is a signal that projects the largest common variation in a smoothed MUDR and can explain most (~70%) of the force fluctuations in a given signal (330).

During force gradation, a greater number of motor units must be recruited when moving from low force outputs to higher force outputs. Differences in discharge rates

between the originally recruited and the newly recruited motor units may be the origin of the overall variability in the discharge rates of active motor units (84, 96). Once beyond the recruitment range of new motor units and as the motor unit approaches its maximal discharge rate, variability is greatly reduced (148, 317, 400). Additionally, numerous research groups have indicated that synchronization has an influence on the force profile and its stability (80, 156, 174, 299, 344, 346, 418). Increases in force fluctuations may manifest as a result of musculoskeletal stiffness, laxity of the involved joint or an increase in the variability of the motor unit discharge rate associated with motor unit recruitment strategies (doublets, synchronization) (80, 344, 346). Force fluctuations resulting from musculoskeletal stiffness comes from the slack in the series elastic component (SEC) that inevitably needs to be taken up prior to force production. Motor unit discharges stretch the SEC and subsequently shorten the muscle fiber length. If synchronization results in increasing the interval between motor unit discharges, the elastic components in the musculotendinous complex would allow the muscle and muscle fibers to lengthen. Lengthening of muscle fibers has been shown to reduce the force output. Therefore, successive motor unit discharges, or synchronous pairs of discharges, would be necessary to shorten the slack musculotendous complex, returning the muscle and its fibers to their previous length and force output (344). The addition of either more active motor units or variability in discharge rate could contribute to fluctuation in force observed during synchronization (253).

Fluctuations in force steadiness may also result from an inability to achieve fused tetanus during voluntary contractions in the female participants (126) leading to greater force fluctuations due to submaximal activation of many motor units. Briefer twitches are

associated with less fusion in motor unit force at a given activation rate which causes greater force variability. Motor unit recruitment and rate coding influence force output by altering the amount of twitch fusion (388). This may be a recruitment strategy that takes advantage of the muscles catch like properties to increase force gradation prior to fused tetanus (35). Taylor and associates (388) reported that MUDR CV is the most likely factor in force variability in the FDI (363). Slight increase in MUDR can affect force fluctuations (388) due to the individual motor units discharge rates being associated with the steep portion of the force frequency curve during moderate force outputs (30) leading to relatively large fluctuations in force output from minor oscillations in the instantaneous discharge rate. It has been reported in experimental and simulation studies that short-term motor unit variability and force fluctuations are causal (102, 263, 317, 418), while others report no specific relationship. A lack of causal relationship may come from the understanding that previously active motor units should overwhelm the variations caused by the newly recruited motor units (67, 84, 111, 133, 136, 286, 330). However, conflicting results may also be due to differences in task, muscle group or the level of force output required during these studies. Nevertheless, the relationship between motor unit variability and fluctuations in force output remain unresolved.

2.7 Sex Differences

The myoelectric signal may be affected by various sex differences. Pincivero and colleagues (342) indicated that there is a distinct sex-related difference in muscle fiber type, which may be displayed in the myoelectric signal. The proportion of muscle fiber types activated is essential to most myoelectric measurements. With frequency measurements closely related to muscle fiber conduction velocity, muscle temperature,

fatigue, skinfold thickness (31, 33, 95, 161, 262, 384) as well as fiber type, any sex differences in these variables could manifest in either the frequency or amplitude of the myoelectric signal. The physiological cross-sectional area of muscle fibers was another reported factor. Physiological cross-sectional area has been shown to be smaller in the legs of females (76, 197, 379). Considering there is a linear relationship between muscle fiber conduction velocity, spectral frequency and physiological cross-sectional area (262), it would be expected that males, who have larger diameter tibialis anterior muscle fibers (197), would have higher frequency components, as has been reported at submaximal levels in the tibialis anterior (76).

Sex differences in skinfold thickness have been reported act as a low-pass filter which can reduce the amplitude of the myoelectric signal. There is a larger volume of tissue between the recording electrode and the active muscle, which acts as a low pass filter, subsequently reducing the amplitude of the summated muscle fiber action potentials in females (104, 159, 280). Christie and Kamen (74) investigated the sex differences in motor unit discharge rate in the tibialis and demonstrated that males had an approximately 9% higher motor unit discharge rate than females (Females: 29.3 ± 7.0 Hz; Males 32.7 ± 6.8 Hz). This was one of the first studies to look at potential sex differences in motor unit discharge rate and reported that the lower motor unit discharge rate at maximum tension in females may be a main factor in sex differences in maximum force production. However, with weak correlation (r = 0.33) between maximum force production and motor unit discharge rate suggests that other factors may be involved. Unfortunately, Christie and Kamen (74) only investigated differences at maximum

tension. It is not known if differences in maximal motor unit discharge rates are evident at submaximal levels, or if maximal effort conditions reflect a neural constraint.

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3 PURPOSE

The overall purpose of this thesis was to examine the neural control of muscle in a healthy population of males and females, to illuminate any potential neural and biomechanical differences that would account for sex differences during dynamic muscle contractions. Central and peripheral mechanisms responsible for force output, motor unit behaviour, and the best statistical approach to studying these differences were explored.

3.1 Manuscript 1

The purpose of the first study was to assess sex differences in maximal effort isometric contractions of the elbow flexors to determine the contribution of strength and rate of muscle activation to sex-related differences in the maximum rate of force development in the upper limb. A secondary purpose was to compare two statistical approaches to data analysis to determine the most theoretically sound approach and their impact on the interpretation of results. The two approaches being compared were normalization through the creation of ratio scores and the use of statistical adjustment through ANCOVA modeling.

3.2 Manuscript 2

The purpose of the second paper was to determine if sex differences in the maximum rate of isometric dorsiflexion force development are determined by maximum isometric dorsiflexion force alone or are other factors, as was observed in the upper limb. Based on the findings from the first study, where neural factors contributed to sex differences in upper limb rate of force development, the second study sought to address whether or not these differences were either central (neural) or peripheral (muscular) in

nature. This was done in the lower limb where males and females are more similar. A second purpose of the study was to address the effects of task familiarization to examine how the ability to achieve a true maximal effort contraction effects the results.

3.3 Manuscript 3

The purpose of the third study was to extend the finding of the previous studies that reported sex differences in the neural control of muscle and potential biomechanical differences. This study was also designed to be an extension of the findings of Christie and Kamen (74) who reported sex differences in motor unit discharge rate during a maximum voluntary contraction. However, they only looked at the differences at maximum. It is not known whether or not differences in MUDRs exist throughout the gradation of force, or if there are different neural strategies involving both rate-coding and motor unit recruitment, as suggested by sEMG data. Therefore, the investigation was designed to evaluate potential sex differences in motor unit behavior between males and females during the force gradation process from submaximal to maximal force output. A secondary goal was to assess how these 'potential' differences in neural strategies are manifested in force steadiness.

4 MANUSCRIPT 1

Sex-related differences in maximal rate of isometric torque development

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Abstract

Sex differences in the maximum rate of torque development ($d\tau/dt_{max}$) may be due to differences in maximum muscle strength, because higher torque values mathematically lead to higher values for the rate of change in torque. The rate of change in the isometric torque-time curve is often normalized to the isometric maximum voluntary contraction (MVC) to evaluate males and females on a relative scale. Normalization eliminates sex differences in $d\tau/dt_{max}$ in the lower limbs because males and females are more comparable (i.e., differences between the sexes are relatively small) with respect to both muscle size and strength in the lower limbs. However, normalization fails to result in parity in $d\tau/dt_{max}$ of the upper limb, leading to the idea that other factors may be involved. This study determined if sex differences in $d\tau/dt_{max}$ in the upper limb can be attributed to differences in isometric MVC and/or a neural variable related to rate of increase in muscle activation (Q_{30}). Forty-six participants (23 males, 23 females) performed maximal isometric elbow flexion contractions, "as hard and as fast as possible". Maximum torque (τ_{max}), $d\tau/dt_{max}$, and the rate of increase in surface electromyographic (sEMG) activity (Q_{30}) were assessed. Muscle plus bone cross-sectional area (M+B CSA) of the upper arm was calculated to estimate differences in muscle size, only for comparative purposes. Maximum strength (55.5%) and muscle size (41.9%) of the elbow flexors in males were much greater than that of females ($p<0.05$). There was a large difference (61.2%) between males and females with respect to $d\tau/dt_{max}$ that was reduced by statistical correction using an analysis of covariance (ANCOVA). The percent differences were reduced to 36.7% ($p<0.05$) for τ_{max} and 54.4% ($p<0.05$) for Q_{30}, but were nearly eliminated to 13.8% ($p>0.05$) when both variables were used simultaneously as

covariates. Since sex differences in the upper limb $d\tau/dt_{max}$ persist, additional neural or biomechanical factors may be involved.

Key Words: Surface Electromyography; Biceps Brachii; Muscle Strength; Elbow Flexion

Introduction

The rate of force development (RFD) is as important as maximal voluntary contraction (MVC) as an important characteristic of skeletal muscle performance. Aside from the obvious implications for sport performance (1, 21), there is an appreciation for the RFD in activities of daily living (48, 50). A rapid increase in muscle force is critical for increasing joint stiffness in response to a sudden perturbation (20, 29, 33, 51, 52), and to restore balance and prevent falls (16, 50). It has also been shown that the RFD in upper limb rehabilitation has a far greater relationship with functional capacity than MVC (48).

Hannah and coworkers (21) recently demonstrated that sex-related differences in the maximum RFD are due to expected differences in MVC. When the maximum rate of torque development ($d\tau/dt_{max}$) during isometric knee extension was normalized with respect to MVC, sex-related differences were completely eliminated. It is possible that normalization was effective because males and females are generally more comparable (i.e., the percent differences are less) with respect to muscle mass and strength in lower versus upper limbs (25, 26, 45). For example, Hannah and coworkers (21) reported that males and females differed in maximal isometric knee extension strength by only 33%. Thus, we asked the question if the findings for the lower limb are generalizable to the upper limb where mass and strength differences between males and females are more pronounced (25, 26, 45).

If sex-related differences in $d\tau/dt_{max}$ are due solely to MVC, then statistical control for MVC using an analysis covariance or normalization by MVC should eliminate significant differences between males and females. However, it is possible that sex-related

differences in $d\tau/dt_{max}$ for the upper limb may persist, because there is evidence for its neural regulation independent of MVC. For example, the superposition of normalized force-time curves pre- versus post-resistive exercise training typically exhibit a distinct shift to the left for the $d\tau/dt_{max}$ portion of the trace (1, 17, 20, 28). Furthermore, previous work has shown that males and females can exhibit differences in the rate of increase in surface electromyographic activity (sEMG) during the maximal speed of limb movement (34). The underlying mechanism for differences in the rate of muscle activation may be that males have higher motor unit (MU) discharge rates during maximal isometric contractions (10), and higher discharge rates are associated with a greater $d\tau/dt_{max}$ (55).

To this end, male and female participants in the present study performed maximal effort isometric contractions of the elbow flexors, to determine the contribution of MVC and rate of muscle activation to sex-related differences in $d\tau/dt_{max}$ in the upper limb. The two prevalent methods of data analysis of sex-related differences in muscle contractile characteristics are: (a) statistical control using an analysis of covariance, and (b) normalization by the creation of ratio scores prior to statistical analysis (5, 21, 25–27). A secondary purpose was therefore to compare the two methods of data analysis and their impact on the interpretation of results.

Methods

2.1 Participants

Forty-six healthy subjects (23 females and 23 males) served as participants for the study. They were recruited from the Brock University undergraduate and graduate physical education programs and were right hand dominant. Participants had no prior orthopedic injuries of the upper limb or neurological disorders. Each subject provided informed consent prior to study participation in accordance with Brock University Research Ethics Board guidelines (REB-02-284).

2.2 Testing procedures

2.2.1 Participant characteristics

Participants reported to the Electromyographic Kinesiology Laboratory for two sessions. The first session was used to familiarize subjects with the demands of the task, obtain their physical characteristics, and complete a physical activity questionnaire (Table 1). The anthropometric data reported in Table 1 were used to calculate muscle (M) and bone (B) cross-section area (CSA) for the biceps according to the following formula:

$$M + B\ CSA = \pi \times (r - (BSF + TSF)/4)^2$$

where r is the radius of the upper arm calculated from the biceps girth, BSF is biceps skinfold, and TSF is triceps skinfold (36, 40). Muscle-bone CSA obtained by anthropometric measures has been validated against the same measures calculated using both x-ray (36), CT (11, 15, 49), and MRI (39). The anthropometric method is known to result in an overestimate in absolute terms but produce the same results when examining

differences between males and females and alterations due to resistive exercise (11, 39). Calculations for the corrected (bone free) arm muscle cross-sectional area proposed by Heymsfield and associates (24) were not performed as Hortobagyi and colleagues (30) demonstrated that the uncorrected formula was effective when studying the relationship between size and strength.

2.2.2 Strength measures

Data collection occurred during the second test session (7). Participants performed a total of seven isometric, maximal voluntary contractions (MVCs) of the elbow flexors. The contractions were held for five seconds and occurred at three minute intervals. The instructions were to flex at the elbow "as hard and as fast as possible", while verbal encouragement was given during each trial (28, 44). Voltage from a load cell (JR3 Inc., Woodland, CA) was monitored using a digital display on a computer-based data acquisition system (DASYLab, DASYTEC National Instruments, Amherst, NH). At the same time, the voltage was presented to participants on an oscilloscope (Hitachi, VC-6525) situated at eye level. The visual gain of the oscilloscope (i.e., volts per division) was the same for every subject and remained constant through the study.

After the first contraction, a target line corresponding to the peak voltage was presented on the oscilloscope. Participants were instructed to move the voltage trace from the load cell, past the target line on the oscilloscope for the next two attempts. The target line was then adjusted to 110% of the peak voltage that occurred during any one of the first three trials. A fourth trial was then performed to determine if participants could reach the new target line. If participants could reach the new target line, then the new target line was 110% of the average of the peak voltages observed on trials 2, 3, and 4. This protocol

has been shown to result in stable MVCs within four contractions and avoid fatigue (4). Two additional horizontal lines were then used to construct a target area that was ± 2.5% of the required force. After five minutes of rest, an additional three MVCs where obtained with the same work to rest ratio. The additional three contractions were required because intraclass reliability coefficients for maximal isometric elbow flexion contractions obtained on one test day begin to plateau (0.96-0.97) between contraction 5 and 7 (9).

2.3 Apparatus and testing position

The apparatus and test position have been described in detail elsewhere (7). Briefly, a seat-belt was used to secure participants at the waist to the testing chair. Shoulder straps crossing the midline of the torso further minimized any extraneous movement of the upper body. The upper arm rested at the back of the elbow on a support so the shoulder and elbow of the arm being tested were at 90 degrees of flexion in the sagittal plane. A wrist cuff was fastened below the styloid process with Velcro® straps while the forearm was in a neutral position (mid-way between pronation and supination). A load cell (JR3 Inc., Woodland, CA) attached to the wrist-cuff was used to record force.

2.4 Recording torque and surface electromyographic activity

The recording surface of the biceps brachii (BB) was shaved, mildly abraded with NuPrep®, and cleansed with alcohol to reduce impedance to less than 10 kΩ (Grass EZM5, Astro-Med Inc., West Warwick, RI). The electrodes (Grass F-E9 Ag/AgCl, Astro-Med Inc., West Warwick, RI) were placed in a monopolar configuration with the recording electrode (G1) placed directly on the motor point and the reference electrode (G2) on the distal tendon. The motor point was electrically identified using a metallic

probe over the skin surface. The lowest possible current that produced a minimally visible twitch was taken as the motor point (8). The ground electrode (CF5000, Axelgaard Manufacturing CO., LTD, Fallbrook, CA) was placed on the clavicle. This electrode configuration results in the greatest sensitivity to changes in muscle electrical activity (8). Since testing was completed within the confines of a Faraday cage, the signal-to-noise ratio (SNR) was greater than 40 dB. The surface electromyographic (sEMG) signal was band-pass filtered (3 – 500 Hz) and amplified (Grass P511, Astro-Med Inc., West Warwick, RI). The force and sEMG signals were 16-bit analogue-to-digital converted (NI PCI-6052E, National Instruments, Austin, TX) at 2048 Hz using a computer-based data acquisition system (DASYLab, DASYTEC National Instruments, Amherst, NH). The data were stored on a Pentium III PC for off-line processing (Seanix Technology Inc., Blaine, WA).

2.5 Data reduction

Figure 1 illustrates a representative trial for sEMG activity of the biceps brachii, torque (τ), and the differentiated torque signal ($d\tau/dt$). The maximum rate of torque development ($d\tau/dt_{max}$) was the maximum of the differentiated torque signal. Onset of sEMG activity was identified as the first point of the signal to rise above the 95% confidence interval for baseline noise and remain above that level for 20 msec (12). Onset of elbow flexion torque was the point in the signal where the rate of change exceeded 1% of the $d\tau/dt_{max}$. The electromechanical delay (EMD) was the time difference between the onsets of elbow flexion torque and sEMG. The rate of muscle activation (Q_{30}) was calculated according to the methods outlined by Gottlieb and associates (19). Surface EMG data was rectified then numerically integrated for 30-msec, starting from sEMG onset. Mean maximal torque (τ_{max}) and root-mean-square (RMS) sEMG amplitude were

calculated on the same segment of data corresponding to 500 msec immediately before the middle of the contraction (17). All calculations were completed in MATLAB (The MathWorks Inc., Natick, MA).

