CHANGES IN RECRUITMENT STRATEGY FOR VARYING JOINT ANGLE

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In settings such as rehabilitation and surgeries, it is assumed that for maximal activation at different angles, the activation dynamics of the muscle will be unchanged since the muscle is still working at its maximum for that angle. If the assumed relationship can be confirmed, muscle models will increase in validity. To investigate this assumption, this study was designed to explore if a change in activation dynamics occurs when the first dorsal interosseous (FDI) muscle is activated through a range of motion. Five males and one female were recruited to perform a series of force producing tasks while a decomposition electromyography (dEMG) machine recorded their FDI muscle activity. The tasks included maximal voluntary contraction, decomposition and following a tracking paradigm at 50% of their maximum across the angles of 0°, 5°, 10°, 15° and 20° index finger abduction. Primary results showed that statistically, length had an effect on the recruitment strategy of the muscle. Further research would be needed to infer these results on a larger population.
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Chapter 1

Introduction

Among the science community, it is known that changes in muscle length vary the maximal force that a muscle can produce through its possible range of motion (Gordon et al., 1966). It is assumed that for maximal activation at different angles, the activation dynamics of the muscle will be unchanged since the muscle is still working at its maximum for that angle (Brinckman et al, 2002). Due to lack of technology, the assumption has not been previously investigated. If the assumed relationship between activation dynamics and muscle length can be confirmed, muscle models will increase in validity.

Each motor neuron in human skeletal muscle innervates a number of muscle fibers in order to make up a motor unit (Gottlieb & Agarwal, 1971). An action potential is sent from the central nervous system down through the motor neuron that excites all of the muscle fibers within a single motor unit. The motor unit, which is the smallest functional unit of the muscle, generates a force when excited (Zajac, 1989). Multiple motor units within a single muscle are excited in order to sum together and produce the net muscle force that is used to carry out a task.

Excitation of the muscle fibers by the alpha motor neuron triggers the contractile proteins of the muscle (Gottlieb & Agarwal, 1971). The contractile mechanism of the muscle produces force, which relays a bioelectrical signal that can be recorded by an electromyogram (EMG) (Gottlieb & Agarwal, 1971). The excitation itself is known as
the activation dynamics or active state. The active state of a muscle is the amount of calcium ions that are bound to troponin or the fraction of cross bridges that are attached (Soest et al, 1995). The activation dynamics of the muscle can be used as a scaling factor for force in both the force-length and the force-velocity curves (Soest et al, 1995).

The central nervous system controls the amount of force that is produced by the muscle, but how exactly it provides this control is still unclear. There are two ways in which an increase in muscle force occurs: through varying the frequency of motor neuron firing or the number of motor units recruited (Zajac, 1989). A possible explanation that has been proposed is that the increase in muscle force could be different between muscles that produce gross motor control and muscles that produce fine motor control. Recruiting inactive motor units provide gross motor control, and varying motor unit discharge frequencies provide finer motor control (Gottlieb & Agarwal, 1971; Lenman 1969). If this were the case, the motor control strategy used would be different for every muscle of the body, depending on its size, fiber type, and task.

The first dorsal interosseous (FDI) muscle of the index finger is a popular and efficient muscle used in this type of experimental study. The muscle belly is distinct and available for the EMG, its function is unique, and its nerve supply is easily accessible (Zijdewind & Kernell, 1994). The FDI has two distinct heads that originate on the lateral, palmer surface of the second metacarpal and the medial aspect of the first metacarpal (Hudson et al, 2009; Masquelet et al, 1986). The FDI allows the index finger to both flex and abduct at the metacarpophalangeal (MCP) joint (Hudson et al, 2009).

Every component involved in the activation of the contractile mechanism is non-linearly dependent on the length of the muscle (Gottlieb & Agarwal, 1971). The amount
of maximum force that a muscle can produce will change according to the length because
the alignment of contractile proteins and the number of cross bridges that are attached
will vary (Brinckman et al, 2002). The excitation itself from the central nervous system
to the motor units of the muscle, i.e. activation dynamics, is used as a scaling factor when
force and length are held constant and assumed to not change with the change in muscle
length. The central nervous system would send the same neural signal to the muscle,
independent of its length, and result in the maximum force possible for the muscle in its
current position. If this concept can be confirmed, the validity of its use in muscle
modeling for research, surgery, and rehabilitation will increase.

