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Assessment of force coordination and neuromuscular quickness in healthy adults

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ASSESSMENT OF FORCE COORDINATION AND NEUROMUSCULAR QUICKNESS IN HEALTHY ADULTS

by
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Dedications

To my parents, Debbie and Mike, and my brother, Eric.

“There’s nothing that makes you more insane than family. Or more happy. Or more exasperated. Or more…secure.” –Jim Butcher

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Thank you, Kodi, for always being up for a walk to unwind at the end of each day. Lastly, my regards to Burt for being an unwavering presence through it all.

Abstract

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Throughout daily life, it is necessary to handle and control innumerable objects. To do so, one’s hands must be precisely regulated. To ensure that an object is effectively manipulated, an individual must apply a grip force (GF) perpendicular to the object’s surface to overcome load force (LF), which acts tangential to the surface to counteract the object’s weight and inertia. Previous studies have shown an elaborate coordination between GF and LF in a variety of object manipulation tasks in healthy populations. This kinetic analysis is clinically important because the GF-LF coordination is shown to deteriorate in aging and neurologically impaired populations. Within this thesis, we explored the coordination between GF and LF and their neuromuscular quickness values in rapid force production tasks that could represent conditions where one has to grasp externally fixed objects to avoid falling. We varied the parameters of surface friction (e.g. high and low friction) and LF direction (e.g. pulling up and pushing down) in order to evaluate variables that could potentially affect the measured outcomes. Overall, this study created a simple, non-invasive measurement technique that quantifies force coordination and neuromuscular quickness in healthy, young adults.

Table of Contents

Table of Contents (Continued)

List of Figures

Figure Page

List of Tables

Chapter 1

Introduction

Hand use is an indispensable human function; one that is used constantly and for an infinite variety of actions. It can be seen in tasks as varied as writing a letter, steering a car, or catching a falling bowl. Unfortunately, both hand function and neuromuscular quickness could be adversely affected by the aging process or the presence of a neurological disease such as multiple sclerosis, stroke, or Parkinson's disease. As a consequence, it becomes difficult to manipulate everyday objects or to generate forces quickly, both of which decrease the patient's quality of life and makes him or her susceptible to falls and other related injuries. If a standard, non-invasive protocol can be developed that will allow doctors in any medical facility to test for muscle and neurological impairment quickly and easily, then these issues may be diagnosed earlier and, therefore, afford the patients a better chance of recovery. Used over time, it will also evaluate the efficacy of the medical treatment or exercise regimen, allowing doctors to suggest alternative options if necessary. Furthermore, this protocol could give insight into how the central nervous system controls hand function and quickness in healthy populations, which will later highlight alterations in these movements in neurologically impaired populations. This introduction will serve as an overview of two important aspects of motor control, hand function and neuromuscular quickness, which will be combined to form a powerful new clinical tool that assesses those aspects within a single measurement technique.

1.1 Hand Function

Throughout a normal day, humans interact with free or externally fixed objects continuously. The ability to handle these objects appropriately is highly dependent upon the capacity of the nervous system to perform the desired movements with precision. Many of the factors that affect this dexterity are discussed below.

Dysfunctions of the central nervous system (CNS) may disrupt the coordination of the forces used in object manipulation. Any impairment of the hand's motor control, whether by aging or disease, could cause excessive or insufficient grasping, or the motion itself could simply be slowed down. To test for the presence and severity of these dysfunctions, many evaluations have been developed. Some take the form of questionnaires, such as the Michigan Hand Outcomes Questionnaire, the Patient Evaluation Measure, and the Patient Rated Wrist Hand Evaluation, while others are performance-based tests such as the Functional Dexterity Test, the Jebson Taylor Function Test, the Nine Hole Peg Test, and the Grooved Pegboard Test [1]. However, though these tests are convenient for a clinical environment, they fail to determine precisely how the CNS is controlling the hand. The test proposed in this study will be a quantitative evaluation that will precisely demonstrate any deviation in coordination from the healthy norm while also giving insight into the neurophysiological mechanisms governing hand function.

The kinetic assessment of force coordination during the manipulation of objects has lately shown to be a robust clinical evaluation for the analysis of hand function [2]–[4]. For this simplified kinetic model, it is necessary to exert a force normal to an object's contact surface in order to deliver a force tangential to the surface, which will either yield a reaction force from the object's external support if the object is fixed or work against the object's weight if it is free to move. The force exerted normal to the surface is designated the grip force (GF), while the tangential forces, including the weight and inertia of the object, is called the load force (LF). The physical attributes of the object as well as the demands of the task to be performed also determine the type of manipulation used. It is common to use a precision grip, one that may involve just the tips of the fingers, to hold fragile objects like a feather or test tube. Uni- or bi-manual power grasps, which may use the full hand including the palm and fingers, help swing a baseball bat or lift weights at the gym. In this study, the precision grip will be utilized because it lends itself to simple measurements of GF and LF. The magnitude of each force, as well as the degree of coordination between GF and LF, is then analyzed to reveal patterns in the motor control of hand function.