2.6 Statistical analyses

The mean of the three trials were averaged and used to analyze the data. Two approaches to statistical analysis were compared. If differences between males and females can be attributed to a particular variable, then it was used as a covariate to eliminate significant differences between groups in an analysis of covariance (ANCOVA). The τ_{max} and Q_{30} variables were selected as covariates to determine if differences between males and females with respect to $d\tau/dt_{max}$ may be attributable to simple strength differences and/or neural control. The number of hours per week spent weight training also served as a covariate to evaluated potential differences related to resistive exercise. The statistical assumption of homogeneity-of-regression was tested prior to the use of each covariate. The test involved the inclusion of a group × covariate interaction term with respect to predicting the dependent variable. None of the interaction terms were significant for any of the covariates. Thus, the statistical assumption of homogeneity of regression for the covariates was tested and maintained (38).

The second approach to data analysis was to create ratio scores by normalizing $d\tau/dt_{max}$ with respect to τ_{max} and Q_{30}, separately, before performing independent *t*-tests for significant differences between groups. Normalization involved dividing $d\tau/dt_{max}$ by τ_{max} and Q_{30} to create two separate ratio scores. The importance of a variable was determined by whether or not the ratio score eliminated significant differences between groups. The

statistical assumptions of normality and homogeneity of variance were tested and maintained (38). All statistical procedures were performed in SYSTAT (SPSS Inc., Chicago, IL) with alpha set at the 0.05 probability level. The means ± standard deviations (SD) for each measure are reported below.

Results

3.1 Participant characteristics

The means and standard deviations for the participant's characteristics are given in Table 1. The two groups engaged in physical activity the same number of hours per week ($p>0.05$). There also was no significant difference between males and females with respect to the number of years of weight training experience ($p>0.05$). Males spent more hours per week weight training than females ($p<0.05$). However, it was not a significant ($p>0.05$) covariate in accounting for sex differences in either maximal isometric elbow flexion torque ($p>0.05$) or the peak rate of torque development ($p>0.05$). While there were obvious differences in anthropometric measures, they were used to estimate muscle-bone cross-sectional area (M+B CSA) of the biceps, which was also greater (41.9%) for males than for females ($p<0.05$).

3.2 Analysis of covariance and ratio scores

Table 2 shows the means and standard deviations for the main criterion measures. As might be expected, males exhibited greater (55.5%) maximal isometric elbow flexion torque (τ_{max}) than females. The correlation between τ_{max} and M+B CSA was $r=0.65$ ($p<0.05$) for females and $r=0.63$ ($p<0.05$) for males.

The peak rate of torque development ($d\tau/dt_{max}$) was 61.2% greater for males than for females. Statistical control for τ_{max} reduced sex differences to 36.7%, which was still significant at the 0.05 probability level (p=0.02). The correlation between τ_{max} and $d\tau/dt_{max}$ was r=0.52 (p<0.05) for females and r=0.39 (p<0.05) for males. Similarly, the correlation between Q_{30} and $d\tau/dt_{max}$ was r=0.50 (p<0.05) for females and r=0.46 (p<0.05) for males and it was a significant covariate in the ANCOVA model (p<0.05). However, the difference between males and females dropped to only 54.4%, which remained well below the 0.05 probability level (p=0.0001). A two covariate model that included both τ_{max} and Q_{30}, reduced the difference between males and females to 13.8%, which was above the 0.05 probability level (p=0.39). It is interesting to note that the difference (27.3%) between males and females with respect to the overall magnitude of muscle electrical activity (RMS) was much less than that for Q_{30} (48.2%). Finally, females had a 22.7% longer EMD.

Dividing $d\tau/dt_{max}$ by τ_{max}, reduced the difference between males and females to 15.9%, which was no longer statistically significant (p=0.08). Normalization of $d\tau/dt_{max}$ by using Q_{30} as the denominator had an even greater impact. The difference between males and females was reduced to 9.8% which was well above the 0.05 probability level (p=0.66).

Discussion

It was believed that correcting for MVC would not resolve sex-related differences in the maximal rate of torque development in the upper limb, as has been previously demonstrated for the lower limb (21). The reason was that males and females are less

comparable (i.e., the percent difference is greater) with respect to muscle mass and strength for the upper limb versus lower limb (25, 26, 45). The rate of increase in muscle activation was therefore explored as an additional factor. The first section of the Discussion describes how the percent difference between males and females in upper limb biomechanical and sEMG variables compares with literature values. The two prevalent data analysis methods used to account for these obvious differences are discussed in relation to the interpretation of results in the second section of the Discussion.

4.1 Comparative values

The percent difference in muscle mass, maximum strength, the maximum rate of torque development, and in sEMG variables for the upper limb are comparable to other studies in the literature. The present study observed a 55.5% difference in MVC between males and females. This value is in the upper range (34% –54%) of what has been reported for sex differences in maximal isometric elbow flexion strength (6, 31, 32). Hannah and coworkers (21) observed that males were only 33% greater than females in maximal isometric knee extension strength.

Muscle-bone cross-sectional area was calculated only to estimate the difference in muscle size of the upper limb between males and females, and compare the observed difference with literature values. No hypotheses are made with respect to M+B CSA and $d\tau/dt_{max}$. The values reported by Cureton and colleagues (11) are similar to those reported in Table 2, so that the percent difference (45.3%) compares favorably with our observations (41.9%). The percent difference in size of the elbow flexors (41%) was

nearly identical to that reported by Miller and associates (45) who used computerized tomography scans. Miller and associates (45) also reported that the total knee extensor CSA for males was only 25% greater than that for females. The moderate correlations observed between muscle size and strength for males (r=0.63) and females (r=0.65) in the present study are consistent with what has been reported in the literature (13, 41, 46).

Methodological differences with respect to measuring elbow flexion forces rather than torques, and the definition of the rate of force development as the percent MVC achieved within a given time interval, makes a comparison of $d\tau/dt_{max}$ to literature values difficult. If the values for $d\tau/dt_{max}$ in Table 2 are converted to the maximal rate of change in force (dF/dt_{max}), Bell and Jacobs (6) reported an identical percent difference (53%) between males and females, compared to the present study. Hannah and coworkers (21) defined the rate of force development as the percent MVC achieved within a given time interval and observed a 29 to 56% difference, depending on the time interval.

Ives and colleagues (34) compared maximal isotonic resisted elbow flexion speed with a 70% MVC pre-load in collage aged males and females and observed a peak biceps brachii sEMG of 2.44 ± 0.8 mV and 1.44 ± 0.73 mV, respectively. The overall magnitudes are similar to our data but the percent difference between males and females is much larger (41%) than what we observed (27.3%). The investigators also measured the slope of biceps brachii sEMG and observed a 39% difference between males and females whereas a 48.1% difference in Q_{30} was observed in the current work. The values for EMD obtained using the double threshold method for sEMG onset depends on background activity, and is only appropriate for signals with a high SNR (40 dB) as occurred in the present study. Bell and Jacobs (6) reported an EMD of 26.7 ± 10.3 msec

for males and 34.5 ± 16.3 msec for females, so the percent difference (22.6%) is nearly identical to what was observed in the present study (22.8%).

4.2 Analysis of covariance and ratio scores

The main findings of the present study depended on the method of data analysis. Statistical control through an ANCOVA revealed that neither τ_{max} nor Q_{30} could completely account for sex-related differences in $d\tau/dt_{max}$, but the two variables in combination could eliminate significant differences. In contrast, normalization by either τ_{max} or Q_{30} through the creation of ratio scores could account for sex-related differences in $d\tau/dt_{max}$.

The τ_{max} measure was more effective in reducing sex differences in $d\tau/dt_{max}$ than Q_{30}. The moderate correlation observed between a measure of muscle size and its strength as observed in this present work has long been an argument for the role of neural factors in the expression of muscular strength (13, 41, 46). Thus, inclusion of τ_{max} as a covariate not only incorporates muscle size differences but also neural factors not captured by Q_{30}. It is reasonable to suggest that τ_{max} also reflects some aspect of neuromotor coordination of the upper limb during maximal efforts. For example, Ives and colleagues (34) monitored biceps and triceps brachii sEMG during maximal speed of forearm flexion against inertial resistances of 0, 40, and 70% MVC. After the point of quick-release, females had a lower rate of increase in biceps sEMG (i.e., Q_{30}) than males. The authors showed the maximal rate of increase in biceps sEMG was constrained by a limited ability of triceps activation to regulate acceleration and control the breaking process. A number of investigators have reported a regulatory role for antagonist muscle activity in maximal isometric

contractions (43, 53, 54). Furthermore, there is increasing evidence that males and females employ different neural strategies for the control of agonist, antagonist, and synergists during maximal voluntary contractions (3, 14, 18, 35).

The elimination of sex-related differences in $d\tau/dt_{max}$ by the inclusion of Q_{30} as a covariate in addition to τ_{max}, indicates that another neural factor is involved. Males have been observed to have higher MU discharge rates than females during maximal isometric contractions (10). Furthermore, Van Cutsem and colleagues (55) have shown that higher MU discharge rates are associated with a greater $d\tau/dt_{max}$. While these finding are for the tibialis anterior, it raises the same possibility for the biceps brachii. That is, males in the present study had higher MU discharge rates as revealed by a greater Q_{30}, which partially contributed to sex differences in $d\tau/dt_{max}$. Range compression is the recruitment of higher threshold MUs at lower force levels during dynamic muscle contractions and has been observed in the upper limb (22, 23). It is possible that there are sex differences in the ability to recruit higher threshold MUs under the extreme time constraints imposed by dynamic force development.

It is tempting to explain the origins of sex differences in τ_{max} and $d\tau/dt_{max}$ on different patterns of upper limb use (26). However, participants in the present study were physical education students. There were no significant differences between groups with respect to total hours of physical activity per week and years of weight training experience. Males did spend more hours per week weight training, but it was not a significant covariate in the ANCOVA model used to test for significant differences between males and females with respect to $d\tau/dt_{max}$. This finding supports the earlier work of Ives and colleagues (34) that previous experience and patterns of upper limb use failed to explain sex differences

in neuromotor coordination during maximal effort movements of the upper limb. Multiple regression analysis of detailed anthropometric measures of the upper limb have revealed that males and females have different predictors of maximal isometric elbow flexion strength, suggesting that upper limb biomechanics may play a role (42).

The construction of ratio measures by dividing the dependent measure (i.e., $d\tau/dt_{max}$) by an independent measure deemed important by the investigator (i.e., τ_{max} or Q_{30}) produced opposite results to the ANCOVA approach. Both measures eliminated statistically significant differences between males and females but it was Q_{30} that had a profound impact on the analysis, not τ_{max}. The investigators could easily interpret this finding as the rate of muscle activation being the only, or at least the main source of sex-related differences in $d\tau/dt_{max}$. In contrast, the ANCOVA approach suggested additional factors requiring further exploration. It has been argued that the ANCOVA approach is more statistically valid because the construction of ratio measures is not an appropriate data transformation. The rank order of the scores and variance structure of data must remain unaltered (2, 47). This condition can only be met when there is a perfect correlation (r=1.0) between the numerator and denominator, and this is not the case between force, its derivative, or any other physiological variable. As a result, statistical analysis of ratio scores can lead to spurious conclusions (37, 47).

Conclusions

Both the normalization and covariate approach cannot be used to establish causal relationships. Maximal voluntary contraction and Q_{30} have been shown in other studies to have a relationship with $d\tau/dt_{max}$. The present work showed that these known linkages

with $d\tau/dt_{max}$ explain only a portion of the differences between males and females while additional neural and/or biomechanical factors may be involved.

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Table 1 – *Characteristics of the study participants*.

Measure	**Females** (N=23)	**Males** (N=23)
	M ± SD	M ± SD
Age (years)	23 ± 1.6	23 ± 3.3
Height (m)	1.7 ± 0.1	1.8 ± 0.1*
Mass (kg)	63.3 ± 8.3	88.9 ± 12.3*
Forearm length (cm)	23.9 ± 1.5	29.1± 2.8*
Elbow circumference (cm)	23.5 ± 1.5	29.1 ± 2.8*
Biceps skin-fold (mm)	10 ± 3	8 ± 4*
Triceps skin-fold (mm)	20.5 ± 4.3	19.0 ± 10.2
Muscle-bone cross-sectional area (cm^2)	45.8 ± 8.0	78.8 ± 15. 7*
Physical activity (hours/week)	5.4 ± 5.4	4.8 ± 3.9
Weight training		
Years	4.3 ± 3.2	5.4 ± 3.8
Hours/Week	2.4 ± 2.1	4.4 ± 3.2*

*Significant at the $p<0.05$ probability level.

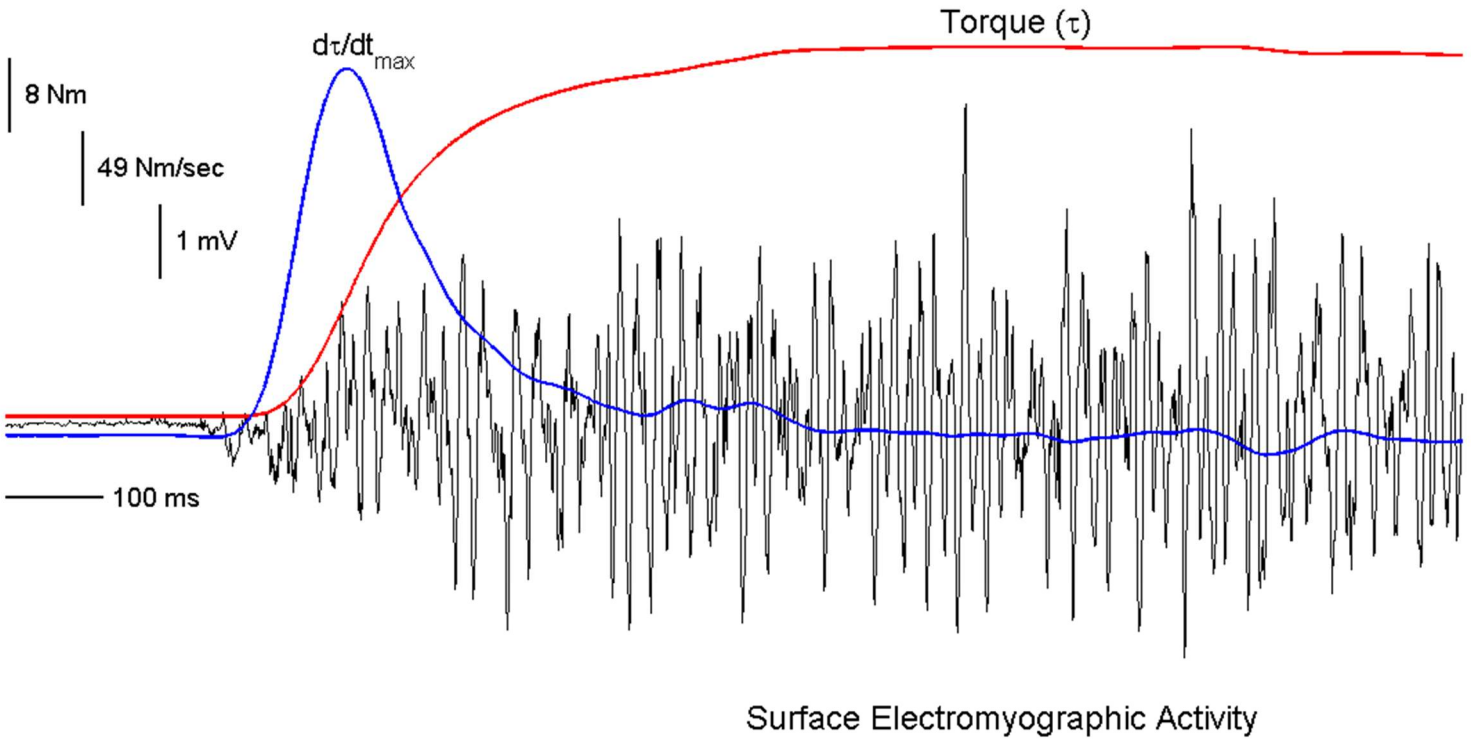
Table 2 – Means (M) and standard deviations (SD) for the following criterion measures: maximum torque (τ_{max}), maximum rate of torque development ($d\tau/dt_{tmax}$), maximum rate of increase in surface EMG (Q_{30}), root-mean-square surface EMG (RMS sEMG), electromechanical delay (EMD).

Measure	**Females** (N=23)	**Males** (N=23)	**Percent Difference**
	M ± SD	M ± SD	
τ_{max} (N·m)	46.3 ± 8.5	103.4 ±24.4*	55.5%
$d\tau/dt_{tmax}$ (N·m /s)	320.6 ± 113.3	825.8 ± 273.9*	61.2%
Q_{30} (mV× s)	27.2 ± 29.6	52.5 ± 31.2*	48.2%
RMS sEMG (mV)	1.6 ± 0.7	2.2 ± 0.4*	27.3%
EMD (ms)	34.5 ± 12.1	26.7 ± 9.3*	22.2%
$d\tau/dt_{tmax}$ (N·m /s) / τ_{max} (N·m)	6.9 ± 2.1	8.2 ± 2.6	15.9%
$d\tau/dt_{tmax}$ (N·m /s) / Q_{30} (mV× s)	19.3 ± 17.3	21.4 ± 13.0	9.8%

*Significant at the $p<0.05$ probability level.

$$Percent\ Difference = \left(1 - \frac{Smaller\ Value}{Larger\ Value}\right) \times 100$$

Figure 1. Torque (τ), surface electromyographic (sEMG) activity, and the peak rate of torque development ($d\tau/dt_{max}$) for a representative subject.