The purpose of this study to investigate if a change in activation dynamics occurs
when the first dorsal interosseous (FDI) muscle is activated through a range of motion.
By having live subjects perform voluntary contractions of the FDI through a range of
motion against a force sensor while connected to the decomposition electromyography
(dEMG) system, activation dynamics at five different angles can be compared in order to
determine if a relationship exists. The hypothesis is that there will be no significant
difference in activation dynamics across a range of motion of the FDI. The prediction is
that there will be no change in activation dynamics as the FDI changes length while
moving through a 20° range of motion.
Chapter 2

Literature Review

2.1 Muscle Models

The force-length property of the muscle can be found when the maximum effort and the fiber length of the muscle being studied are held constant (Zajac, 1989). Full activation of a muscle occurs when the muscle tissue is maximally excited until the protein machinery is fully engaged (Zajac, 1989). Zajac (1989) explained that isometric active muscle force could be recorded in order to indicate muscle activation. During an isometric contraction, length and velocity are both held constant allowing for force to be the only input. It was the assumption that the activation would scale the isometric, active force-length curve when length was constant, making active force at a given length proportional to activation (Zajac, 1989). This implies that the central nervous system would send the same neural signal to the muscle, independent of its length, and result in the maximum force possible for the muscle in its current position.

The ideal model of a muscle should be able to incorporate individual motor units since it is the discharge of each single motor neuron that collectively generate force (Zajac, 1989). It is possible that the whole muscle force-length curve may have more of a curvature than the curve of each individual motor unit (Zajac, 1989). In the bilinear differential equation used by Zajac (1989),
\[ \frac{a(t)}{dt} + \left[ \frac{1}{\tau_{\text{act}}} \cdot (\beta + [1 - \beta]u(t)) \right] \cdot a(t) = \left( \frac{1}{\tau_{\text{act}}} \right) \cdot a(t) \]

when the muscle is fully activated and activation dynamics \((u(t))\) is held constant, the rate constant \((\frac{1}{\tau_{\text{act}}} \cdot (\beta + [1 - \beta]u(t)))\), time constant \((\frac{1}{\tau_{\text{act}}})\), and motor unit excitation \((a(t))\) all increase linearly in order to increase the EMG signal before leveling off. The time it takes the motor unit to level off may be different between all individual motor units and produce an average curve when they are summed together. Past models have only consisted of one input (excitation of muscle) and one output (force) (Zajac, 1989). By having one input and one output, the models assume that the overall system of coordination can be represented by second-order dynamics and a net activation is produced through a net neural excitation action through first-order activation dynamics (Zajac, 1989). A single input and single output also makes the difference between firing-rate control and recruitment control indistinguishable (Zajac, 1989). An increase in either type of control would result in an increase in the overall neural signal. A model that separates individual motor units would help distinguish which type of control the muscle being studied is under.

2.2 Force-Length Property

In a study conducted by Gordon, Huxley, and Julian (1966), isolated frog muscle fibers were studied for the tension produced by the different degrees of stretch throughout the full range of lengths where the thin and thick filaments of the fiber overlapped. Markers were used to mark the parts of the frog fibers where the striation spacing was uniform and the length between these markers was always recorded by a photoelectric
device. Outputs from the device allowed for the length or the tension to be held constant depending on conditions. The results showed that a plateau of constant tension was reached between 2.05 and 2.2µm and the tension declined sharply on either end of the plateau (Gordon et al., 1966). On either side of the plateau, they found the tensions to reach zero by 1.3µm on the lower side and 3.65µm on the upper side while a maximum tension was found at 2.25µm (Gordon et al., 1966). The authors concluded that the optimum length for the muscle fibers were between striation spacings of 2.05 and 2.2µm.