Within this simplified model, it can be noted that the amount of grip force required to manipulate the object is dependent upon the friction between the skin and the surface of the object. When utilizing a device in a vertically oriented fashion, the minimum grip force (GF_{min}) is noted to be the GF at the point just prior to object slip. Therefore, this value may be calculated as:

$$GF_{min} = LF/2COF$$

which is the ratio of the LF to the coefficient of friction (COF) for the fingers on either side of the object [4], [5]. However, humans almost always apply a GF higher than the necessary GF_{min} during manipulation tasks [5]. This ensures that any voluntary or involuntary change in LF will not immediately cause slippage. The difference between the engaged GF and the minimum ($GF-GF_{min}$) is called the safety margin (SM) [5]. The

SM is variable across subjects, but in general it is seen as the adaptation of GF to LF changes that is both low enough to prevent muscle fatigue and crushing of the object and high enough to avoid dropping the object [5], [6]. It would appear that the central nervous system (CNS) is able to quickly alter the GF to fit these requirements, and the changes are remembered for future manipulations of the same objects [7]. These preprogrammed reactions allow the hand to grip an object with high precision more quickly [6], [8].

De Freitas et al. (2009) noted that either the relative SM, taken as a percentage of GF_{min}, or the absolute SM, given as the difference in GF and GF_{min} in Newtons (N), could be observed. In a study of both dynamic and free holding tasks, it was found that the CNS was able to hold the absolute SM, but not the relative SM, mostly constant throughout trials in which friction, LF range, LF frequency, and given instructions were varied [6], [8]. This suggests that the absolute SM should be the metric used in future coordination studies.

When modeling a hand-object system, it is also necessary to account for forces in the surface-to-skin interface. When two objects come into contact, the force required to move one over the other must overcome the resistance caused by friction. According to Amonton's laws of friction, this frictional force is proportional to the load but independent of the contact area. This allows the ratio of the frictional force to the load to be a constant, which is called the coefficient of friction [9]. The coefficient of static friction is considered when the objects are not moving relative to each other, while the coefficient of dynamic friction is considered once movement has been initiated and is a much smaller value than its counterpart. Using these laws in a simple mechanical model, the coefficient of friction, COF, may be calculated as

$$\text{COF} = \frac{1}{(2 * \text{slip ratio})}$$

where the slip ratio is the ratio between the grip force, GF, at the moment just prior to the beginning of object slide and the load force, LF. The slip ratio is doubled because it is assumed that the LF is distributed equally on the surfaces on both sides of an object [5], [10].

Skin does not fully follow Amonton's laws because it exhibits viscoelastic properties. Skin displays two main mechanisms of viscoelasticity: adhesion and hysteresis. These mechanisms increase the force required to move an object against the skin [11]. The COF of skin may also be dependent on numerous factors including, but not limited to, age [11], [12], normal force [9], [12], sliding speed [11], area of contact surface [9], [13], roughness of the skin [10], [11], roughness of the object [5], [14], hydration [9], [11], sweat [15], and the anatomical sites being used [9], [16], [17]. It is clear that the COF of the skin may vary from any number of factors at different times. As such, without further investigation into the intricacies of skin friction, this study, like many before it, makes the simplifying assumption that skin follows Amonton's laws. It is common to assess the static COF using standard slip method test developed by Johansson and Westling in 1984. Therefore, the COF is measured as the ratio between the normal force and the static frictional force [5], [10], [17]. During object manipulation, the GF modulates to maintain an adequate safety margin proportional to this value [5], [14]. The modulation of the grasping forces happens rapidly, often between initial grasping and lift onset [18].

Traditionally, GF-LF coordination has been assessed through three indices: GF scaling, GF-Lf coupling, and GF modulation. The scaling of GF, often discussed as the

GF/LF ratio and calculated with either the average or peak values, incorporates information from skin afferents to vary with the object's surface properties, such as roughness [7]. A relatively low and stable GF/LF ratio – even with changing LF – is an indication of precise force coordination [5], [7], [14]. The coupling of GF to LF involves the maximum cross-correlation coefficient (r) and the time lag between the waveforms of the two forces. Commonly, studies have found that larger r values and shorter lag times represent instances of high force coupling [8], [19]–[21]. Additionally, GF modulation gives an illustrative description of how the GF adapts to changing LF. A diagram of GF vs. LF shows GF gain as the slope of the regression line, while GF offset is the intercept [21]. Low GF offset and high GF gain are taken as indices of proficient coordination. These indices describe which force may be impaired, at what time during the manipulation, and how significantly these changes affected the motion.