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5 MANUSCRIPT 2

Neural, biomechanical, and physiological factors involved in sex-related differences in the maximal rate of isometric torque development

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Abstract

Resent research has reported that lower maximal rate of torque development ($d\tau/dt_{max}$) exhibited by females, relative to males, during knee extension can be accounted for by normalization to a maximal voluntary contraction (MVC), however, this was not seen in the upper limb. **PURPOSE:** The aim of the current work was to examine the contribution of maximum strength (τ_{max}), twitch contraction time (CT), muscle fiber condition velocity (MFCV), and rate of muscle activation (Q_{30}) to sex differences in the $d\tau/dt_{max}$ during maximal isometric dorsiflexion. **METHODS**: Thirty-eight participants (20 males; 18 females) performed both maximal voluntary and evoked isometric contractions of the tibialis anterior across three days. Ten maximal compound muscle action potentials were elicited and subsequently followed by three, 5-second contractions. From the recordings MFCV, $d\tau/dt_{max}$, τ_{max}, CT, electromechanical delay (EMD), root-mean squared (RMS) amplitude, peak-to-peak voltage (Vpp), and Q_{30} were calculated. **RESULTS**: An ANCOVA showed that τ_{max} accounted for all the sex differences in $d\tau/dt_{max}$ (p=0.96). There were no significant differences between groups with respect to MFCV, RMS amplitude, Vpp amplitude or CT. However, there was a significant sex difference in $d\tau/dt_{max}$, τ_{max}, and Q_{30}. Females had longer evoked EMD times compared to males (15.69 ± 10.57 ms versus 9.95 ± 3.46 ms; p=0.01), but the voluntary EMD times were not different. **CONCLUSION**: The current research supports the work by Hannah and colleagues (2012) that normalization to MVC in the quadriceps is able to account for all sex differences in rate of toque development in the lower limb.

Key words: Electromechanical delay; Muscle activation rate; Neural control; Strength testing; Surface electromyography; Muscle fiber conduction velocity; Contraction time

Introduction

The rate of tension development has received increased attention as a critical aspect of dynamic muscle performance during activities of daily living (e.g. balance maintenance) and sport performance (1, 32, 40, 42, 44, 48). A number of studies have shown that there is an inextricable link between the ultimate strength of the muscle and its rate of tension development (2, 26). However, neural factors can also play an important role, as training-related increases in the maximal rate of tension development are associated with an increase in muscle activation at the onset of contraction (49, 50). Inglis and associates (28) recently showed that neural factors also play a role in sex differences in the maximal rate of tension development in the upper limb. Maximum strength was used as a covariate and was only able to account for a portion of the sex differences. The addition of a second variable, the maximum rate of electromyographic activity at the onset of contraction (Q_{30}), was able to eliminate statistically significant sex differences.

Given the inherent relationship between muscle strength and the maximum rate of tension development, few studies have explored additional mechanisms that would explain sex differences in the maximal rate of tension development (1, 2, 13, 50). However, sex differences in lower extremity musculoskeletal injury rates (10) and falls incidences (21, 46) may be linked back to this critical aspect of muscle contraction (3). Hannah and associates (19) explored potential neural and biomechanical factors involved in sex differences in the maximal rate of tension development in the quadriceps, in addition to maximum strength. Twitch properties, electromechanical delay, and muscle activation using surface EMG were also assessed. Consistent with the more general

findings for the relationship between maximum strength and the rate of tension development, when the peak rate of tension development was normalized with respect to maximum isometric strength of the muscle, the sex differences were completely eliminated.

The results by Hannah and associates (19) in the lower limb are contrary to the finding of Inglis and associates (28) who showed a role for neural factors in the upper limb. It may be hypothesized that that since males and females were more comparable in absolute strength in the quadriceps (Δ33%) than the biceps (Δ55.5%; (28)), strength may entirely explain sex differences in the rate of tension development in the lower limb. In contrast, the difference in maximum strength between males and females in the upper limb observed by Inglis and associates (28) was much greater (Δ55.5%), possibly allowing for additional factors, such as the rate of muscle activation, to play a role. Inglis and associates (28) explored the possibility that the interpretation of the results may differ based on normalization to maximum voluntary strength versus the use of a covariate approach. It was found that normalization to maximum voluntary strength also failed to account for all sex differences in the maximal rate of tension development.

Since there is a strong relationship between maximum strength and the rate of tension development, sex differences may be accounted for by maximum strength only when the two groups are 'more' comparable with respect to maximal strength as exists in the lower limb compared to the upper limb. For example, it has been shown that males and females are more comparable in maximal isometric dorisiflexion strength (Δ28.9%) and identical with respect to RMS surface EMG magnitude (22, 24, 33). Unfortunately, Lenhardt and colleagues (2009) did not assess the maximum rate of torque development.

The 28.9% strength difference between the sexes is consistent with Holmbäck and colleagues (25) who concluded that muscle cross-sectional area was the principal determinant of dorsiflexion strength. In general, the muscle cross-sectional area of the TA in males is only 20% larger than that for females (25, 29).

The purpose of the current paper was to determine if sex differences in the maximum rate of isometric dorsiflexion torque development are determined by maximum isometric dorsiflexion torque alone, or other factors as observed for the upper limb where the strength differences are more pronounced. Based on the work of Hannah and colleagues (19), it was hypothesized that maximum isometric strength would account for sex differences in the maximum rate of torque development in the TA because males and females are more comparable in maximum isometric strength in the lower limb. Studying mechanisms behind sex differences in distal muscles that are responsible for balance and explosive activity can guide specific training interventions or rehabilitation techniques.

Methods

Participants

A power analysis was performed prior to data collection based on research by Lenhardt and colleagues (33), showing a subject pool of 18 males and females was sufficient to show differences in the rate of muscle activation (Q_{30}). However, to protect against subject drop out 20 subjects (males and females) were recruited. Pre-tension on the load cell was observed in two female subjects, so their data was removed from the analysis. Thus, thirty-eight healthy Brock University Kinesiology students (20 males and 18 females) were analyzed in the current study. The participants were free of any

orthopedic or neuromuscular disorders, right leg dominant and provided written informed consent prior to study participation in accordance with the Brock University Research Ethics Board guidelines (REB-02-283). Each participant was familiarized with the Electromyographic Kinesiology Laboratory prior to the first testing session. Prior to testing participants were asked about their history of physical activity and weight training (years of experience) as well as the duration (hours per week and per day) and the percentage of weight training focusing on the upper or lower body.

Experimental setup and scheduling

All testing was performed as the participant sat in a custom built testing chair designed to isolate the dorsiflexors during maximal isometric contractions. Participants sat with their hip and knee joints secured at 90° of flexion and the ankle joint secured at 110° of plantar flexion (27). Slight plantar flexion was chosen as previous research has shown that a certain degree of plantar flexion produces a maximal torque and 110° may place the TA closer to optimal length for both maximal evoked and voluntary dorsiflexion torque production, which takes into account the lever arm length (36). A load-cell (JR3, Woodland, CA, USA) was secured under the foot plate of which the foot was restrained by a minimally padded steel bar located proximal to the metatarsals for all torque recordings (6). There were three days of testing to assess the reliability of the measures as participants can exhibit a learning effect during maximal strength assessment (18). On each day participants were asked to perform the same tasks. These tasks included both voluntary isometric maximal dorsiflexion contractions and maximal evoked isometric torque. Each of the three testing days was separated by at least 48 hours in order to avoid any complications, which may arise as a result of fatigue.

sEMG recordings

Participants lay supine on a gurney so that the most prominent TA motor point may be electrically identified using a metallic probe over the skin surface (7). The lowest possible current that produced a minimally visible twitch was taken as the motor point (5). Following motor point identification, the recording areas were shaved, mildly abraded (NuPrep; Weaver and Co., Aurora, Colarado), and finally cleansed with alcohol to minimize skin-electrode input impedance. The sEMG recording electrode had three parallel stainless steel bars which resulted in two bipolar signals. Each stainless steel bar was 1mm in diameter, 10 mm long and was mounted with an inter-bar distance of 5 mm. The recording electrode was prepared with double-sided adhesive tape, electrolyte gel (Signal Gel; Parker Laboratories, Inc., Fairfield, New Jersey) and placed in line with the muscle fibers, 1 cm distal to the motor point. Alignment and final placement of the electrodes for recording MFCV followed the procedures outlined in McIntosh and Gabriel (37).

Finally, a ground electrode (CF5000; Axelgaard) was placed on the lateral malleolus (37). Electrode-skin input impedance (Grass EZM5, Astro-Med Inc., West Warwick, RI) was assessed before and after the experiment to ensure it remained below 10kΩ. Skin temperature (Electrotherm TM99A; Cooper Instrument Corp., Middlefield, Connecticut) was also monitored before and after the experiment to verify there was no change, which could affect the stability of the myoelectric signal.

Surface EMG was band-pass filtered (between 10 and 1000 Hz) and amplified (Grass P511; Astro-Med) to maximize the resolution on a 16-bit analog-to-digital

converter (MI PCI-6052E; National Instruments, Austin, Texas). All signals were collected at 5000 Hz and acquired on a computer-based data acquisition system (DASYLab; DASYTEC National Instruments, Amherst, New Hampshire). The data were stored on a PC (Celeron; Dell, Round Rock, Texas) for offline analysis.

The data window for sEMG analysis of the maximal voluntary contraction was 500 ms, terminating before the middle of the contraction (28). The sEMG signals were up-sampled to 25 kHz prior to calculating the cross-correlation coefficient to increase the time resolution of the action potential propagation (11). Muscle fiber conduction velocity calculation was based on the time delay identified by the peak of the cross-correlation function and the known inter-bar distance of 5 mm. The root-mean-square (RMS) amplitude of sEMG activity was also calculated. The rate of muscle activation was calculated by first rectifying the sEMG data and then numerically integrating the first 30ms starting from the sEMG onset (Q_{30}) that represents the rate of increase in the sEMG over the first 30 ms of muscle activity (16).

Electromechanical delay comprises an important portion of the rate of tension development phase of the contraction (14), where changes in motor unit activity patterns have been demonstrated to play a critical role in the maximal rate of torque development (50). Electromechanical delay was determined from the time lag between the onset of dorsiflexion torque and sEMG. Surface EMG onset threshold was identified as the first point of the sEMG signal to rise above the 95% confidence interval for baseline noise and to remain above the 95% confidence interval for 20 ms (9). Visual inspection was utilized to ensure the accuracy of the established threshold's ability to detect either torque or sEMG onset (28).

Evoked isometric compound muscle action potentials.

Compound muscle action potentials were evoked through a cathode stimulating electrode placed over the fibular nerve along with an anode placed on the medial condyle of the fibula to evoke an isometric dorsiflexion twitch contraction. The evoked potentials were monitored on an oscilloscope (VC-6525; Hitachi) to ensure a consistent maximal response had been achieved for ten stimulations. The peak-to-peak voltage (Vpp) was extracted from the CMAP while contraction time (CT) was obtained from the torque-time curve. Evoked contraction time (CT) was calculated from the time difference between the CMAP onset to its time to peak tension (9). Contraction time is used in this paper to separate differences in the rate of torque development associated with voluntary control versus muscle fibers properties. Ultimately, it was important to determine if potential sex-difference in Q_{30} could be due to initial differences in peripheral factors (MFCV, Vpp, or CT) or overall voluntary activation (RMS).

Maximal voluntary isometric dorsiflexion contractions

After the evoked contractions a 15-minute rest period was given. Participants then performed three voluntary maximal effort isometric dorsiflexion contractions by pulling with the top of their foot against a padded metal plate while minimizing toe extension. During the MVC's each participant was asked to contract "as hard and as fast as possible" with an emphasis on the "hard" (43). Each contraction lasted approximately five seconds in duration and was separated by a five-minute rest period. The force data was converted to torque values using the lever arm length, measured from the ankle joint to the metatarsals, where the load cell was located. A target line was given which

represented 110% of the previously determined maximal effort, which was identified in real time on an oscilloscope. Furthermore, during each voluntary contraction the participants were verbally encouraged to surpass the target line. Figure 1 shows a representative torque trace, rate of change in torque, and surface electromyographic activity of the tibialis anterior during a maximal effort dorsiflexion contraction. The manner in which the signals were collected is described below.

Maximum rate of torque development was calculated from the equation provided by Andersen and Aagaard (2). The calculation involves determining the slope ($\Delta\tau/\Delta t$) over non-overlapping, successive 20 ms intervals, starting from the onset of the torque-time curve. The onset was determined as the point in the signal where the rate of change surpassed 1% of $d\tau/dt_{max}$. The $d\tau/dt_{max}$ was then the maximum slope, which is synonymous with the 'peak' of the $d\tau/dt$ curve (14).

Statistical analysis

Statistical analysis was conducted in two stages. First, to evaluate the reliability of the criterion measures using the intraclass correlational analysis of variance technique for males and females, separately. Second, significant differences between males and females (Sex) in the magnitude of the means, changes in the means across test sessions (Days), and the interaction (Sex x Days) was evaluated using a repeated measures analysis of covariance to determine the impact of potential variables that may underlie sex differences in the rate of torque development.

Intra class correlation

Intraclass correlational analysis of variance (ANOVA) was performed to evaluate the reliability of the criterion measures for each group, which requires the consideration of both the stability of means and the consistency of scores. A one-way repeated measures ANOVA was used to assess the stability of means across the three test sessions while the intraclass correlation coefficient (model 2,k) was used to evaluate the consistency of scores within subjects. We adopted the convention delineated by Fleiss (12) where an intraclass correlation coefficient (R) below 0.40 indicates poor reliability, between 0.40 and 0.75 is fair reliability, while values greater than 0.75 represent excellent reliability.

The magnitude of the intraclass correlation coefficient was further evaluated using the standard error of measurement (SEM) within an individual (17). The SEM was calculated as the square of the mean square error for the ANOVA table using the variables 'Sex' and 'Day' (51). The intra-subject coefficient of variation was the grand mean across the three test sessions divided by the SEM.

Analysis of covariance (ANCOVA)

A repeated-measures analysis of covariance was then used to determine the impact of a covariate on significant differences between males and females with respect to the $d\tau/dt_{max}$ development. Maximal torque and Q_{30} were the primary variables of interest. However, EMD, MFCV, and twitch contraction time (CT) were also explored. All statistical procedures were performed using SAS statistical software (SAS Institute Inc.,

Cary, North Carolina) with alpha set at the 0.05 probability level. The means ± standard deviations for each measure are reported below.

Results

Participant characteristics

The means and standard deviations for the participant's characteristics are given in Table 1. Significant differences were seen in all anthropometric measurements between males and females ($p< 0.05$). However, there were no significant differences in hours per week engaged in physical activity or weight lifting ($p>0.05$). The grand means and standard deviations for the criterion measures across the three test sessions for males and females are presented in tables for the voluntary and evoked contractions (see Tables 2 and 3).

Reliability analysis

Table 2 shows that the consistency of scores within subjects for the criterion measures obtained during the voluntary contractions for females was excellent (R=0.76-92) except for EMD which had an intraclass correlation coefficient of R=0.35 (See Table 2). Female participants exhibited a 27.2% reduction in EMD from session 1 to session 3 (F [2, 51] = 5.83, p=0.0045). The lack of stability, as assess by the repeated measures ANOVA, resulted in a reduced intraclass correlation coefficient. However, the intra-

subject coefficient of variation (Grand Mean/SEM) was 24.5%, which was deemed acceptable for further analyses (17).

The means of the criterion measures generated during voluntary contractions as seen in table 2 were highly stable across test sessions in males while the consistency of scores within subjects ranged from fair to excellent (R=0.61-83). The higher intra-subject variation (56.8%) for males Q_{30} resulted in a lower intraclass correlation coefficient (R=0.61) but was still acceptable.

The same was true for the criterion measures generated during evoked contractions (R=0.85-95), Table 3 shows that consistency of scores within subjects for the criterion measures obtained during the evoked contractions in males ranged from fair to excellent (R=0.57-96). The lowest intraclass correlation coefficient was for EMD but it had an intra-subject coefficient of variation of only 22.8%. While there was a slight decrease in the means across sessions, a limited range of scores contributed to a decreased intraclass correlation coefficient. Thus, the measure was still deemed acceptable for further analyses.

Between groups analyses

A repeated measures ANOVA revealed a significant difference between groups and across days for voluntary maximal dorsiflexion isometric torque (F [1, 36] = 45.97, p=0.001). Males had on average a 50.6% greater torque output than females. The difference between males and females with respect to the maximum rate of torque development was similar in magnitude. The maximum rate of torque development was 44.6% greater in males than for females (F [1, 36] = 16.46, p=0.0003). In contrast,

females had a 34.7% greater rate of increase in muscle activation as assessed by Q_{30} (F [1, 36] = 4.30, p=0.0454). There were no significant differences between males and females with respect to EMD, RMS amplitude, or MFCV (F [1, 36] = 0.00, p=0.9635; F [1, 36] = 0.31, p=0.5834; F [1, 36] = 0.86, p=0.3598, respectively).

Similar to the voluntary contractions, maximal evoked dorsiflexion torque was 53.9% greater for males than for females (F [1, 36] =28.52, p=0.001), and males also had a comparably greater maximum rate of torque development of 55.4% (F [1, 36] =13.99, p=0.0006). The EMD was 36.6% shorter for males than for females during the evoked contractions (F [1, 36] = 7.02, p=0.0119). In contrast, MFCV, evoked CT and Vpp of the CMAP were not significantly different between groups (F [1, 36] = 0.22, p=0.6451; F [1, 36] = 0.03, p=0.8589; F [1, 36] = 2.42, p=0.1288, respectively).

Analysis of Covariance

When maximal torque was used as the covariate in the repeated measures ANOVA for the maximum rate of torque development, the difference between means decreased to 1.2% (F [1, 36] = 0.01, p=0.9264). The rate of increase in muscle activation (Q_{30}) was not assessed as a covariate because females were actually greater than males. The only other significant difference between males and females was EMD during evoked contractions. However, evoked EMD had little impact as a covariate. Sex-related differences in the maximum rate of torque development was reduced to 33.8%, which was still significant (F [1, 36] = 8.27, p=0.0067).