Zajac (1989) stated that the force-length property of a less than fully activated muscle is a scaled version of the fully activated muscle and a less than fully activated muscle develops proportionately less active force. The less than fully active muscle has not fully activated some or all of the muscle fibers within the muscle by either inactivating some motor units or exciting the muscle units at a lower frequency. Scaling the force length curve of a muscle by activation is consistent with the universally accepted assumption that forces of motor units in parallel sum and forces within an individual fiber sums (Zajac, 1989).

2.3 Effects of Joint Angle and Muscle Length on Force

Joint angle may effect recruitment or rate-coding strategies of the motor units within the muscle or the recruitment strategies between muscles. It is not clear what effect joint position has on motor neuron excitation patterns, surface EMG activity, or the relationship between EMG and force (Doheny et al., 2007). It is suggested that the relationship between muscle length and maximal voluntary EMG activity may be muscle or joint dependent. Many studies have been done to study the change in force with regard to joint angle and muscle length (Doheny et al., 2007; Hansen, 2003; Koo, 2002;
Leedham & Dowling, 1995; Linnamo, 2006; Liu et al., 2013; Solomonow et al., 1986; Vredenbregt & Rau, 1973).

In a study by Vredenbregt and Rau (1973), one subject was studied for the effect of varying angles and force levels between the forearm and upper arm during an isometric bicep curl. A strain gauge dynamometer measured the force produced at the wrist in the same direction as the longitudinal axis of the biceps brachii while an EMG was attached to the biceps brachii. The study resulted in a more than linear increase of the surface EMG of the biceps brachii with increasing force at the wrist. The ratio of activity level and maximum force were calculated and plotted as a function of the mean value of EMG activity. The results showed that one force curve could be drawn through the data on this graph from zero to maximum force for all lengths without increasing their deviations (Vretenbregt & Rau, 1973). The results help explain that the muscle length defines the maximum force at the same level of activity because for maximum force at different lengths the values for activity were equal (Vretenbregt & Rau, 1973).

Solomonow et al. (1986) aimed to quantify the level of activity of the elbow antagonist muscle pair (bicep brachii and triceps) as a function of the developed force by the joint, the effect of gravity and the joint position/muscle length on such relationships. In their experiment, twelve subjects, including six men and six women, aged 19-23 years old, all right-handed, performed isometric contractions at various elbow angles and various percentages of maximal voluntary contractions in both the flexed and extended direction while having electrodes attached to their biceps and triceps. The results found that the agonist muscle reacts in a predictable constant slope as the prime mover of the joint in a given direction, and the antagonist provides the necessary regulation of the
torques about the elbow in order to compensate for the variation of the elbow angle and changes in lever arm length. This shows that joint angle had no effect on the normalized myoelectric signal (MES)-force relationship of the biceps and triceps muscles acting as antagonists (Solomonow et al., 1986).

In a study by Van Zuylen et al. (1988), eight subjects were studied for the activity of motor units of muscles across the elbow joint during voluntary isometric torques at different angles of the elbow. A force-sensor measured the force at the wrist while the forearm was held in a neutral position between full supination and full pronation and the elbow joint was varied between 40 and 180°. The results found that the recruitment level in maximum voluntary contractions vary with joint angle (Van Zuylen et al., 1988). The variation in recruitment threshold is due to muscles with the greater mechanical advantage at a specific angle receiving different neural input to account for the differing mechanical advantage. The change in muscle activation at different joint angles and muscle lengths both influence the maximum force that the muscle can produce at that position.

In a study by Doheny et al. (2007) force generated at the wrist and EMG signals from the biceps, brachioradialis, and triceps were found during a series of isometric contractions at eight elbow joint angles ranging from 10-120°. The experiment consisted of 12 volunteers, including seven female and five male, with a median age of 24.75 years, where all 12 volunteers participated in the six angles from 45-120° and eight of the volunteers participated in the additional angles of 10° and 30°. The results showed that maximum force decreases at more extended or flexed joint angles compared to less extreme angles and the dependency of max force output on the muscle is determined by
the mechanical properties of the muscle. In addition, the level of excitation of each muscle is determined by the required percentage of available force (max voluntary force at that angle) rather than the absolute force required because there was no significant effect of elbow angle on the maximum voluntary contraction EMG activity or the force-EMG relationship (Doheny et al., 2007). The author points out that in order to isolate the effects of neural inputs, reducing mechanical property effects of the muscle is necessary. This could be obtained by normalizing the force-EMG relationship with respect to the max voluntary contraction obtained at the angle (Doheny et al., 2007).