Such precise coupling of GF and LF is only possible through the use of the CNS as a control system. It is suggested that the force coordination is controlled primarily by feedforward or predictive control mechanisms, wherein pre-manipulation visual clues lead to a selection of one set of pre-programmed motions [7]. In fact, delayed visual feedback has been shown to increase the time lag between GF and LF significantly [22]. But when visual clues are present, they allow response trajectories to an assumed amplitude to be predicted [23], [24]. This mechanism leads to a coordinated reaction of GF and LF. Following the initial movement, a reciprocal activation of the antagonist muscles occur at peak dF/dt in order to counter the agonist muscle contraction, and the opposing forces prevent over- or under-gripping [25]. If the CNS does not anticipate LF precisely, then it will adjust GF based on digital feedback from tactile afferents and non-digital mechanoreceptors [7], [26], [27]. This adjustment begins after a short delay, which is measured as the time lag. This feedback is necessary on at least an intermittent level because deafferented subjects will otherwise inefficiently over grip and display large variations in lag times [28]. Through practice and further experience with various loads and surface frictions, this lag may be reduced to yield a more efficient movement [24].

Beyond discrete manipulations such as picking up or setting down an object, daily tasks include continuous motions, such as gripping the handrail on a rocking boat or shaking a container of orange juice. During these tasks, it is necessary to either repeat one motion at a given frequency or reverse forces to utilize force control in two dimensions. When performing a series of bidirectional oscillatory manipulation pulses, if any force is required in opposition to the main LF direction, then the force control yields lower GF-LF coordination [19]. Similarly, in unidirectional oscillatory tasks a change in LF frequency leads to significant decreases in GF-LF coupling and GF modulation, while varying the range of LF change has a relatively weak effect [20], [21]. However, regardless of whether the motion was horizontal or vertical or varying surface texture, grip force always modulates in phase with the load [8], [29]. Therefore, dynamic testing is a valid method of evaluating coordination, and the frequency, but not the range, of LF should be considered when designing the protocol.

Issues arising with increasing age as well as neurological disease also influence the elaborate coordination between GF and LF. In general, aging populations demonstrate increased GF and larger relative safety margins [30]. For middle-aged subjects, increased skin slipperiness (decreased COF) necessitates a firmer grip, while subjects over the age of sixty experience declining cutaneous afferent function and compensate with exaggerated frictional scaling [11], [30]. Furthermore, in force modulation tasks, both young and older adults improved force tracking accuracy with practice, but older adults required more practice for lower precision [31].

Similar changes were noted in individuals with neurological impairments. Hermsdörfer et al. observed that when participants who had suffered a stroke were asked to manipulate an object, they demonstrated larger applied GF and decreased GF-LF coupling, although with almost no GF time lag during varying LF, proportional to the severity of their disease [2]. Furthermore, simple lifting tasks have shown to be valid tests for the diagnosis of multiple sclerosis, as subjects with the disease demonstrate larger peak GF, higher GF/LF ratio, and significantly larger time lags between GF and LF than healthy subjects [3], [32]. Similar impairments were noted in children with hemiplegic cerebral palsy [33] and in subjects with Parkinson’s disease [34]. With this great number of clear examples of motor control impairment’s effect on the given variables, it is imperative that this method becomes a standard for neurological examinations. This method could not only serve as a diagnostic tool by indicating coordination deficiencies, but it could also be used throughout the treatment process to determine if a patient’s impairments have been reduced.

1.2 Neuromuscular Quickness

Unfortunately, falls are inevitable in daily life, and, while they are often inconsequential, for many people even a small spill could be devastating. This is especially evident for the elderly, as falls have become the leading cause of unintentional injury deaths in Americans over the age of sixty-five [35]. Falls also create an enormous financial burden. For instance, in 2000, healthcare networks treated 2.6 million non-fatal fall related injuries, and direct medical costs summed to about $200 million for fatal injuries and $19 billion for non-fatal [36]. Often, what matters most in preventing injury is how quickly the individual can step or reach out to support him- or herself. Quickness is a critical quality of movement that is represented as a submaximal muscular effort used to complete a movement in the shortest necessary time [37], and it may be illustrated in such scenarios as jerking a steering wheel to avoid an accident or thrusting out a hand to grab an externally fixed object after tripping. As rapid movements are a significant part of daily life, it is imperative that a method be developed that will quantify an individual's ability to exert rapid submaximal forces, which will aid in the assessment of treatment options for those who exhibit slowness in force production such as aging and neurologically impaired populations.