Discussion

The main finding of the current work was that maximum torque was able to account for almost all of the sex-related differences in the maximum rate of torque development, as the percent difference in least square means was reduced to 1.2% by using maximum torque as a covariate. There also were no significant differences between males and females in surface EMG RMS amplitude magnitude as observed by Lenhardt and colleagues (33). The absence of sex differences in the sEMG signal amplitude, Vpp, and CT may reflect a comparable number of hours per week of training for the two groups (1). Contraction time was used in the study to determine if differences in the maximal rate of torque development were associated with either descending voluntary control or peripheral factors associated with muscle composition (8). Based on the fact that no sex-related differences in the CT (0.01%) were found suggests that the two groups were also similar with respect to muscle fiber composition, which is consistent with other research (23).

Based on the work of Lenhardt and colleagues (33) it was expected that males and females would be more comparable in maximum strength in the lower limb than the upper limb, which is consistent with other studies on sex differences in maximum strength (22, 24). Instead, a 50.6% difference was observed, which was the same order of magnitude as previously observed for the upper limb (28). One reason may be related to differences in the cross-sectional area (CSA) of the TA in the present sample versus that of Lenhardt and colleagues (2009). Although CSA was not directly measured, the males and females in the present study had a larger difference in lower leg girth (8.4%) than the

5.2% observed by Lenhardt and colleagues (33), which may suggest a larger difference in TA CSA.

The large difference in maximum strength between males and females was associated with a comparably large difference in the maximum rate of isometric dorsiflexion torque development (44.6%). Greater co-activation of the antagonist muscle group in females could contribute to both lower maximum strength and rate of torque development. Macaluso and associates (35) showed that females may use greater antagonist co-activation to stabilize the joint due to greater joint laxity, smaller agonist musculature, and potentially lower quality muscle mass (torque/CSA), as has been seen in older versus younger adults (45, 47). However, antagonist co-activation was not measured in the present study as it is markedly lower during isometric contractions compared to dynamic contractions and particularly lower when the muscle is placed in a shortened position as in the present study (41).

There were several novel findings in the present study. First, there were no significant differences between males and females with respect to MFCV. This result may be due to comparable muscle fiber diameters between the sexes, which is a large determinant of MFCV (31, 34, 38, 39, 54). It has been shown the type I (slow twitch) fibers are situated predominately in the anterior portion of the TA (20). Moreover, it is a general result that, while males have larger fiber diameters than females, the type I muscle fiber diameters in females are larger than their type II muscle fibers. The small inter-electrode distance (5 mm) will record from a small pick-up volume that encapsulates these superficial type I fibers of the TA, which are comparable in muscle fiber diameter between the sexes.

A second novel finding is that there were pronounced sex differences in Q_{30} that were not evident in either the voluntary (RMS) or evoked (Vpp) sEMG magnitude. Females were significantly greater than males with respect to the rate of muscle activation as assessed by Q_{30}. The difference increased further (4%) with repeated testing. This change was also associated with a significant training-related reduction in voluntary EMD associated with repeated testing. This reduction in voluntary EMG but "not" evoked EMG, highlight the change in neural control (15). We suggest that the greater Q_{30} and reduction in voluntary EMD for females was associated with a different motor unit activity pattern at the onset of muscle contraction. However, it is not possible to distinguish exact motor unit behavior that may be responsible for the sex-related differences in the rate of increase in surface EMG. Van Cutsem and associates (50) established a link between the maximum rate of isometric dorsiflexion torque, the rate of increase in surface EMG and the incidence of doublets associated with progressive resistive training. Similarly, Gabriel and associates (14) found that only three training sessions were sufficient to produce an increase in the maximum rate of isometric elbow extension torque development and mean spike frequency of the surface EMG signal during the torque development phase of the contraction in female participants.

Bojsen-Moller and colleagues (4) reported a positive correlation between the rate of torque development and tendon structure stiffness, indicating that 30% of the variation in torque development can be accounted for by the tendons mechanical properties. Winter and Brookes (53) further hypothesized that joint laxity might also play a role in the tension development phase of the contraction as reflected in the EMD. We believe that the greater rate of increase in surface EMG activation for females reflects a compensatory

mechanism to transmit force to the tendon more effectively, as evident in the observed decrease in voluntary EMD (30, 52, 53). In support of this idea, Rozzi and colleagues (1999) have shown that females exhibited greater integrated surface EMG activity upon landing from a jump as a compensatory mechanism for greater knee joint laxity, which includes both the musculotendinous unit and ligamentous restraint.

Limitations

Surface EMG can only provide an indirect measurement of differences and changes in underlying changes in motor unit activity patterns. Similarly, the use of CT scans to look at fiber composition differences only allows the association of either similarities or differences. Although the possible influence of different muscle structure and size in males versus females was discussed in the paper not directly measuring it with ultrasound is a limitation. It was also assumed that there were differences in TA tendon stiffness without having actually measured it. Moreover, caution must be applied when extrapolating the results to older adults. The use of a healthy college aged population may not account for differences between the sexes with aging, such as the loss of type II fibers, which may render the sexes more similar in the 6th and 7th decades.

Conclusion

The results support the work of Hannah and associates (19) as maximal strength of the TA as a covariate accounted for nearly all of the sex differences in the maximum rate of torque development. This was true, even though the difference in maximal strength was quite pronounced (50.6%). Thus, the hypothesis that maximum strength accounts for sex differences in the maximum rate of tension development in the lower limb due solely

to the fact that males and females are more comparable in maximum strength than in the upper limb was not supported. Rather, the maximum strength accounted for sex difference in the maximum rate of tension development despite a large discrepancy in strength between males and females. Furthermore, females had a greater rate of increase in surface EMG activation, and exhibited a significant reduction in EMD with repeated testing, suggesting that they might utilize a different motor unit activity pattern at the onset of contraction. Future research may focus on training modalities and rehabilitation techniques that could optimize RTD rather than only focusing on maximum strength to assist an older population in balance maintenance and recovery from imbalance.

Grants

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Table 1. Demographic characteristics of the study participants.

Measure	Females (N=18) M ± SD	Males (N=20) M ± SD	Difference %
Age (years)	24 ± 3.3	24 ± 2.4	0
Height (m)	1.6 ± 0.1	1.8 ± 0.1	11.1*
Mass (kg)	56.5 ± 8.5	79.7 ± 3.9	29.1*
Body Mass Index (kg/m^2)	21.4 ± 2.2	24.4 ± 0.8	12.3*
Foot Length (cm)	23.7 ± 0.7	28.5 ± 1.5	16.8*
Leg Length (cm)	39.2 ± 1.8	47.3 ± 2.3	17.1*
Leg Girth (cm)	37.0 ± 1.7	40.4 ± 1.9	8.4*
Physical activity (hours/week)	7.5 ± 3.4	9.3 ± 2.2	19.4
Weight-lifting (hours/week)	4.4 ± 3.7	5.5 ± 2.8	20.0

Significant differences were set at a $p<0.05$ level and are indicated with *

Table 2. Data under the voluntary condition

Female	vMFCV (m/s)	vEMD (ms)	vRTDpk (Nm/s)	vTorque (Nm)	Q_{30} (mV x s)	RMS (mV)
Test Day	(M ± SD)	(M ± SD)	(M ± SD)	(M ± SD)	(M ± SD)	(M ± SD)
1	5.16 ± 1.61	37.19 ± 11.06	81.15 ± 33.65	29.10 ± 7.96	7.90 ± 5.27	0.21 ± 0.10
2	5.28 ± 1.40	33.86 ± 10.07	82.74 ± 36.02	29.17 ± 7.78	8.26 ± 6.97	0.21 ± 0.13
3	5.33 ± 1.69	27.09 ± 5.30	91.43 ± 34.88	27.10 ± 6.61	8.20 ± 6.74	0.17 ± 0.11
Grand	5.26 ± 1.55	32.71 ± 9.94*	85.11 ± 34.51	28.46 ± 7.39	8.12 ± 6.26	0.20 ± 0.11
SEM	0.67	8.02	10.35	2.76	3.06	0.03
R	0.81	0.35	0.91	0.86	0.76	0.92
Male						
Test Day						
1	4.92 ± 0.99	32.92 ± 12.48	160.03 ± 70.29	57.31 ± 17.12	5.80 ± 2.91	0.18 ± 0.11
2	5.13 ± 1.25	32.52 ± 12.08	157.04 ± 70.00	58.52 ± 21.69	4.83 ± 3.21	0.18 ± 0.13
3	4.70 ± 1.09	33.08 ± 13.74	142.01 ± 82.01	56.97 ± 20.54	5.26 ± 2.73	0.17 ± 0.08
Grand	4.92 ± 1.11	32.84 ± 12.57	153.03 ± 73.47	57.60 ± 19.55	5.30 ± 3.01	0.18 ± 0.11
SEM	0.55	6.12	30.29	8.06	1.88	0.06
R	0.76	0.70	0.83	0.83	0.61	0.72
Percent Difference	6.5	0.4	44.4*	50.6*	34.7*	10

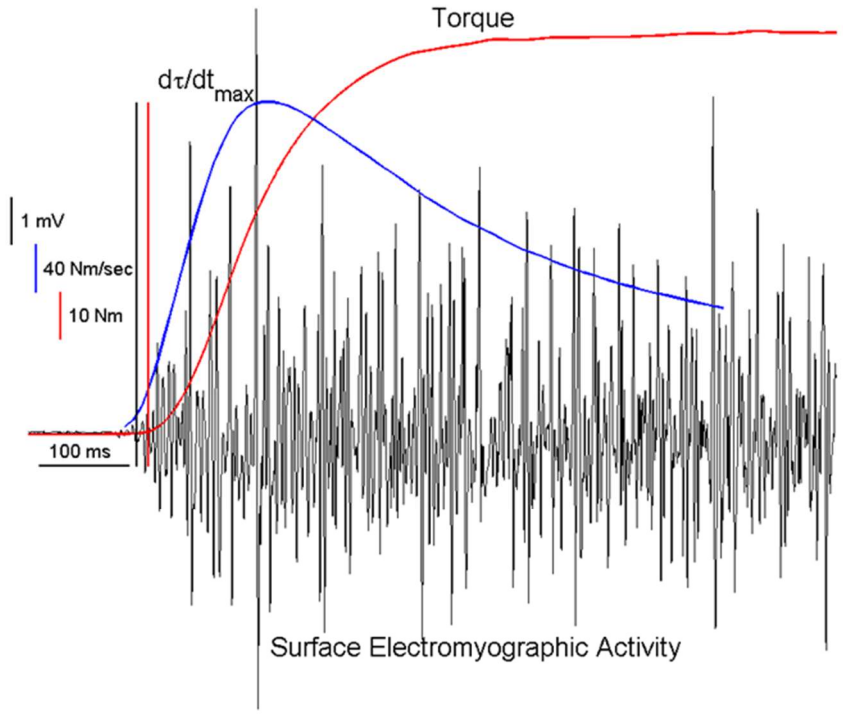
Means (M) and standard deviations (SD) for voluntary muscle fiber conduction velocity (vMFCV), voluntary electromechanical delay (vEMD), voluntary peak rate of torque development (vRTDpk), voluntary maximum torque (vTorque), voluntary rate of EMG increase over the first 30 ms (Q_{30}), and voluntary root-mean squared amplitude (RMS). Significant differences were set at a $p<0.05$ level and are indicated with *

Table 3. Data under the evoked condition.

Female	eMFCV (m/s)	eEMD (ms)	eRTDpk (Nm/s)	eTorque (Nm)	Vpp (mV)	CT (ms)
Test Day	(M ± SD)	(M ± SD)	(M ± SD)	(M ± SD)	(M ± SD)	(M ± SD)
1	4.22 ± 1.76	16.71 ± 11.21	46.07 ± 18.41	2.59 ± 1.43	2.17 ± 0.92	77.20 ± 17.63
2	4.11 ± 0.99	15.38 ± 10.98	46.75 ± 18.43	2.69 ± 1.28	2.22 ± 0.91	80.74 ± 12.18
3	4.34 ± 1.51	14.97 ± 10.02	46.25 ± 14.43	2.68 ± 1.10	2.07 ± 0.98	69.21 ± 22.78
Grand	4.22 ± 1.43	15.69 ± 10.57	46.36 ± 16.87	2.66 ± 1.25	2.16 ± 0.92	75.72 ± 18.37
SEM	0.50	4.25	4.13	0.28	0.36	4.41
R	0.88	0.84	0.94	0.95	0.85	0.91
Male						
Test Day						
1	4.33 ± 1.50	10.57 ± 3.30	106.48 ± 39.66	5.89 ± 2.33	2.59 ± 1.14	68.65 ± 7.47
2	4.71 ± 1.66	10.17 ± 4.56	101.01 ± 41.67	5.54 ± 2.48	2.51 ± 1.13	79.55 ± 11.98
3	4.17 ± 0.99	9.12 ± 2.06	104.50 ± 35.12	5.86 ± 1.99	2.73 ± 1.01	76.60 ± 8.09
Grand	4.40 ± 1.40	9.95 ± 3.46	104.00 ± 38.31	5.77 ± 2.24	2.61 ± 1.08	74.93 ± 10.33
SEM	0.76	2.27	8.57	0.45	0.40	2.32
R	0.71	0.57	0.95	0.96	0.87	0.64
Percent Differences	4.1	36.6*	55.4*	53.9*	17.2	0.01

Means (M) and standard deviations (SD) for evoked muscle fiber conduction velocity (eMFCV), evoked electromechanical delay (eEMD), evoked peak rate of torque development (eRTDpk), evoked maximum torque (eTorque), evoked peak-peak voltage (Vpp), and the evoked contraction time (CT). Significant differences were set at a $p<0.05$ level and are indicated with *

Figure 1. Torque (τ) (dark grey), surface electromyographic activity (light grey) and the rate of torque development (black) over the first second of a trial for a male representative subject. The first vertical line (black) represents the EMG onset, the second vertical line (grey) represents the torque onset.



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6 MANUSCRIPT 3

Sex differences in the Gradation of Muscle Force

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Abstract

The purpose of this study was to evaluate potential sex differences in motor unit behaviour during the gradation of force. Forty-eight (F24; M24) participants performed isometric dorsiflexion contractions at 20, 40, 60, 80, and 100% MVC. Tibialis anterior electromyography (EMG) was recorded both by surface and intramuscular electrodes. Males and females were comparable with respect to physical activity levels and resistance training of the lower limb. Compared to males, however, females had a greater motor unit discharge rate averaged across all submaximal intensities (Δ=1.24 pps, 6.25%), motor unit recruitment (Δ=4.33 counts, 5.44%), and incidence of doublets discharges (Δ=49.03 discharges, 50.00%) below 100% MVC (*P*'s<0.01). Males exhibited dramatic increases in these variables to either reach parity or surpass females at 100% MVC (*P*'s<0.01). Females had a greater force variability (16%) and a greater coefficient of variation in motor unit discharge rate (8.4%) than males (*P*'s<0.01). Overall, there was a significant repeated measures correlation between the root-mean-square force error and the standard deviation of motor unit discharge rate (r=0.56, P<0.01). The demographic data suggest that sex differences in muscle size (body mass Δ=20.4%; leg length Δ=8.0%; leg girth Δ=2.6%), not physical activity levels or resistance training of the lower limb, is the primary reason for absolute strength differences. Although, sex differences in motor unit variables with respect to the gradation of force and force variability (steadiness) may arise from neural strategies utilized by females to compensate for musculoskeletal differences, not inherent disparities in neural drive to the motoneuron pool or muscle.

Key words: Motor unit discharge rate; recruitement; doublets; force steadiness; variability; tibilialis anterior; isometric contraction

New & Noteworthy

Previous reports suggest males have greater motor unit discharge rates (MUDR) and force steadiness compared to females. This study showed females had higher MUDRs, recruited more motor units and a greater incidence of doublets submaximally. Females also had greater variability in MUDR. These differences in submaximal motor unit behaviour may be a female compensation strategy for biomechanical differences resulting in reduced force steadiness.

Introduction

It has been suggested that males have a greater tension generating capacity due to differences in muscle activation, inferred from the surface electromyographic (sEMG) signal (16, 37, 81). However, compared to males, females may rely more on neuromuscular adaptations rather than hypertrophy, to resistive exercise (22). Indirect evidence is based on the observation that females have a higher sEMG magnitude at a given level of force (52), possibly due to the recruitment of a larger number of motor units at various contraction intensities, prior to substantial increases in motor unit discharge rates (MUDR) (16, 95). Males and females may also have different motor unit activity patterns based on their differential response to muscle fatigue. Fatigue-related force fluctuations have been associated with sex differences in root-mean-square (RMS) sEMG amplitude, thought to reflect a combination of rate coding and recruitment during the force gradation process (26).

The majority of studies on sex difference in force gradation have made inferences on motor unity activity based on sEMG (4, 5, 7, 12, 16, 17, 26, 65, 74, 80, 82, 86, 95) and

it is now becoming increasingly apparent that there are difficulties with its interpretation (26). Unfortunately, there is a paucity of direct motor unit recordings. Recent research by Peng and colleagues (79) and Harwood and associates (39) evaluated sex differences in MUDRs using intramuscular fine wire electrodes. Peng and colleagues (79) investigated muscle activation patterns of the vastus medialus and the vastus medialus oblique during a straight leg raise task in two hip rotation positions. The task was performed at a rate of rise from 7.5% MVC/s and 75% MVC/s. Task position was a major factor in both motor unit discharge rates and recruitment, and females had greater MUDRs than males.