A recent study by Liu et al. (2013) aimed to relate biceps and triceps EMG activity to elbow torque at angles 45°, 60°, 75°, 90°, 105°, 120°, and 135° and examine three non-linear models and EMG processors. The subjects included 12 volunteers, including nine males and three females, aged 18-52 years old. Each subject performed three trials of a sequence of constant-posture, quasi-constant-torque contractions. The study found the EMG processors that included signal whitening and multi-channel combinations had consistent improvement in joint torque estimation and the models that included EMG polynomial degrees of two and accounted for muscle co-activation produced the lowest estimation error (Liu et al., 2013). The results supported past research that suggests the EMG-torque relationship, during the rise in force, at the elbow maintains the same shape across angles and only differs by a multiplicative gain factor as a function of angle (Liu et al., 2013). The authors explained that EMG models aim to mimic the natural relationship between the central nervous system and the peripheral muscles and joints. It is important for the relationship to account for changes in both
muscle length and joint angle because of muscle length-tension relationship and muscle moment arms.

2.4 Using the First Dorsal Interosseous (FDI)

The FDI muscle of the index finger is a muscle that is easily used to find the effects of joint angle on force because of the simplicity of the joint compared to other joints in the body such as the elbow or wrist. The FDI is a small muscle of the body that has a flexion and abduction moment arm at the metacarpophalangeal joint (Infantolino & Challis, 2010). While there is evidence that the FDI plays a role in index finger flexion (as seen in Hudson et al., 2009), it is the only muscle that abducts the index finger.

Previous research has been done using the FDI for these purposes.

Zijdewind and Kernell (1994) conducted a study on how force generated at the index finger was affected during various angles in adduction, abduction, and flexed directions. The study consisted of 16 healthy volunteers, including eight female and eight males aged 20-32 years old. During the experiment, the volunteer’s dominant arm was placed on a table slightly flexed and the hand was vertical with the fingers extended. While the hand, forearm, and middle finger were held immobile, a force transducer measured the index finger forces produced in each direction. The study found that when the muscle was excited by electrical nerve stimulation the abduction force produced by the index finger was dependent on the abduction angle between the index finger and middle finger; the greater the abduction angle between the index and middle fingers, the smaller the abduction force that was produced (Zijdewind & Kernell, 1994). In addition, large maximum voluntary contractions were produced in all directions, however, during abduction the force created during the neutral and maximum angles were significantly
smaller than when the finger was abducted in the minimum angle position (parallel with the middle finger) (Zijdewind & Kernell, 1994). The results show that if the index finger muscles have an optimum length, then increasing degrees of finger abduction from zero degrees would decrease the force production of the FDI.

A study conducted by Hudson et al. (2009), aimed to find the relationship between the amount of force produced by the FDI and how the muscle was driven during a voluntary contraction. The experiment consisted of 15 healthy adults, including seven females and eight males, ages 22-42 years old. The FDI force was measured during the thumbs up (thumb abducted) and thumbs down (thumb extended) positions at various angles. The force was measured using a force-sensor while the moment arm of the FDI was measured using an ultrasonography. The results showed that the FDI’s contribution during finger flexion is altered by the position of the thumb, but it did not change across the various angles once in the position (Hudson et al., 2009). They also found that the neural drive of the FDI was distributed according to the muscle’s ability to contribute to the finger flexion (Hudson et al., 2009). The distribution is because the normalized EMG during voluntary flexion contractions was greater with the thumb down (when the FDI contributes more to finger flexion) than with the thumb up and there were no changes in neural drive to the long flexors. In addition, Hudson et al. (2009) had their subjects perform voluntary flexion contractions of the index finger in both the thumbs up and thumbs down position. They found that the drive to the FDI increased by 28% during the thumbs down position compared to the thumbs up position (Hudson et al., 2009). This proved that the FDI is at a greater mechanical advantage during finger flexion when the thumb is abducted because the flexion moment arm of the FDI is increased. As the
mechanical advantage of the muscle is improved, the neural stimulation and muscle contribution increases (Hudson et al., 2009).