Harwood and associates (39) investigated the effect of different vibration frequencies on the elbow flexors during a contraction at 15% of maximum. Contrary to the findings of Peng and colleagues (79), Harwood and associates (39) reported that females had a 6% lower MUDR during these submaximal contractions. Further to this, Christie and Kamen (14) explored short-term training adaptations in MUDRs and after-hyperpolarization duration in the tibialis anterior using intramuscular recordings of the myoelectric signal. They reported that males had a 9% greater MUDR compared to females at maximum. However, they only examined MUDRs at maximum. It is not known whether or not differences in MUDRs exist during the gradation of force, or if there is a different neural strategy involving both rate-coding and motor unit recruitment as suggested by sEMG data. The purpose of this study was, therefore, to evaluate potential differences in motor unit behaviour between males and females during the force gradation process from submaximal to maximal muscle tension.

Methods

Participants

Forty-eight (24 males and 24 females) college aged (18-25 yrs) students were recruited to participate in the study. All participants were right leg dominant and free of any neuromuscular or orthopedic disorders. Written informed consent in accordance with the Brock University Research Ethics Board guidelines (REB-12-027) was provided prior to participation in the study. Prior to testing, participants reported their history of physical activity and weight training as well as the duration (hours per week and per day) and the percentage of weight training focusing on upper body or lower body.

Experimental Setup

All testing was completed within a grounded Faraday cage. Participants were seated in a testing chair used to isolate the ankle joint during dorsiflexion contractions. The hip and knee joints were secured at 90° with the ankle joint fixed at 20° of plantarflexion (42). A padded bar was secured to a foot plate across the distal metatarsal to allow the recording of isometric dorsiflexion contractions. Force output was recorded from a load cell (JR3 Inc., Woodland, CA) secured to the foot plate below the distal metatarsals (13).

Auditory feedback was provided (Advent, 1002, Lake Mary, FL) to ensure indwelling EMG signal quality, as well as visual feedback from an oscilloscope (Tetronix, TDS 460A, Beaverton, OR) (19, 75, 89). Visual and auditory feedback were utilized to locate areas of maximal motor unit activity inside the muscle. Adjustments of the intramuscular electrode were made to maximize the indwelling EMG signal quality (90). A second oscilloscope (Hitachi, VC-6525, Woodbury, NY) provided real-time force

feedback to the participants with error bars set at ± 2.5% of the desired level of contraction.

Subject Information and Anthropometric Measurements

Participants provided demographic information including age, weight, height, training, and physical activity background. Lower leg anthropometric measurements were collected as follows: lower leg length (fibular head to lateral malleolus), lower leg circumference (mid-calf), whole foot length (calcaneus to distal first digit), lateral malleolus to bottom of foot length, lateral malleolus to metatarsals length, and calcaneus to metatarsals length.

Surface Electromyography

The most proximal electrically identified tibialis anterior (TA) motor point was located by using a low intensity electrical stimulus applied to the TA through an anode adhered to the midpoint of the posterior aspect of the lower leg, and a hand-held probe cathode on the belly of the TA (Grass Telefactor S88; Astro-Med, Warwick, RI, USA). The motor point was identified as the area where the lowest voltage of stimulation was able to produce a muscle twitch (42). A large portion of the TA and patellar tendon was then shaved, mildly abraded (NuPrep®; Weaver and Company, Aurora, CO, USA), and cleansed with alcohol. Electrolyte gel (Signal Gel®; Parker Laboratories, Fairfield, NJ, USA) was applied to Ag/AgCl electrodes (Grass F-E9, Astro-Med Inc., West Warwick, RI), and placed in a monopolar configuration on the tibialis anterior with the recording electrode placed over the motor point and a reference electrode placed over the distal tibial tendon. A ground electrode (CF5000; Axelgaard Manufacturing CO., LTD,

Fallbrook, CA) was placed on the lateral malleolus. Skin impedance was measured both prior to and following each session with an impedance meter (Grass EZM5; Astro-Med Inc., West Warwick, RI) to ensure the impedance remained below 10 kΩ.

Intramuscular Electromyography

An approximately 5 cm area of skin distal to the surface recording electrode was shaved and sterilized with ChloraPrep® One-Step (Chlorhexidine Gluconate 2% w/v and isopropyl 70% v/v solution). An intramuscular 1.5 inch, 25 gauge quadrifilar needle recording electrode (Viasys Healthcare UK; Surrey, England) was inserted approximately one cm distal to the surface recording electrode. The ground electrode for intramuscular recordings was placed on the patella. The intramuscular needle electrode was inserted while participants performed and held a 20% maximal voluntary contraction (MVC). Auditory feedback during the low-level contraction was used to determine when the electrode passed through the fascia and into the active muscle. The needle cannula contained four 50 μm diameter platinum-iridium wires that were exposed through a side port, which allowed for the detection of individual motor units (MU). The electrode remained in the muscle for the duration of the testing protocol, with minor adjustments made as needed to ensure signal recording quality.

Signal Collection Equipment

The gains of the surface and an intramuscular EMG activity were adjusted using a Grass P511 amplifier (Astro-Med Inc., West Warwick, RI), to maximize the resolution on a 16-bit analogue-to-digital converter (NI PCI-6052E; National Instruments, Austin, TX, USA), for offline analysis (Dell; Round Rock, TX, USA). Surface EMG was band-pass

filtered between 3 and 1000 Hz, while the intramuscular EMG was band-pass filtered between 1 and 10 kHz (10, 11). All signals were sampled at 25,600 Hz using a computer-based data acquisition system (DASYLab; DASYTEC National Instruments, Amherst, NH, USA).

Experimental Protocol

Participants (N = 48) visited the Electromyography and Kinesiology Laboratory for two sessions separated by a minimum of 24 hours. The first session was a familiarization session and the second involved the testing protocol.

Familiarization Day

On the first visit participants, read the information letter outlining study requirements, and provided written informed consent. This was followed by completing a physical activity readiness questionnaire (PAR-Q) and activity journal. Anthropometric measurements were taken, and task familiarization was performed. During task familiarization, the participants were seated in the testing chair and performed five-second contractions at 20, 40, 60, 80, and 100% MVC. The contractions were used to insure the participants were able to both achieve and hold both maximal and submaximal contractions at the targeted level of force output. Each participant was asked to contract "as hard and as fast as possible" against a bar that was placed over the top of the foot, with an emphasis on "hard" (84). Great care was taken to ensure that they were not using their toe extensor muscles. Participants held the contraction at the desired force level until instructed to completely relax.

Test Day

Once the electrodes were placed and the participant was in the test position, they were instructed to perform three eight-second MVCs, each separated by three minutes of rest. Each participant performed three maximal isometric voluntary dorsiflexion contractions prior to and following the submaximal bouts. The pre and post MVCs were used to test for the presence of fatigue. Following the first set of MVCs, each participant performed three consecutive submaximal contractions at 20, 40, 60, and 80% MVC. Each of the 'Test Day' contractions were also eight-seconds in duration with three minutes of rest in-between and performed "as hard and as fast as possible". The submaximal conditions were presented in balanced order, which means that the 24 permutations were completed across subjects. Hence, there were forty-eight subjects to ensure the conditions were balanced for both groups (males and females). Desired force outputs were displayed on the oscilloscope for real-time feedback, with two horizontal error bars positioned at ± 2.5% of the target level of force.

Data Reduction and Analysis

All criterion measures were extracted from a two-second epoch which was representative of the most stable portion of the force output. Force output at each percent of MVC was the mean value of the force output in the same data window. Force steadiness, the root-mean-square error for the non-normalized (RMS_{error}) and normalized ($nRMS_{error}$) force traces (15, 72, 73, 85, 91). To calculate $nRMS_{error}$, force was normalized by dividing the entire trace by the maximum value of that trace and determining the RMS_{error} for the two second window of data. The result was multiplied

by 100 for percent error ($nRMS_{error}$). The sEMG measures were root-mean-square (RMS) amplitude and mean power frequency (MNF).

The intramuscular interference patterns were completely decomposed and then the decompositions were checked a second time in EMGlab (62) by a single operator who had intraclass correlation coefficients of 0.95 and 0.98 for agreement with two other expert users (44). The discharge times were used to create a data vector of Dirac pulses, which were convolved with a 0.95s Hanning window; the reciprocal of the result was then taken as the instantaneous MUDR (48, 55). Motor unit action potential trains (MUPTs) that had a coefficient of variation (CV) equal to or less than 20%, were selected for further analysis (90). Interpulse intervals less than 20 ms or greater than 200 ms were not used in the calculation of MUDR. However, interpulse intervals between 2 and 20 ms were used for the determination of doublet discharges (31). The number of motor units detected (MUNum) at each level of force served as a measure of MU recruitment. A total of 16,719 MUPTs were analyzed for this present study. The shimmer plots for each trial were inspected by a second investigator as a final check to ensure the accuracy of the decomposed MUPTs used for the study (**Figure 1**). Calculation of the indwelling and surface EMG measures were completed in MATLAB (Mathworks, Natick, MA).

Statistics

Demographic differences were evaluated using unpaired *t*-tests, while pre and post-testing of impedance and temperature measures were analyzed using a paired *t*-test to observe any changes over the course of the protocol. Similarly, sEMG activity was monitored to test for the presence of fatigue. Mean maximum force, RMS, and MNF of

the sEMG signal for the pre versus post set of 100% MVCs, were analyzed using a paired *t*-test to detect the presence of fatigue.

Changes in the bivariate frequency-distribution for MUDR and motor unit potential (MUP) peak-to-peak (P-P) amplitude across the different force levels was assessed using the Kolmogorov-Smirnov test, with Bonferroni adjustments for the multiple comparisons across force levels (18, 96). Preliminary analysis showed that there was no significant difference between trials, so they were averaged for each condition. Males and females (Sex), across levels of MVC, and their interaction (Sex × MVC) were evaluated using a two-factor, mixed model repeated-measures analysis of variance. Different levels of force output were used to identify potential sex differences in the following key variables: (1) motor unit discharge rate; (2) motor unit potential peak-to-peak amplitude; (3) number of motor units detected; (4) the incidence of doublet discharges (5) force variability; and (6) the CV of the MUDR. Post-hoc analyses were accomplished with orthogonal polynomials trend analysis of means while orthogonal contrasts were used for specific comparisons between means when necessary. All statistical procedures were performed in SAS® (SAS Institute Inc.; Cary, NC, USA) with alpha set at the 0.05 probability level.

Results

Methodological Controls

The average skin impedance decreased by 0.83 kΩ (Δ 11%). No practical significance can be placed on small changes in impedance below 10 kΩ, which is well within the accepted range for sEMG (40). The average skin temperature increased 0.31

degrees Celsius (Δ 1%). While it was statistically significant (t = 1.98; p = 0.04), far greater changes (± 15°C) are required to alter the sEMG signal (83, 97).

The data were also analyzed for the presence of fatigue from beginning to end of the testing protocol. Force output was averaged for the three pre-submaximal MVC trials and the three post-submaximal MVC trials. There was a 7 N (Δ 4%, t = 2.22; p = 0.03) decrease in MVC force while RMS (Δ 5%, t = 1.39; p = 0.17) and MNF (Δ 2%, t = -1.33; p = 0.19) for surface EMG remained unchanged. The small magnitude of the observed changes are considered trivial, which is consistent with previous findings wherein a similar number of isometric contractions at different percentages of MVC are present within the same test session in balanced order, with similar rest periods (28). Overall, EMG data did not exhibit the classic signs of muscle fatigue: a decrease in MPF and an increase in RMS amplitude (66, 67).

Participants

The means, standard deviations, and t-ratios for the physical and demographic characteristics of the participants are presented in **Table 1**. As might be expected, there were significant anthropometric differences. Height, weight, and BMI were significantly different between the sexes (p's < 0.01). Lower leg length and lower leg girth were similar between the sexes (p's > 0.01). Males and females were significantly different from each other for two out of the three anthropometric measures of foot dimensions. There was no significant difference between males and females with respect to the total amount of physical activity per week (p > 0.01). Males engaged in more weightlifting

hours per week ($p < 0.01$), but both sexes were similar in the percentage of total weightlift time focused on the lower limb ($p > 0.01$).

Descriptive Statistics for Motor Unit Behaviour

Changes in the bivariate frequency-distribution for MUP P-P amplitudes and their discharge rates for males and females across force levels are depicted in **Figure 2**. The colour map shows the number of MUs within each P-P amplitude-discharge rate bin. The red cross-hair in each panel illustrates the bivariate mean. Both groups exhibited an increase in MUDR and MUPs with greater P-P amplitudes from 20 to 100% MVC ($p < 0.001$). At 20% MVC, 90% of the MUP P-P amplitudes were at or below 1.5 mV for females, while it was 88% for males. In general, females had more MUs with P-P amplitudes less than 1.5 mV below 60% MVC, but the difference completely disappeared at 100% MVC.

The Kolmogorov-Smirnov test for changes in the bivariate distribution for MUP P-P amplitude-discharge rate between 20 and 100% MVC, revealed that males had a dramatic increase in the number of MUPs with P-P amplitudes below 1.5 mV. These later recruited MUs also had higher MUDRs ($D = 0.701$, $p < 0.01$). Females exhibited the same pattern of change, but to a lesser degree ($D = 0.4678$, $p < 0.01$). The colour maps for males and females also show pronounced increases in the number of MUs with P-P amplitudes greater than 1.5 mV with higher discharge rates, but they were less than 20% of the total MU population.

Sex Differences in Motor Unit Variables

The means, standard deviations and *F*-ratios for force, RMS_{error}, $nRMS_{error}$, and the CV in MUDR are presented in **Table 2**. Males were 35.76% stronger than females ($p < 0.01$). Since the submaximal force levels were a percentage of 100% MVC, the higher absolute force values for males resulted in a steeper ascent to maximum. The result was a significant Sex × % MVC interaction term for the linear trend component, which accounted for 96.44% of the variance in force means between males and females ($p < 0.01$).

The higher absolute force values for males resulted in a greater (32.1%) RMS_{error} when compared to females ($p < 0.01$). When RMS_{error} was calculated from the normalized force trace ($nRMS_{error}$), females had a 16.2% greater $nRMS_{error}$ than males ($p < 0.01$). However, both sexes followed a similar pattern of change across force levels. There was a linear decrease from 20 to 40% MVC that accounted for 29.39% of variance in means across force levels ($p < 0.01$). Between 40 and 60% MVC, females plateaued then increased between 60 and 80% MVC to another plateau. In contrast, males exhibited an increase from 60% MVC to maximum. The result was a quadratic trend component for both groups that accounted for 59.95% of the remaining variance ($p < 0.01$). Both sexes followed the same pattern of change in the CV in MUDR across force levels, with females having an overall greater level of variability of 11.5% ($p = 0.049$). Changes in the CV in MUDR did not necessarily mirror $nRMS_{error}$, but there was a net increase between 20% and 100% MVC ($p < 0.01$). However, the relationship between RMS_{error} and the standard deviation of MUDR was more tightly linked, as depicted in **Figure 3**.

The repeated-measures correlation for the relationship between RMS_{error} and the standard deviation of MUDR was $r = 0.56$ ($p < 0.01$).

The means, standard deviations and *F*-ratios for MUDR, incidence of doublets discharges, total number of MUs (MUNum), and MUP P-P amplitude are presented in **Table 3**. Both sexes had similar mean MUDRs. However, males and females exhibited different patterns of change across force levels ($p < 0.01$). Orthogonal contrast testing confirmed that females had greater MUDRs than males below 100% MVC ($p < 0.05$). Motor unit discharge rate for females plateaued between 80 and 100% MVC ($p = 0.36$), while males exhibited a dramatic increase between these force levels ($p < 0.01$). As a result, males had a significantly higher MUDR than females at 100% MVC ($p < 0.01$). Overall, females exhibited a 36.36% greater incidence of doublets discharges than males ($p = 0.03$). Orthogonal contrast testing also confirmed that females had a greater incidence of doublets discharges compared to males below 100% MVC ($p < 0.01$). Similar to MUDR, between 80 and 100% MVC, males exhibited a dramatic increase in doublets discharges reaching parity during the maximal effort contraction ($p < 0.01$).

The MUNum was similar between males and females. However, similar to the MUDR, both groups had a different pattern of change across force levels ($p < 0.01$). While males and females exhibited a progressive increase in MUNum from 20 to 100% MVC ($p < 0.01$), females had a greater MUNum than males up to 60% MVC but males overtook them at 80% MVC and above ($p < 0.01$). Motor unit potential P-P amplitude and the pattern of change across force levels were similar between the sexes. Both sexes exhibited a 30.75% decrease until 80% MVC, but orthogonal contrasts revealed a 9.13% increase between 80 and 100% MVC ($p < 0.01$). As a result, the pattern of change in

MUP P-P amplitude exhibited both linear and quadratic trend components that accounted for 79.83% ($p < 0.01$) and 16.27% ($p < 0.01$) of the variance in means, respectively.

Discussion

The purpose of this study was to explore sex-related differences in motor unit behaviour at both maximal and submaximal levels of force output. The main findings uncovered sex differences at all levels of tension for MUDR. Females had higher submaximal MUDRs but at MVC, MUDRs of males were higher. Females recruited more motor units compared to males, however, at maximum tension there were no sex differences. There was a greater incidence of doublet discharges across submaximal force levels for females compared to males, but at maximum these differences disappeared. Females also had greater fluctuations in force output as assessed by $nRMS_{error}$ and a greater CV in MUDR.

Force Output

The percent difference in maximum strength of the lower limb between males and females observed in the present work (36%) is comparable to other studies in the literature. Lenhardt and colleagues (56) reported maximum dorsiflexion force was 29% greater in males than females. The 36% and 29% strength differences are slightly lower than Brown and colleagues (8) who reported males' maximal dorsiflexion force being on average 39% greater compared to females'. It has been suggested that females may exhibit a higher degree of coactivation (53, 54, 88), which would lower overall force output at the joint. The role of coactivation in the present study was minimized by testing the ankle in 20 degrees of plantarflexion, which would place the triceps surea complex in

passive insufficiency for both sexes (58, 60, 63). Ultimately, the demographic data suggest that strength differences observed in the present study may be explained by differences in muscle size (body mass Δ = 20.4%; leg length Δ = 8.0%; leg girth Δ = 2.6%) but not the level of physical activity or amount of resistance training in the lower limb, which is consistent with the findings of Hannah and colleagues (38).

Force Error and MUDR Variability

Collapsed across force levels and sexes the $nRMS_{error}$ of dorsiflexor force was 2% which is comparable to Jesunathadas and associates (47) who reported CV below 1%. The study by Jesunathadas and associates (47) had 9 males and 2 females perform isometric dorsiflexion contractions from 2 to 90% MVC. Interestingly, there was a steep drop from 2 to 1% in the CV of force between 2 and 20% MVC, and it remained relatively stable to 90% MVC. In the present study both males and females exhibited a quadratic change in $nRMS_{error}$ similar to that reported by Yoon and colleagues (99) from 20 to 80% MVC. Specifically, force error decreased from low to moderate contraction intensity and increased above 50% MVC. Further, females had an overall greater (16%) $nRMS_{error}$ compare to males, as has been seen previously by Brown and colleagues (9) in the biceps (~ 50%) and Yoon and associates (99) in the dorsiflexors (~17%) across various force levels.

Taylor and associates (92) reported that MUDR CV is the most likely factor contributing to force steadiness in the FDI. In support, the current work showed that females had a greater MUDR CV (8.4%) and reduced force steadiness (16% greater $nRMS_{error}$) compared to males who had a MUDR CV of 7.4% across all levels of force.

The association between force steadiness and the variability of MUDR was further demonstrated by a significant correlation ($r = 0.56$) between RMS_{error} and the standard deviation of MUDR. The standard deviation of MUDR accounted 34.8% of the variance in force steadiness. The percent variance accounted for (VAF) is less, but supports the significant positive relationship (44.3 - 64.3% VAF) reported for the first principal component of smooth MUDRs and the CV of force by Negro and colleagues (71). Differences in the magnitude of the correlational relationship may easily be attributable to the fact that the variability in MUDR was averaged over 16,000 MUs across force levels ranging from 20 to 100% MVC, compared to 220 MUs with the force level limited to 10% MVC in the study by Negro and colleagues (71).

Motor Unit Variables and Muscle Force

There are several unique findings in the present study. First, females had higher MUDRs below 100% MVC where males surpassed them at maximal effort. The same was true for the occurrence of doublets, with females having dramatically greater occurrences compared to the males submaximally. Below 100% MVC, females also recruited more MUs. Overall, there was no significant differences in MUP P-P amplitude between males and females, however, females had more MUP P-P amplitudes of 1.5 mV or less below 100% MVC. The demarcation line of 1.5 mV was selected because 90% of the MUs at 20% MVC had that P-P amplitude, and there were distinct changes in the bivariate frequency-distribution curve relative to that specific P-P amplitude.

The dramatic increase in MUDR at 100% MVC for males was associated with a commensurate rise in the number of MUs detected at 80% MVC and above. The bivariate

frequency-distribution for MUP P-P amplitude and MUDRs revealed that these later recruited MUs at or below 1.5 mV had higher discharge rates relative to the mean. It is reasonable to suggest that they were high threshold MUs but the smaller P-P amplitudes were because they were farther away from the needle electrode (6). Between 20 and 80% MVC, the recruitment of MUs farther from the electrode, decreased the average MUP P-P amplitude. At 100% MVC, the signal strength of the larger MUs was evident in the colour maps of the bivariate distribution of MUP P-P amplitude-discharge rate. Their numbers were sufficient to increase the mean MUP P-P amplitude between 80 and 100% MVC. The higher (>1.5 mV) amplitude MUs were less than 20% of the total number of active MUs.

The bivariate frequency-distribution for MUP P-P amplitude and MUDRs in the present work are entirely consistent with other findings for the tibialis anterior (12, 27, 94) and more recently for the gastrocnemii (32, 50). Changes in the bivariate frequency-distribution for MUP P-P amplitude and MUDR across force levels suggest that higher MUDRs and the additional recruitment of high threshold MUs may explain the higher force output for males beyond muscle size (25, 32, 49, 50). It is interesting to note that these later recruited MUs do not conform to 'onion skin' theory (20).

The 'onion skin' theory suggests that the earlier recruited MUs will discharge at higher rates compared to later recruited, higher threshold MUs (20). We did not track individual MUs across force levels to rule out the possibility that the same MUs were discharging at higher rates. However, between 80 and 100% MVC, the later recruited MUs had distinctly larger P-P amplitudes and higher discharge rates, as evidenced in the right panels of **Figure 2**. Similar findings have been reported by others (1, 36, 41, 47, 68,

78) and has been referred to as the ‘reverse onion skin’. Later recruited high threshold MUs would require a greater discharge rate to achieve fused tetanus (35). Therefore, as force output increases throughout the full range of force output, as was the case in the current study, it would be expected that the later recruited high threshold MUs would discharge at greater rates between 80 and 100% MVC compared to earlier recruited low threshold motor units (78).

The differences between males and females across submaximal force levels may be the result of a neural strategy during submaximal efforts to compensate for a biomechanical difference when the muscle length and pennation angle are not optimal. Manal and colleagues (58) examined sex differences in the pennation angle of multiple muscles, both upper and lower limb. They reported that when going from rest to maximal effort there is a sex difference in the pennation angle as it moves towards optimal in order to produce maximal tension down the tendons line of action. At rest the TAs pennation angle for males and females are within half a degree of one another (9.3° and 8.9°, respectively). However, as tension increases to maximum the differences in pennation angle increase to over 2 degrees (males: 14.3°; females: 12.1°). Although small (Δ 15%), this difference is significant in regard to the production of tension. A greater pennation angle allows for a greater physiological cross-sectional area in the active muscle, which subsequently arranges more contractile fibers in parallel to contribute to force production (57, 59).

Additionally, with the ankle joint being placed at 20° of plantarflexion this may further disadvantage the female’s pennation angle. Changes in ankle joint angle can also result in either the lengthening or shortening of muscle length. It has been suggested that

changes in muscle length result in either increases or decreases in MUDR (3, 30, 64). In a paper by Mela and colleagues (64), they placed the ankle joint at three angles, 9° dorsiflexion, 16° and 44° plantarflexion, to investigate joint moment outputs at various stimulation frequencies elicited at the peroneal nerve. The authors suggested that based on the observation of a leftward shift of the normalized moment/frequency curve that when moving from a shortened (9° dorsiflexion) to lengthened (16° and 44° plantarflexion) muscle length, there is a shift as the muscle moves towards an optimal muscle length, thus requiring a lower stimulation frequency to achieve the same moment output. Therefore, a suboptimal joint angle in our female group may have required an increase in MUDR in order to achieve the desired level of submaximal muscle tension at our test position of 20° plantarflexion.

Muscle fibers are at an optimal length when the muscle is fully activated (58, 59). Based on the work by Manal and colleagues (58) and Mela and associates (64) it is possible that the female's TA pennation angle and muscle length were less optimal in the test position. Perhaps the differences in submaximal MU behaviour may have arisen as a compensation mechanism to overcome a less optimal ankle joint angle due to sex differences from both musculoskeletal and biomechanical factors, that may not be seen at maximum.

It is also possible that sex differences in MU variables may reflect a neural strategy to overcome a greater tendency towards greater joint laxity or tendon compliance (29, 33, 34, 43, 46, 51, 76, 77, 87). Our previous work showed that females had a 36.6% greater electromechanical delay (EMD) than males during evoked contractions of the TA, in the same testing device and with the ankle placed in the same plantarflexed position.

However, males and females were nearly identical (Δ 0.4%) with respect to maximal voluntary EMD. It was suggested that the nearly identical voluntary EMDs resulted from a 34.7% greater rate of increase in sEMG activity. The compensation strategy was most likely be due to a greater occurrence of doublets as observed in the present study (93). In the current study, females had a significantly greater (25.5%) occurrence of doublet discharges during the rate of force development, similar to the remaining portion of the contraction. Doublet discharges would facilitate the uptake of slack in the musculoskeletal-tendon unit and a more efficient transmission of force (61).

Oscillations in the common synaptic input to the MU can explain a large portion of fluctuations in force that are influenced by descending commands (69, 70). Synaptic noise is the result of motoneurons receiving synaptic inputs from thousands of connections. These numerous inputs can lead to variability in action potential discharge times as the motoneuron membrane potential fluctuates (2, 23). Random fluctuation in the membrane potential or 'synaptic noise' can be observed in the CV in the motoneuron inter-spike interval (ISI) (23). However, the variability in the ISI can also be influenced by oscillations in common synaptic inputs to the motoneuron population (21). It has been previously suggested that force steadiness is associated with either the ISI of motor unit discharge times (24, 68) or 'motoneuron noise', which is the common low-frequency oscillations in neural drive (motor unit action potential trains, MUAPT) (71).

It is likely that the reason females exhibited less force steadiness than males was a result of a neural strategy to compensate for either a less than optimal pennation angle during testing or a tendency towards greater tendon compliance and laxity at the ankle joint. The higher MUDR and incidence of doublets increased the summation of twitch

force twitches in a way that reduced force steadiness (98). Since the standard deviation of MUDR is linked with the magnitude of the scores, the observed correlation ($r = 0.56$, $p < 0.01$) between the standard deviation of MUDR and RMS_{error}, provides indirect support for a link between MUDR and force error. Thus, we suggest it is a biomechanical difference that leads to a neural compensation strategy that manifests itself as a difference in force variability, not due to an inherent difference in neural drive to the muscles. A similar observation was made by Ives and associates (45) with respect to deceleration of the upper limb during isotonic, maximal effort elbow flexion against an inertial resistance. In this case, sex differences in the strength of the antagonist muscle group resulted in a compensatory neural strategy in the way in which females decelerated the limb.

In summary, females had greater MUDR, MU recruitment, and incidence of doublet discharges below 100% MVC, where males exhibited a dramatic increase either reaching parity or surpassing females on these same variables. Females had an overall greater level of force variability, and there was a significant correlation between RMS_{error} and the standard deviation of MUDR. Demographic data suggest that these differences do not arise from differences in physical activity levels or resistance exercise of the lower limb. It was hypothesized that females invoked a neural strategy to compensate for musculoskeletal differences that resulted in lower force steadiness than males. Thus, sex differences in the gradation of force and force steadiness were not due to inherent differences in neural drive as seen in the lack of difference in the RMS amplitude. However, may be due to neural compensation during the submaximal force outputs to achieve fused tetanus.

Table 1. Subject demographics, anthropometrics and physical activity.

	Females (*N*=24) M ± SD	Males (*N*=24) M ± SD	Percent Difference	*t*-ratios and *p*-values
Age (years)	21.54 ± 1.69	22.00 ± 2.06	2.1	
Height (m)	1.68 ± 0.05	1.82 ± 0.06	7.7*	t = -9.99; $p < 0.01$
Mass (kg)	64.86 ± 6.97	81.45 ± 11.46	20.4*	t = -6.16; $p < 0.01$
Body Mass Index (kg/m^2)	23.31 ± 2.31	24.59 ± 2.85	5.2	t = -1.73; $p = 0.09$
Leg Length (cm)	36.33 ± 2.24	39.50 ± 2.10	8.0	t = -5.06; $p < 0.01$
Leg Girth (cm)	36.92 ± 2.09	37.92 ± 2.98	2.6	t = -1.35; $p = 0.18$
Foot Length (cm)	23.73 ± 0.93	26.44 ± 1.21	10.2*	t = -8.69; $p < 0.01$
Lateral malleolus to bottom of the foot length (cm)	7.27 ± 0.74	8.00 ± 1.12	9.1*	t = -9.50; $p < 0.01$
Lateral malleolus to metatarsals length (cm)	12.02 ± 1.71	12.50 ± 1.78	3.8	t = -0.95; $p = 0.34$
Calcaneus to metatarsals length	15.25 ± 1.51	17.04 ± 2.14	10.5*	t = -3.35; $p = 0.01$
Physical activity (hours/week)	4.52 ± 1.36	4.02 ± 2.05	11%	t = 1.00; $p = 0.32$
Weight-lifting (hours/week)	2.67 ± 2.46	5.24 ± 3.13	49%*	t = -3.16; $p < 0.01$
Ratio of lower body weight-lifting (%)	48.82 ± 14.53	43.50 ± 14.61	11%	t = 1.11; $p = 0.28$

*Significant at the p<0.01 probability level. All values are means ± standard deviation (M ± SD).

$$Percent\ Difference = \left(1 - \frac{Smaller\ Value}{Larger\ Value}\right) \times 100$$

Table 2. Force, force error, normalized force error and the coefficient of variation of the motor unit discharge rate.

Percent MVC		**Force** (Newtons) (M ± SD)		**Force Error** (Newtons) (M ± SD)		**Force Error** (Normalized) (M ± SD)		**MUDR CV** (M ± SD)	
		Female	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**
20		29.50 ± 8.06	43.37 ± 10.91	0.95 ± 0.62	0.95 ± 0.54	2.72 ± 1.90	1.88 ± 1.02	7.84 ± 2.23	7.26 ± 2.27
40		55.54 ± 15.14	83.82 ± 18.93	1.49 ± 1.36	1.49 ± 1.07	2.23 ± 1.80	1.60 ± 1.00	8.45 ± 2.30	6.89 ± 1.68
60		80.00 ± 21.29	125.72 ± 27.31	1.99 ± 1.11	2.38 ± 1.30	2.22 ± 1.24	1.77 ± 1.15	8.71 ± 2.39	7.28 ± 1.70
80		105.98 ± 27.67	163.21 ± 38.78	2.94 ± 1.97	3.69 ± 2.11	2.50 ± 1.66	2.10 ± 1.17	8.28 ± 2.13	7.59 ± 1.78
100		128.07 ± 35.63	199.37 ± 47.93	3.23 ± 1.68	6.68 ± 15.48	2.50 ± 1.77	2.91 ± 5.22	8.72 ± 2.13	8.11 ± 1.45
***F*-ratios**	*df*								
Sex	[1,46]	36.38*		4.74*		4.25*		4.08*	
Force	[4,184]	1324.71*		14.40*		4.56*		3.07*	
Sex × Force	[4,184]	20.79*		2.97*		1.89		1.60	

ABOVE: All values are means ± standard deviation (M ± SD). Force error was calculated from the Root-Mean-Square error (RMS_{error}) and normalized to force in Newtons ($nRMS_{error}$). BELOW: ANOVA *F*-ratios for Sex, Force and the Sex by Force interaction. *Significant at the $p<0.01$ probability level.

Table 3. Motor unit behaviour and the peak-to-peak amplitude of the motor unit potentials.

Percent MVC		**MUDR (**pulses/sec) (M ± SD)		**Doublets** (Counts) (M ± SD)		**MU Number** (Counts) (M ± SD)		**MUP P-P Amplitude** (mV) (M ± SD)	
		Female	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**
20		17.30 ± 3.22	15.36 ± 2.99	42.56 ± 74.35	25.60 ± 91.71	16.00 ± 6.35	12.72 ± 3.92	0.698 ± 0.327	0.859 ± 0.704
40		20.50 ± 4.70	18.98 ± 3.01	80.99 ± 141.71	35.30 ± 66.55	18.62 ± 5.60	17.36 ± 5.30	0.608 ± 0.248	0.733 ± 0.413
60		22.62 ± 3.78	21.94 ± 3.61	102.11 ± 130.42	56.36 ± 114.57	22.78 ± 5.36	20.63 ± 5.93	0.538 ± 0.177	0.638 ± 0.315
80		24.52 ± 3.90	23.66 ± 2.82	149.38 ± 136.38	61.70 ± 71.58	22.26 ± 5.13	24.62 ± 6.12	0.523 ± 0.167	0.555 ± 0.250
100		26.11 ± 4.84	27.68 ± 4.47	165.82 ± 146.66	165.25 ± 167.86	26.26 ± 6.78	26.32 ± 6.11	0.538 ± 0.185	0.649 ± 0.466
ANOVA *F*-ratios	***df***								
Sex	[1,46]	0.08		4.55*		0.90		0.62	
Force	[4,184]	148.07*		69.12*		97.03*		9.46*	
Sex × Force	[4,184]	4.17*		3.43*		4.36*		0.40	

ABOVE: All values are means ± standard deviation (M ± SD). Motor unit discharge rate (MUDR pps), incidence of doublet discharges (Doublets), motor units recruited (MU Number) and motor unit potential peak-to-peak amplitude in millivolts (MUP P-P Amplitude mV). BELOW: ANOVA *F*-ratios for Sex, Force and the Sex by Force interaction. *Significant at the $p<0.01$ probability level.