2.5 Activation Dynamics

Activation dynamics is the transformation of neural excitation to the activation of contractile apparatuses and then the transformation from muscle contraction dynamics to the activation of muscle force (Zajac, 1989). The sequence of motor unit discharge contains the information of the neural excitation to the motor unit (Zajac, 1989). The neural excitation is coupled with the contractile machinery of the muscle fiber through muscle activation, also known as \( q \). It is still unclear as to whether activation is coupled with the muscle contraction dynamics, or not.

Adrian and Bronk (1929) explained that the degree of voluntary activation, or the neural drive to the muscle, has an influence on the muscle’s force generation. Although the mechanism of how the muscle regulates the excitation of the motor units has been a subject of speculation for many years, different motor units will respond differently to excitation. The relationship between force and frequency is not fixed; force and frequency are different recruitment strategies. An early study by Henneman in 1957 found that the response to increasing excitation of motor units was the recruitment of motor units by size. This phenomenon became known as the recruitment theory and is now accepted by the scientific community.

A study by Eccles et al. in 1958 found by measuring the spike potentials, after-potentials and conduction velocities of different motor axons in slow and fast twitch muscles, that electrically stimulated, higher threshold, larger diameter motor neurons produced a shorter afterhyperpolarization and a greater firing rate than the lower
threshold, smaller diameter motor neurons. They believed that the afterhyperpolarization stage was a large contributor to determining the firing rate of each motor neuron. Their study suggests that earlier recruited motor units would fire with a lower firing rate than the later recruited motor units. Studies conducted since have found that the opposite is true. These studies suggest those higher threshold motor units, which are activated later at higher levels of excitation, have a lower firing rate than the lower threshold motor units recruited first (DeLuca & Contessa, 2012).

DeLuca and Contessa conducted a study in 2012 that aimed to formulate a model that provides a general description of the firing behavior of a set of motor neurons regulating voluntary isometric contractions where the force varies linearly with time. They studied the *vastus lateralis* and the first dorsal interosseous because the two muscles have different firing rate properties. The study consisted of eight healthy subjects with no known neurological disorders and included five men and three women, aged 19-35 years old. The subjects performed leg extension and index finger abduction while isometric force was measured via load cells attached to the lever arms of the restraints. Three main concepts emerged from the study’s results. First, once a motor neuron is activated, the firing rate increases as a negative exponential function as the excitation increases (DeLuca & Contessa, 2012). Second, they found that parameters that define the exponential function (such as time constant, firing rate at recruitment, and peak firing rate) can all be expressed as linear functions of the recruitment threshold, with the time constant being directly proportional and the firing rate at recruitment and peak firing rate being inversely proportional (DeLuca & Contessa, 2012). Last, the trajectory of the firing rate is only weakly influenced by the force rate of the contraction within the range
of 2% and 10% maximal voluntary contractions per second (DeLuca & Contessa, 2012). The study also found that firing rates at recruitment decrease with increasing recruitment thresholds and the velocity of the firing rates gradually decrease until the firing rate trajectory reaches a peak value that is inversely proportional to the recruitment threshold. This means that the rate of increase is determined by the time constant and the larger--diameter, later recruited motor units have a slower firing rate increase and their firing rate value will be lower at any time and force than the earlier recruited motor units (DeLuca & Contessa, 2012). From the results of their study, DeLuca and Contessa (2012) found that the control strategy of muscles does not optimize the force output of the muscle. The earlier recruited motor units have a short duration and they do not tetanize. The later recruited motor units fire slower and tetanize, but their firing rate trajectory decreases quickly in order to economize energy when the force output reaches its maximum.