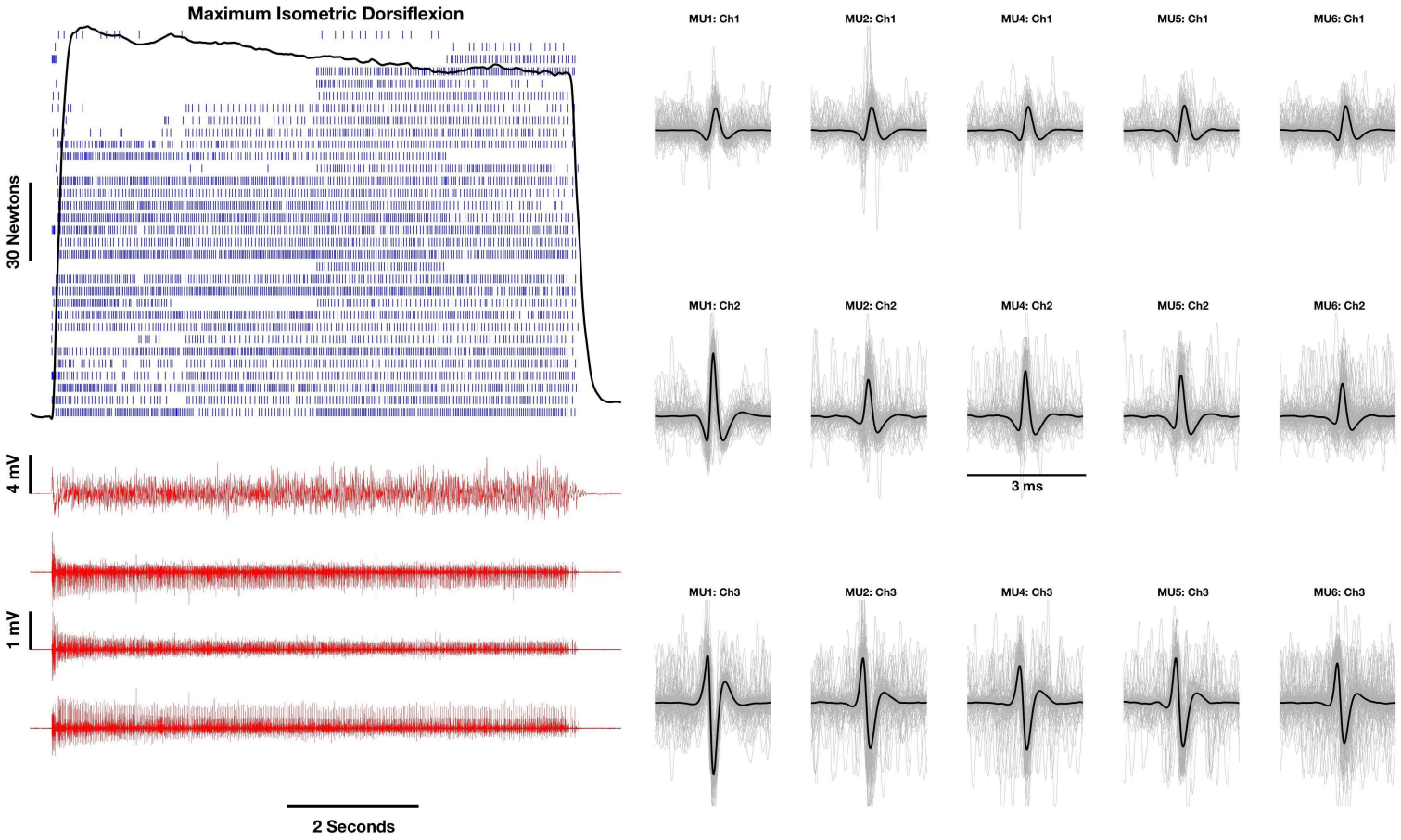

Figure 1. Representative trial at 100% MVC. UPPER LEFT: Black line represents the force output, vertical blue lines represent each identified discharge from each motor unit (n=25). LOWER LEFT: top red trace is the surface interference pattern and the three red traces below are the intramuscular interference pattern recorded from each channel. Pooled motor unit discharge rates were calculated a two second epoch based on the most stable portion of the force trace. RIGHT: Five sample motor units identified across all three channels. Greyed portion is the shimmer of each motor unit discharge throughout the entire trial. Black line is the average of all of the motor unit discharges.

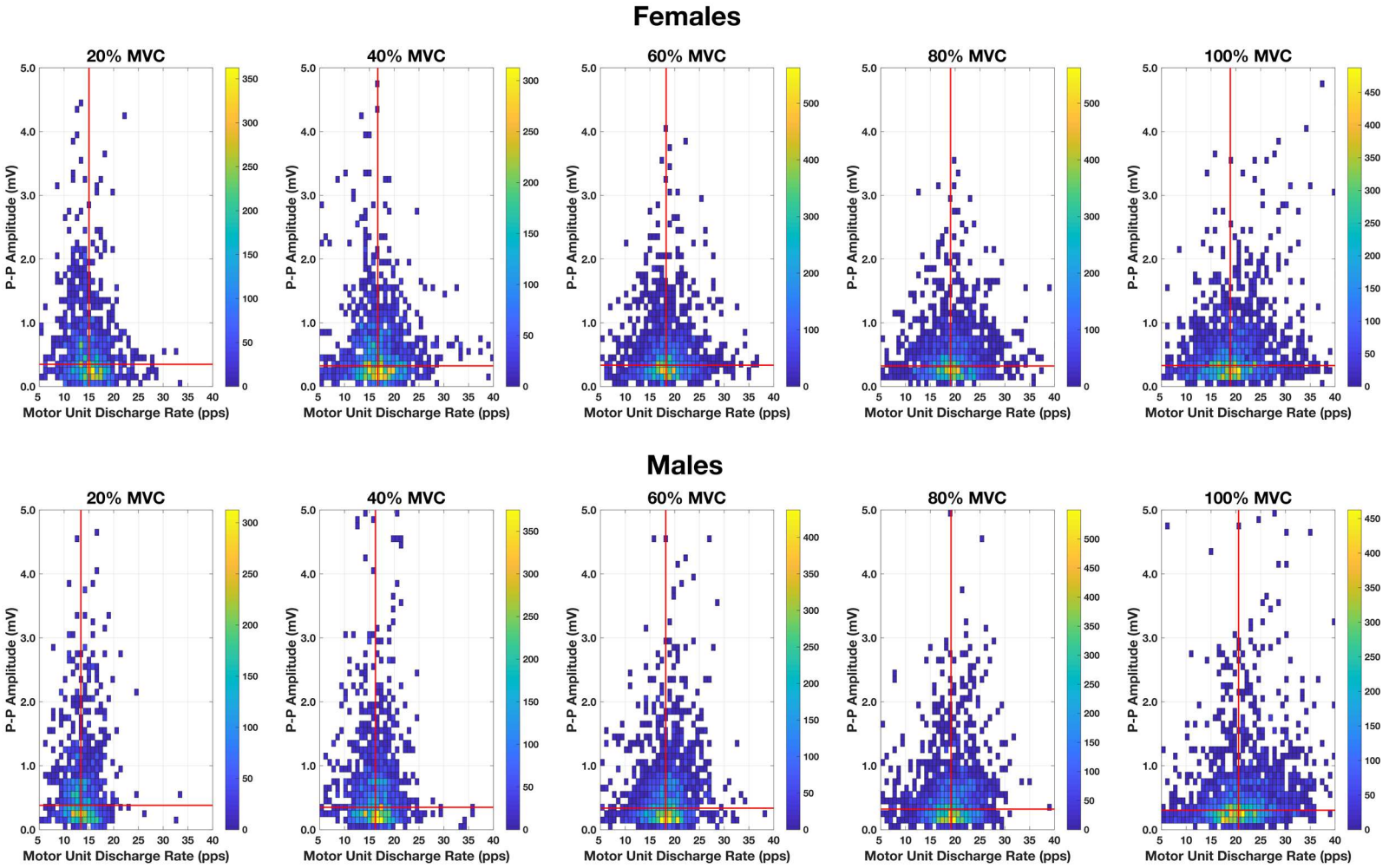

Figure 2. Bivariate frequency-distributions for motor unit potential (MUP) peak-to-peak amplitude (P-P) and its discharge rate across all force levels for females and males. The colour map shows the number of MUs within each P-P amplitude-discharge rate bin. Red cross-hair in each panel illustrates the bivariate mean.

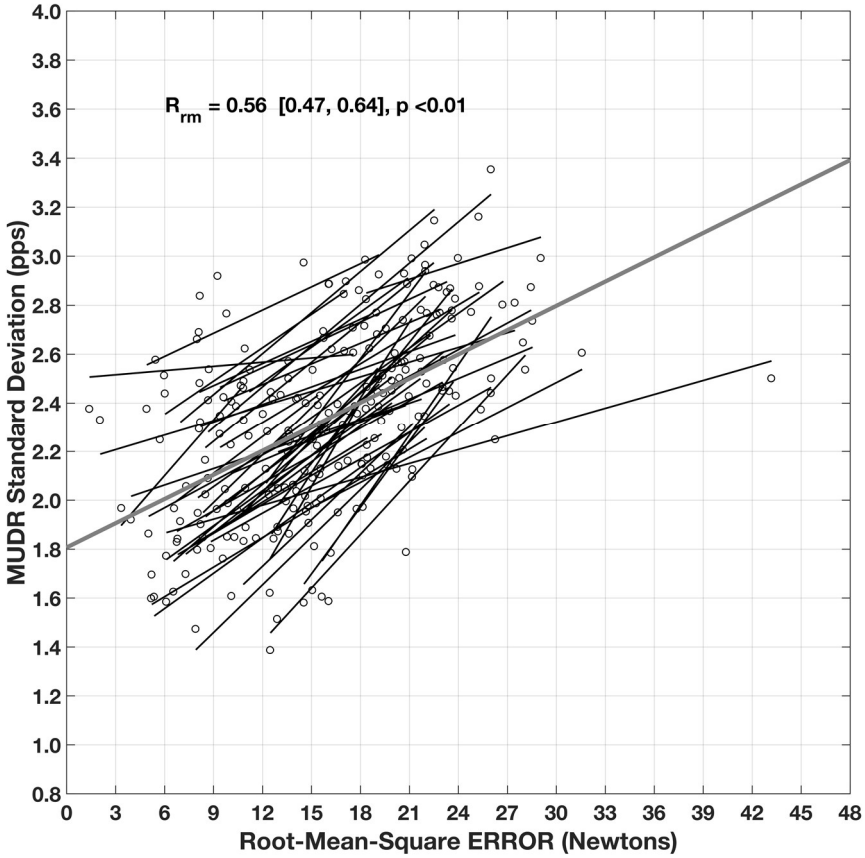

Figure 3. Correlation between the root-mean-square error and the standard deviation of motor unit discharge rate (MUDR). Black lines correspond to individual linear correlations and the grey line is the overall linear correlation.

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7 GENERAL DISCUSSION

Sex differences have been identified in both the upper and lower body and are usually attributed to differences in muscle size or activities of daily living (33, 35, 67). The study of sex differences provides insight into not only the size differences attributing to greater strength in males, but also the mechanisms behind these differences that go beyond anthropometrics or daily physical activity. Previous literature has indicated that sex differences in muscle strength are considerably greater in the upper versus lower limb. This makes the upper limb an ideal model to establish the best statistical approach for hypothesis testing and correct data interpretation. In the lower limb, where the sex differences in size and strength are not as great (33, 35, 55), potential neural and muscular mechanisms may be responsible for the remaining differences in strength, that go beyond size (4–7, 10, 20, 47, 63, 65). It has been suggested that males have a greater tension generating capacity due to differences in muscle activation, inferred from the sEMG signal (12, 29, 62). However, compared to males, females may rely more on neuromuscular adaptations than hypertrophy when adapting to resistive exercise (15). It is not known whether or not sex differences in MUDRs exist during the gradation of force, or if they utilize different neural strategies involving both rate-coding and motor unit recruitment as suggested by sEMG data.

The overall purpose of this dissertation was to further the study of sex differences in strength and force development in relation to the neural control of muscle while establishing the most appropriate approach to data analysis, observing the effect of task familiarization and the effect of force output on neural control. The utilization of both surface and intramuscular EMG allowed for the investigation of neural drive and motor unit behaviour in relation to force gradation and force steadiness. Sex differences have

been studied previously, but the literature lacks consistency in statistical approach and interpretation. Therefore, manuscript 1 examined sex differences in maximal isometric strength and the rate of force development in the elbow flexors where the magnitude of the differences is greatest to reveal the impact of the analysis of approach. Manuscript 2 compared neural (central) and muscular (peripheral) contributions to the maximal rate of tension development through comparing maximal voluntary versus evoked muscle contractions in the dorsiflexors. This was done to identify specific neural or muscular mechanisms responsible for sex differences, beyond muscle size. Finally, manuscript 3 directly assessed sex differences in neural control throughout a range of force gradation by recording motor unit activity. The purpose of direct measurement of the motor unit behaviour through a large range of force output was to determine if sex differences at maximal effort are due to a neural constraint or are present as part of an overall sex difference in force regulation across the continuum.

7.1 Summary of Findings

The first study looked at the sex differences in the rate of force development in the upper limb and tested two statistical approaches to the data analysis. Prevalent methods of data analysis of sex-related differences in muscle contractile characteristics are: (a) statistical control using an analysis of covariance, and (b) normalization by the creation of ratio scores prior to statistical analysis (4, 30, 33, 35, 36). Each method was used to analyze the data to see if they could account for the sex differences in the rate of force development. Each participant performed 3 voluntary isometric contractions of the elbow flexors and both force output and monopolar surface EMG was recorded from the biceps brachii. The rate of force development was calculated according to Andersen and

Aagaard (2) using the rate of increase in force output across 20 ms sequential non-overlapping epochs. The peak was the epoch with the greatest rate of increase. The rate of muscle activation (Q_{30}) was calculated according to the methods outlined by Gottlieb and colleagues (24) by rectifying the raw EMG signal and numerically integrating it to get the rate of increase in sEMG over the first 30 ms.

As might be expected, males exhibited greater (55.5%) maximal isometric elbow flexion force than females. The correlation between maximum force and biceps cross-sectional area was $r = 0.65$ ($p < 0.05$) for females and $r = 0.63$ ($p < 0.05$) for males. The peak rate of force development (RFD) was 61.2% greater for males than for females. Statistical control for maximum strength reduced sex differences to 36.7%, which was still significant at the 0.05 probability level ($p = 0.02$). The correlation between maximum strength and maximum RFD was $r = 0.52$ ($p < 0.05$) for females and $r = 0.39$ ($p < 0.05$) for males. Similarly, the correlation between Q_{30} and maximum RFD was $r = 0.50$ ($p < 0.05$) for females and $r = 0.46$ ($p < 0.05$) for males and it was a significant covariate in the ANCOVA model ($p < 0.05$). However, the difference between males and females dropped to only 54.4%, which remained well below the 0.05 probability level ($p = 0.0001$). A two-covariate model that included both maximum force and Q_{30}, reduced the difference between males and females to 13.8%, which was above the 0.05 probability level ($p = 0.39$).

The use of a ratio score by dividing the maximum RFD by maximum force, reduced the difference between males and females to 15.9%, which was no longer statistically significant ($p = 0.08$). Normalization of maximum RFD by using Q_{30} as the denominator had an even greater impact. The difference between males and females was

reduced to 9.8% which was well above the 0.05 probability level ($p = 0.66$). A reasonable case could be made for normalization by either variable, leading to a more ambiguous interpretation of the results. The statistical controls through an ANCOVA revealed that neither maximum force nor Q_{30} could completely account for sex differences in the rate of force development. When these variables were combined, they were however able to eliminate statistically significant the sex differences. More importantly, sex differences (13.8%) still persisted despite statistical controls. The main findings of study 1 were that the findings themselves depend on the method of data analysis.

There was also a moderate correlation between the measure of muscle size and its strength. This led to the argument regarding the role of neural factors in the expression of muscular strength (17, 46, 56). Past research has provided evidence that males and females employ different neural strategies for the control of agonist, antagonist, and synergists during maximal voluntary contractions (1, 19, 23, 39, 41). Persistence of sex differences in the rate of force development despite the additional inclusion of Q_{30} as a covariate indicated that other neural factors were involved. The measure Q_{30} has been associated with motor unit discharge rate at the onset of muscle contraction (66). This led to the suggestion that males in the study had a higher motor unit discharge rate at the onset of muscle contraction, which contributed to the observed sex differences in the rate of force development. Therefore, the first study also showed that the known linkages between rate of force development explain only a portion of the differences between males and females, while additional neural and or biomechanical factors may be involved.

The purpose of Study 2 was to determine if sex differences in the maximum rate of isometric dorsiflexion force development are determined by maximum isometric dorsiflexion force alone, or other factors as observed for the upper limb where the strength differences are more pronounced. Based on the work of Hannah and associates (30), it was hypothesized that maximum isometric strength would account for sex differences in the maximum rate of force development during dorsiflexion, because males and females are more comparable in maximum isometric strength in the lower limb. Therefore, Study 2 sought to investigate if sex differences in the maximum rate of force development in distal muscles (tibialis anterior) required for balance are a result of either central or peripheral factors. The role of task familiarization on the ability to achieve a true maximum effort contraction was also assessed. Studying mechanisms behind sex differences in distal muscles that are responsible for balance and explosive activity can guide specific training interventions or rehabilitation techniques.

Each participant performed three maximal isometric voluntary contractions, and through the stimulation of the peroneal nerve produced 10 isometric compound muscle action potentials on three separate days. Both force output and sEMG were recorded simultaneously with a three-bar surface recording electrode placed one cm distal to the most proximal electrically identified tibialis anterior motor point, which produced two bipolar recordings of the myoelectric signal.

We found that the consistency of scores within subjects for the criterion measures obtained during the voluntary contractions for females was excellent (R = 0.76–92) except for EMD which had an intraclass correlation coefficient of R = 0.35. Female participants exhibited a 27.2% reduction in EMD from session 1 to session 3 (p =

0.0045). The means of the criterion measures generated during voluntary contractions were highly stable across test sessions in males, while the consistency of scores within subjects ranged from fair to excellent (R = 0.61–83). The same was true for the criterion measures generated during evoked contractions (Females: R = 0.85–95; Males: R = 0.57–96).

Males had on average a 50.6% greater force output than females. The difference between males and females with respect to the maximum rate of force development was similar in magnitude. The maximum rate of force development was 44.6% greater in males than for females (p = 0.0003). In contrast, females had a 34.7% greater rate of increase in muscle activation as assessed by Q_{30} (p = 0.0454). There were no significant differences between males and females with respect to EMD, RMS amplitude, or MFCV (p = 0.9635; p = 0.5834; p = 0.3598, respectively).

Similar to the voluntary contractions, maximal evoked dorsiflexion force was 53.9% greater for males than for females (p = 0.001), and males also had a comparably greater maximum rate of force development of 55.4% (p = 0.0006). The EMD was shorter for males than for females during the evoked contractions (36.6%, p = 0.0119). However, MFCV, evoked CT, and Vpp of the CMAP were not significantly different between groups (p = 0.6451; p = 0.8589; p = 0.1288, respectively).

When maximal force was used as the covariate in the repeated-measures ANCOVA for the maximum rate of force development, the difference between means decreased to 1.2% (p = 0.9264). The rate of increase in muscle activation (Q_{30}) was not assessed as a covariate, because females were actually greater than males. The only other significant difference between males and females was EMD during evoked contractions. However,

evoked EMD had little impact as a covariate as the sex-related differences in the maximum rate of force development were reduced to 33.8%, which was still significant ($p = 0.0067$).

Study 2 showed that maximum force was able to account for almost all of the sex differences in the maximum rate of force development as the percent differences was reduced to 1.2%. Contraction time was used to determine if differences in the maximum rate of force development was associated with either descending voluntary control or peripheral factors associated with muscle composition (13), which was not observed in the study. This suggests that there were no sex differences in the muscle fiber composition (34).

With sex differences in muscle contractile characteristics eliminated, there were several novel findings in Study 2 that added to the understanding of sex differences in the rate of force development. First, there were no sex differences in MFCV which may be due to comparable muscle fiber diameters (49, 50, 54, 59, 69). Second, the pronounced differences in Q_{30} were not evident in either the voluntary (RMS) or evoked (Vpp) myoelectric signal magnitude, which eliminates the possibility of a confound in the interpretation of sex differences in the rate of muscle activation. The tibialis anterior in females had a significantly greater rate of increase in sEMG as assessed by Q_{30}. Females also exhibited a ~ 4% increase across the three days of testing, which was associated with a decrease in voluntary EMD. The lack of change in evoked EMD of females, highlights a voluntary control strategy to change neural drive at the onset of force production (21). However, it is not possible to distinguish the exact motor unit behaviour based on the recordings of the myoelectric signal from the skins surface, but Van Cutsem and

colleagues (66) established a link between the maximum rate of force development, the rate of increase in EMG and an increase in the incidence of double discharges due to training adaptations. This suggests that females may utilize a different motor unit activation pattern compared to males at the onset of contraction.

Harwood and colleagues (31) reported that in the elbow flexors females had a 12% lower MUDR when compared to males during a 15% MVC submaximal trial. It was also reported by Christie and Kamen (11) that short-term training adaptations in the tibialis anterior resulted in a 9% greater MUDR for males compared to females at 100% MVC. Taken together, the two studies cited above, there may be sex differences in the gradation of muscle force. If for example, females 'generally' have lower MUDRs, are there other neural strategies (i.e., doublets or recruitment) that compensate during the force gradation process? Therefore, Study 3 sought to extend the findings of sex differences in neural variables through the assessment of differences in sEMG activity reported in Study 1 and 2, by directly measuring motor unit variables throughout the gradation of muscle force.

Each participant was asked to perform three voluntarily isometric dorsiflexion contractions at 20, 40, 60, 80 and 100% MVC. Force output was recorded along with both sEMG and intramuscular EMG with a quadrifillar intramuscular electrode. The monopolar surface EMG electrode was placed on the most proximal electrically identified tibialis anterior motor point. The intramuscular electrode was placed 1 cm distal to the surface electrode.

For females, 90% of the MUP P-P amplitudes were at or below 1.5 mV at 20% MVC, compared to 88% for males. Between 20 and 100% MVC males increased the number of MUPs with P-P amplitudes below 1.5 mV, so that sex differences disappeared

by 60% MVC. The later recruited MUPs also had higher MUDRs ($p < 0.01$). Females exhibited the same pattern of change, but to a lesser degree ($p < 0.01$). There was also a pronounced increase in the number of MUs with P-P amplitudes greater than 1.5 mV with higher discharge rates in both males and females, but they accounted for less than 20% of the total MU population. Females had greater MUDR than males below 100% MVC (Δ 3%, $p < 0.05$), whereupon males had a dramatic increase between 80 and 100% MVC and surpassed them (Δ 8%, $p < 0.05$). The same was true for the occurrence of doublets (Δ 36.36%, $p < 0.05$). Finally, the normalized RMS_{error} for females was 16% greater and the coefficient of variation in MUDR was 11.5% greater than the males.

The present work is the first study to document differences in MU behavior between males and females with respect to the force gradation process. First, changes in the bivariate frequency distribution for MUP P-P amplitude and MUDR across force levels suggest that higher MUDRs and the additional recruitment of high threshold MUs may explain the higher force output for males beyond muscle size (18, 25, 42, 44). Notable, the later recruited MUs for both sexes had distinctly larger P-P amplitudes and higher discharge rates, which is similar to findings reported by others (3, 28, 37, 40, 57, 61) and has been referred to as the 'reverse onion skin'. Thus, Study 3 provided definitive evidence that the 'onion skin' relationship between MU recruitment and MUDR is not necessarily the rule, which has been a long-held neurophysiological principle.

The details on MU behavior during the gradation of muscle force is both novel and of high impact. Below 100% MVC, females had higher MUDRs, recruited more MUs at each force level, and had a greater incidence of doublets than males. Higher MUDR and incidence of doublets increases the summation of twitch force twitches and can reduce

force steadiness, in a manner similar to what has been shown with synchronization (68). As a result, females were observed to have less force steadiness with 11.5% greater normalized RMS_{error} than males. Since the standard deviation of MUDR is linked with the magnitude of the scores, the observed correlation ($r = 0.56$, $p < 0.01$) between the standard deviation of MUDR and RMS_{error}, provides indirect support for a link between MUDR and force error.

Demographic data suggest that sex differences do not arise from differences in physical activity levels or resistance exercise of the lower limb. Rather, the higher MUDRs, incidence of doublets, and recruitment of MUs was most likely a neural compensation mechanism for either sex differences in pennation angle (51–53) or a greater tendency towards greater joint laxity in (22, 26, 27, 38, 45). At the very least, the longer evoked EMD for females observed in Study 2, suggest greater compliance in the muscle-tendon unit. Sex differences in the gradation of force and force steadiness were, therefore, not due to inherent differences in neural drive to the muscle. A similar observation was made by Ives and associates (39) with respect to deceleration of the upper limb during isotonic, maximal effort elbow flexion against an inertial resistance. In this case, sex differences in the strength of the antagonist muscle group resulted in a compensatory neural strategy in the way in which females decelerated the limb.

In summary, females had greater MUDR, MU recruitment, and incidence of doublet discharges below 100% MVC, where upon males exhibited dramatic increase to either reach parity or surpass females on these same variables. Females had an overall greater level of force variability, and there was a correlational link between the RMS_{error} and the standard deviation of MUDR. Demographic data suggest that sex differences do

not arise from differences in physical activity levels or resistance exercise of the lower limb. It was hypothesized that females invoked a neural strategy to compensate for musculoskeletal differences that resulted in lower force steadiness than males. Thus, sex differences in the gradation of force and force steadiness were not due inherent differences in neural drive to the muscle.

This dissertation provided insight and future directions not only to sex differences in the neural control of muscle, but also suggested that structural differences can lead to neural compensation to achieve a desired force output. First, the use of the most appropriate statistical analysis technique was established to facilitate a parsimonious interpretation of the results. The importance of multiple familiarization sessions was highlighted so that sex differences in performance were not associated with the task itself. Multiple familiarization sessions also revealed sex differences in neural strategy do exist as females increased the rate of muscle activation (Q_{30}) to overcome potential structural or biomechanical differences.

The suggested differences in neural drive seen in the measurement of Q_{30} from the sEMG signal of the tibialis anterior in Study 2 was confirmed through direct indwelling EMG measurement of MU variables in Study 3. It is clear at submaximal force outputs females recruit more motor units, have a greater motor unit discharge rate and utilize different recruitment strategies, such as doublet discharges, to compensate for potential sex differences in muscle architecture, musculotendinous stiffness and structural differences in the joint itself. The longer evoked EMD for females observed in Study 2, suggest greater compliance in muscle-tendon unit. It is clear from all of the findings that sex differences in motor unit behaviour are not a result of neural constraints (39), but are

actually due to neural accommodation for musculoskeletal biomechanics. We suggest the neural compensation strategy also manifests itself as a difference in force variability, and not due to an inherent difference in neural drive to the muscles. These neuromechanical sex differences in distal muscles that are responsible for balance, posture and explosive activity can guide specific training interventions or rehabilitation techniques.

It must be stated that when interpreting the motor unit potentials recorded from the intramuscular electrode caution must be applied when distinguishing the inferred threshold based on MUP amplitude alone. As contraction intensities increase so does the motor unit size and discharge rate (48). This can be governed by: (1) physiological factors according to the size principle (32), (2) electrophysiological factors, such as the distance between the recording electrode and the active muscle fibers (58) and (3) technical factors related to the decomposition of the interference pattern (8). The contribution of lower amplitude motor units to the interference pattern during high force outputs can be masked by either superposition of motor units or the size of larger motor units (9, 14, 58). Generally, the detected motor units from a needle electrode represents the active motor units within 0.5 mm of the detection surface (8). Therefore, the detection of higher amplitude motor units may not be direct evidence of the recruitment of high threshold motor units but may be the result of motor unit distance from the recording surface.

Additionally, as has been reported previously by Kawakami and colleagues (43) that muscle fiber pennation angles increase in hypertrophied muscle. With this in mind, it is possible that the sex differences reported in pennation angle of the tibialis anterior (52) may be a result of males larger muscle cross-sectional area. Therefore, the sex differences

in motor unit behaviour during submaximal force outputs found in the third study may in fact be more related to the joint and musculotendonous stiffness (60, 64) separate from muscle size and pennation angle.

7.2 Limitations

Surface EMG can only provide an indirect measurement of differences and in underlying changes in motor unit activity patterns. Additionally, not measuring motor unit synchronization may result in an incomplete understanding of the mechanisms responsible for force steadiness. Although the possible influence of different muscle architecture, size and stiffness in males versus females was discussed in the dissertation, not directly measuring it with ultrasound is a limitation. It was also assumed that there were differences in tendon stiffness and joint laxity without having actually measured it. Moreover, caution must be applied when extrapolating the results to older adults. The use of a healthy college aged population may not account for differences between the sexes with aging, such as the loss of type II fibers, which may render the sexes more similar in the 6th and 7th decades. This may also be true regarding the generalizability of the findings from Study 1 & 2. These studies focused on contractions at maximum which may limit their application to healthy athletic populations and not directly transferrable to either clinical or elderly population. The reason is that clinical or elderly populations perform most of their activities of daily living during submaximal efforts.

7.3 Future directions

The present work has identified that sex differences in the neural control of muscle are not isolated to maximal contractions. The dissertation has led to the hypothesis that

there is an interaction between musculoskeletal biomechanics and neural control during submaximal contractions. Specific neural strategies may be employed to compensate for pennation angle, muscle length and moment arm length. Previous work by Manal and colleges (52, 53) and Maganaris and Baltzopolous (51) have looked at both joint angle and the resulting muscle pennation angles between the sexes at rest and maximum force output. However, the question remains as to how sex differences in pennation angle during the transition between rest to maximum governs MU behavior. This is important because most activities of daily living occur within this transition range. The same is true for joint stiffness, which is comprised of ligamentous restraints and muscle-tendon units of the surrounding musculature. Sex differences in joint stiffness and muscle activity have been linked with injury mechanisms during dynamic gross motor behavior (7, 10, 16), but there are no basic data on underlying MU behavior as reported in this dissertation. It is also important to understand how sex differences in musculoskeletal biomechanics alter MU behavior to provide insight into injury mechanisms and properly prescribe both training and rehabilitation regimen.

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APPENDIX A

Ethics Approval Study 1 and 2

Brock University
Research Ethics Office
Tel: 905-688-5550 ext. 3035
Email: reb@brocku.ca

Bioscience Research Ethics Board

Certificate of Ethics Clearance for Human Participant Research

DATE: July 7, 2011

PRINCIPAL INVESTIGATOR: GABRIEL, David A. - Physical Education & Kinesiology

FILE: 02-283 - GABRIEL

TYPE: Faculty Research STUDENT:
SUPERVISOR:

TITLE: Analysis of Surface Electromyographic Spike Activity

ETHICS CLEARANCE GRANTED

Type of Clearance: MODIFICATION Expiry Date: 5/31/2012

The Brock University Bioscience Research Ethics Board has reviewed the above named research proposal and considers the procedures, as described by the applicant, to conform to the University's ethical standards and the Tri-Council Policy Statement. Clearance granted from **7/7/2011** to **5/31/2012**.

The Tri-Council Policy Statement requires that ongoing research be monitored by, at a minimum, an annual report. Should your project extend beyond the expiry date, you are required to submit a Renewal form before **5/31/2012**. Continued clearance is contingent on timely submission of reports.

To comply with the Tri-Council Policy Statement, you must also submit a final report upon completion of your project. All report forms can be found on the Research Ethics web page at http://www.brocku.ca/research/policies-and-forms/research-forms.

In addition, throughout your research, you must report promptly to the REB:

a) Changes increasing the risk to the participant(s) and/or affecting significantly the conduct of the study;
b) All adverse and/or unanticipated experiences or events that may have real or potential unfavourable implications for participants;
c) New information that may adversely affect the safety of the participants or the conduct of the study;
d) Any changes in your source of funding or new funding to a previously unfunded project.

We wish you success with your research.

Approved:

Brian Roy, Chair
Bioscience Research Ethics Board

Note: Brock University is accountable for the research carried out in its own jurisdiction or under its auspices and may refuse certain research even though the REB has found it ethically acceptable.

If research participants are in the care of a health facility, at a school, or other institution or community organization, it is the responsibility of the Principal Investigator to ensure that the ethical guidelines and clearance of those facilities or institutions are obtained and filed with the REB prior to the initiation of research at that site.

APPENDIX B

Ethics Approval Study 3

Brock University
Research Ethics Office
Tel: 905-688-5550 ext. 3035
Email: reb@brocku.ca

Bioscience Research Ethics Board

Certificate of Ethics Clearance for Human Participant Research

DATE:	8/3/2012		
PRINCIPAL INVESTIGATOR:	GABRIEL, David - Kinesiology		
FILE:	12-027 - GABRIEL		
TYPE:	Ph. D.	STUDENT:	Greig Inglis
		SUPERVISOR:	David Gabriel

TITLE: Sex Differences in motor unit discharge rates at various force levels

ETHICS CLEARANCE GRANTED

Type of Clearance: NEW Expiry Date: 8/30/2013

The Brock University Bioscience Research Ethics Board has reviewed the above named research proposal and considers the procedures, as described by the applicant, to conform to the University's ethical standards and the Tri-Council Policy Statement. Clearance granted from 8/3/2012 to 8/30/2013.

The Tri-Council Policy Statement requires that ongoing research be monitored by, at a minimum, an annual report. Should your project extend beyond the expiry date, you are required to submit a Renewal form before 8/30/2013. Continued clearance is contingent on timely submission of reports.

To comply with the Tri-Council Policy Statement, you must also submit a final report upon completion of your project. All report forms can be found on the Research Ethics web page at http://www.brocku.ca/research/policies-and-forms/research-forms.

In addition, throughout your research, you must report promptly to the REB:

a) Changes increasing the risk to the participant(s) and/or affecting significantly the conduct of the study;
b) All adverse and/or unanticipated experiences or events that may have real or potential unfavourable implications for participants;
c) New information that may adversely affect the safety of the participants or the conduct of the study;
d) Any changes in your source of funding or new funding to a previously unfunded project.

We wish you success with your research.

Approved:

Brian Roy, Chair
Bioscience Research Ethics Board

Note: Brock University is accountable for the research carried out in its own jurisdiction or under its auspices and may refuse certain research even though the REB has found it ethically acceptable.

If research participants are in the care of a health facility, at a school, or other institution or community organization, it is the responsibility of the Principal Investigator to ensure that the ethical guidelines and clearance of those facilities or institutions are obtained and filed with the REB prior to the initiation of research at that site.

APPENDIX C

Demographic Information

Demographic Information

Age: _____

Weight: _____

Height: _____

University Major: _____

How many times a week do you weight train?_____

How many hours per week do you weight train?_____

What percentage of time weight training do you spend training:

Upper body: _____

Lower body: _____

How long have you been weight training (please circle):

0-3 months 4-6 months 7-12 months 1-5 years more than 5 years

How many times per week do you do physical activity, other than weight training? _____

Other than weight training, what other physical activity are you participating in?

_____

_____

Anthropometric Measures and EMG Data Collection Sheet

Anthropometric Measures and EMG Data Collection Sheet

Date: _____

Subject Name: _____ Age: _____ Number: _____

Gender: M F

TYPE	Measurement (cm)
Lower Leg length	
Lower Leg circumference	
Whole Foot	
Malleolus to bottom	
Malleolus to Meta Tarsals	
Calcaneus to Meta Tarsals	

EQUIPMENT SETTINGS

Amplification: Indwelling: _____ sEMG: _____

Filter Settings: Indwelling: _____ sEMG: _____

Skin Impedance Pre: _____ Post: _____

Skin Temperature Pre: _____ Post: _____

Pre MVC: _____

20% : _____

40% : _____

60% : _____

80% : _____

POST MVC : _____