DeLuca and Contessa (2012) explain that the use of a dEMG allows for the analysis of firing rate behavior of each motor unit when the motor neurons are excited to a certain level. They found that as excitation increases the firing rates of all of the activated motor units increase. They have coined this the ‘common drive’. The common drive represents the net excitation that is received by all of the motor neurons in the same pool. On the dEMG, the hierarchical increase in firing rate and the inverse relationship to the recruitment threshold present an image that resembles an onionskin (DeLuca & Contessa, 2012). The authors explain that the onionskin property control scheme has optimized the combination of force magnitude and time duration and provides a greater economy of force generation during functional daily activities. Using the decomposition
electromyography (dEMG), we can come closer to answering whether motor units recruit the same every time or if recruitment changes with the length of the muscle.
Chapter 3

Methods

3.1 Participants

Five males and one female (ages 20-29) volunteered to participate in this study. The Pennsylvania State University Institutional Review Board for Human Subjects approved the study and all participants signed a written informed consent form before testing. All participants were recruited from the Pennsylvania State University Kinesiology student population of the Berks Campus.

3.2 dEMG Signal Detection and Processing

Prior to sensor attachment, the participant’s skin was debrided of dead skin and cleansed with an alcohol wipe in order to remove any dead skin or dirt that could cause interference. A grounding pad was placed on the participant’s right olecranon and the five-pronged dEMG electrode was placed on the web space between the thumb and index finger of the right posterior hand. The participant was instructed to place his/her right hand onto a wooden board that would position his/her index finger at the correct angle against the force sensor. The participant’s other fingers and thumb are positioned to rest comfortably away from the force sensor and restricted in order to keep them from influencing movement.
Once a signal of ±8 µV or less during an array signal check was established, the electrode was taped into place using hypoallergenic tape. The participant was asked to stay relaxed unless instructed to perform a task.

The EMG data acquisition software used during the study was EMGWorks (Delsys Incorporated, Boston, Massachusetts, USA). Starting with 0° of abduction at the second metacarpophalangeal (MCP) joint, the participant was guided through a maximal voluntary contraction (MVC) task that asked the participant to contract at his/her maximum force against the force sensor for three seconds. The software located the participant’s maximum force for the angle before moving on. The participant was then guided through a decomposition check that ensured that the signal could be decomposed. During this task, the participant was asked to maintain a force around 50% of the maximum for 15 seconds. The signal had to fall within the acceptable limits of baseline noise, 50/60 Hz noise (which represents electrical interference) and acceptable signal to
noise ratio (SNR) in order to be decomposed. If the signal did not pass the
decomposition check, the signal was rechecked to ensure that it was still within the ±8
µV range and the decomposition check was redone until the signal passed. If necessary,
the dEMG electrode was repositioned and the entire process was started from the
beginning.

Once the signal passed the decomposition check, the participant was guided
through a tracking paradigm at 0° of abduction. The tracking paradigm brought the
participant through force exertion that ranged from 0 to 50% of the participant’s
maximum for that angle for 40 seconds. The tracking paradigm increased force 12.5% of
MVC per second for four seconds, held 50% max MVC for 32 seconds then decreased
12.5% of MVC per second for four seconds. Following the task, the signal was analyzed
for its quality again. If the signal fell within the limits for each category, the force sensor
was moved to the next angle. If not, the signal quality and the signal array were checked
and the tasks were redone. The participant was asked to rest between the tasks for each
angle to ensure that the muscle did not fatigue during the tasks. For each of the following
angles (5°, 10°, 15° and 20°), the participant was guided through the MVC and tracking
paradigm tasks. After the completion of all tasks for each angle, the data was
decomposed and analyzed by EMGWorks Analysis (Delsys Incorporated, Boston,
Massachusetts, USA).

3.3 Analysis

The data was transferred to Matlab and analyzed using custom written code in
order to best fit the data to the equation from Deluca and Contessa (2012):
\[ \hat{\lambda}_i(t, \tau_i) = \hat{\lambda}_r_i + (\hat{\lambda}p_i - \hat{\lambda}r_i)[1 - e^{(\tau - t)/\theta_i}] \]  

Where,
- \( \hat{\lambda}_i \) is the mean firing rate as a function of time and recruitment threshold
- \( \tau_i \) is the recruitment threshold as a function of normalized MVC
- \( \theta \) is the time constant of the fitted curve
- \( \pi_i \) is the estimated value of the firing rate at recruitment from the fitted curve
- \( \hat{\lambda}p_i \) is the estimated value of the peak firing rate obtained from the fitted curve

The slope coefficients (\( \theta \)) from the fitted curve of each motor unit for one angle were plotted against its corresponding motor unit (MU) number. This was repeated for the following four angles. The slopes from the \( \theta \) versus MU number graphs for each angle were then plotted against their corresponding angle in a separate graph and fitted with a best-fit line in order to analyze whether \( \theta \) changed based upon length. The \( \theta \) values for all subjects were averaged for each angle and plotted against the corresponding angle and fitted with a best-fit line in order to find an average \( R^2 \) value for all subjects across all angles. A regression test was then run on the average slope values of all subjects by each angle to find significance.
Chapter 4

Results

There were reasonable fits to the motor unit firing rate data for each motor unit using equation 1. Table 4.1 displays the average per angle percent root mean square error (%RMSE) of the fits of equation 1 to the actual firing rates for each subject. The %RMSE values ranged from 6.7% to 12.8% demonstrating a good fit to the equation.

Table 4.1: Average %RMSE of the fits of equation 1 to actual firing rates

<table>
<thead>
<tr>
<th>Subject</th>
<th>0°</th>
<th>5°</th>
<th>10°</th>
<th>15°</th>
<th>20°</th>
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<td>9.6</td>
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</tbody>
</table>

Figure 4.1 demonstrates an example of the fits for the first, second, and last motor unit activated during the tracking paradigm for one subject at one angle. Figure 4.2 demonstrates an example of the best fit of slope coefficients from the $\theta$ value of each motor unit from figure 4.1 plotted against its corresponding motor unit (MU) number. Figure 4.3 demonstrates an example of the best fit of the slopes from figure 4.2 for each angle in which the rate of change over the $\theta$ values across the range of motion was analyzed for each subject.
Figure 4.1: Example of fit of equation 1 to individual motor units

Figure 4.2: Example of the best fit of slope coefficients from the $\theta$ value of each motor unit as a function of MU number
Figure 4.3: Example of the best fit of the slopes from the theta values as a function of MU numbers for each angle

Table 4.2 displays the slopes and coefficient of determination (R^2) values for each subject across all five angles. The R^2 values for each subject individually are not high. All of the subjects had positive slopes except for one.

<table>
<thead>
<tr>
<th>Subject</th>
<th>R^2 Value</th>
<th>Slope</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.34829</td>
<td>0.0009</td>
</tr>
<tr>
<td>2</td>
<td>0.60574</td>
<td>0.0016</td>
</tr>
<tr>
<td>3</td>
<td>0.36311</td>
<td>0.0008</td>
</tr>
<tr>
<td>4</td>
<td>0.02731</td>
<td>0.0002</td>
</tr>
<tr>
<td>5</td>
<td>0.37407</td>
<td>-0.0005</td>
</tr>
<tr>
<td>6</td>
<td>0.73615</td>
<td>0.0012</td>
</tr>
</tbody>
</table>

Figure 4.4 displays the best fit of the average slope values for all subjects at each angle and Table 4.3 displays the average slope values for each angle. The slope of the best-fit line had a positive slope of 0.0007 and a R^2 value of 0.87362. The R^2 value of the average θs was higher than the individual R^2 values for any subject across all of the
angles. The regression test of the average slope values of all subjects by each angle was statistically significant from zero (p= 0.0198).

Figure 4.4: Pearson Correlation by average slope values for all subjects by angle

Table 4.3: The average slope values for all subjects by angle

<table>
<thead>
<tr>
<th>Angle (°)</th>
<th>Slope</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.0409</td>
</tr>
<tr>
<td>5</td>
<td>0.039733</td>
</tr>
<tr>
<td>10</td>
<td>0.046783</td>
</tr>
<tr>
<td>15</td>
<td>0.047567</td>
</tr>
<tr>
<td>20</td>
<td>0.05445</td>
</tr>
</tbody>
</table>
Chapter 5

Conclusion

From the results of the study, it appears that there is a statistical effect of length on recruitment strategy. The results fall into disagreement with the assumption that for maximal activation at different angles, the activation dynamics of the muscle will be unchanged since the muscle is still working at its maximum for that angle (Brinckman et al., 2002). The results rejected the hypothesis that there would be no significant difference in activation dynamics across a range of motion of the first dorsal interosseous (FDI) muscle. The statistical difference ($p < 0.05$) of the slope of the $\theta$ versus angle line from zero indicated a correlation between $\theta$ and angle. A slope of zero would imply that length does not have an effect on recruitment strategy. With a slope greater than zero, the data shows an effect but whether that effect is meaningful is not well understood. Within a small study the effect of length on recruitment strategy may be more influenced by outliers than a larger study would be. The effect could become non-significant when induced in a larger and more diverse population of people.

Some limitations existed with the study. The sample sized used in the study was small. Due to the small sample size, the results do not extrapolate well to a larger population. More subjects would be needed in order to apply the results to a larger population. More male subjects were used compared to female subjects because a clean signal could not be obtained from several females. The females were later rejected from the subject population after multiple attempts at finding an acceptable signal were tried
and the source of error could not be found. Therefore, it would be difficult to generalize to a female population. The age of the subjects would also cause difficulties when applying the results to a larger population. The subjects were all recruited from a college campus and were all between the ages of 20 and 29 years old. The results of the study would not accurately predict the outcomes of younger and older communities. The FDI is also not a clinically significant muscle in terms of its importance in daily bodily functioning. Using a muscle that is worked through a larger range of motion and having the mechanical advantage of the muscle changing throughout that range of motion may show more significant changes in the data itself because recruitment strategy is changed in order to optimize muscle energy and force (Van Zuylen et al., 1988). Another large limitation to the study was that muscle fatigue might have skewed the data. When one muscle exerts a constant force for a long period of time without rest, the muscle fatigues but the overall activity of the muscle increases (Vredenbregt & Rau, 1973). When the muscle fatigues, the EMG activity increases due to additional motor units being recruited by the muscle to maintain the constant force (Edwards & Lippold, 1956).

Future studies could be completed to improve on the results of the current study. Incorporating participants that complete the study in a randomized order would help reduce the effects of muscle fatigue on the results. The participants would no longer be asked to work from 0° abduction to 20° abduction, increasing not only in angle but also in muscle fatigue. Randomizing the angle order would allow the larger angle (10°, 15° and 20°) tasks to be conducted using a muscle that was not already fatigued. The results would include tasks from each angle that were completed using both a previously
fatigued muscle and an unused muscle. The results would then consider the effect that muscle fatigue has on EMG activity and provide more accurate data.

In addition, a larger muscle should be considered in order to increase the amount of length/angle change. Repeating the methods of the current study on a larger muscle would investigate if the correlation between length and recruitment strategy were found when worked through a larger range of motion. Studying a larger muscle would also investigate if the correlation were present in a muscle with a more significant contribution to daily bodily functioning. The FDI does not contribute to many tasks that are performed consistently throughout daily life. As seen in Van Zuylen et al. (1988) and Hudson et al. (2009), muscles that contribute more to a force and have a greater mechanical advantage receive different neural inputs when it comes to recruitment. The effect of length on the recruitment strategy may change as the type of recruitment changes in order to optimize muscle energy.

In spite of the limitations of the study discussed above, the study was able to provide evidence that a change in the length of a muscle does affect the recruitment strategy of the muscle during maximum voluntary contractions. However, further research would need to be conducted in order to extend the results to a larger population and increase the validity of muscle modeling.
REFERENCES


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