Quickness has been assessed in various levels of motor control including kinematic, muscular, and kinetic levels. Freund and Büdingen also utilized kinematics to investigate the "relationship between speed of the fastest possible voluntary contractions and their amplitude for hand and forearm muscles" by performing isometric and isotonic movements as quickly as possible, which showed that the rise time of goal-directed voluntary contractions was approximately constant regardless of size [38]. In order to more deeply evaluate CNS control for these movements, motor unit recruitment and firing rate is investigated by fine-wire electromyography (EMG). Studies using EMG noted the presence of a highly ordered motor unit recruitment scheme based upon motor neuron excitability and found that the initial agonist burst magnitudes are strongly related to the peak force achieved [39], [40]. Wierzbicka et al. (1986) also utilized fast, goal-directed voluntary movements in order to study the triphasic pattern of agonist and antagonist muscle bursts and showed that the time to peak displacement is a function of antagonist input [41]. Although all of these methods are readily applicable to studies of quickness, EMG and kinematic testing often require invasive methods and expensive equipment that hinders the practicality of these methods in clinical settings. Also, while functional tests do measure the timing of movements, they fail to give insight into how the CNS controls the rapid muscle coordination.

In lieu of the above testing methods, kinetic analysis is traditionally utilized for quickness assessments because it both quantifies the force production and clarifies the methods of control utilized by the CNS. In this method, subjects are evaluated through submaximal force production tasks under the instructions to produce them as quickly as possible at varying amplitudes. However, subjects are instructed not to aim for precision in amplitude selection, because it has been shown that the CNS will alter the rise time of the force in order to accurately select a chosen amplitude [23]. Using these pulses, the peak force (PF) and corresponding rate of force development (RFD) is found [38]. Previous findings by Freund [38] and Wiezerbicka [41], [42] have shown that PF and RFD have a strong linear relationship. The slope of this line– the amount the RFD must adjust to match the contraction amplitude– is called the RFD scaling factor (RFD-SF) and may be used as an index of quickness [37], [43]. A kinetic study by Bellumori et al. utilized rapid force pulses to confirm the above findings and showed that the scaling of force development was generally comparable among muscle groups [37]. This study also showed that the time required to reach peak force is relatively invariable regardless of contraction size.

Depending on the quickness desired and the amount of force needed for these movements, the rate of rise of tension in the muscles also vary. Freund and Büdingen showed that the speed with which the force increases depends upon the amplitude of the target force [38]. Correspondingly, the increase or decrease in the rate of force production with amplitude ensures that the rise time is approximately constant [23], [38]. Because it has been found that elderly adults display a decline in the maximal motor unit discharge frequency, and therefore exhibit slower RFD and longer rise times, these variables have proven to be valid measurements of age-related quickness [43], [44].

When compared to healthy, young adults, age-related and neurological issues cause the previously discussed variables to fluctuate significantly. A study by Graafmans et al. showed that elderly adults who reported falls were more likely to experience recurrent falls, with the strongest risk factor being mobility impairment [45]. Clearly, quickness degeneration has affected mobility, thereby making these adults unable to control their falls. In fact, Kim and Ashton-Miller showed that older adults react less efficiently than younger, leading to shorter available fall response times [46]. Further studies clearly show limited scaling of the RFD and less consistency in maximum force production in older adults [43]. In a different manner, Parkinson’s disease (PD) patients are able to maintain fast muscle contraction, but the motor output is improperly scaled, leading to larger than normal agonist bursts [42]. Furthermore, Park and Stelmach showed that when subjects are asked to perform rapid, isometric forces as quickly as possible, those with PD have reduced rates of force development and take a longer time to reach peak force [47]. The proposed simple, noninvasive test of these functions would be a straightforward method of diagnosing these dysfunctions or verifying given treatment methods in a single measurement setting.

1.3 Combined Testing

Both hand function and neuromuscular quickness are critical qualities of human movement. Coordination testing illustrates the ability of the two forces, GF and LF, to work together to complete an object manipulation efficiently, while quickness measurements demonstrate how rapidly those forces act to complete a movement in the shortest amount of time. However, coordination studies do not analyze rapid submaximal force production, which is a major factor in daily life and is critically lacking in neurologically impaired populations. Furthermore, the studies of quickness focus only on the rate of LF production and ignore such variables as the rate of GF production and GF-LF coordination. This thesis proposes a merger between these two methods of analyzing motor function. The goal is to evaluate GF and LF when subjects perform rapid manipulation tasks on an externally fixed force measuring device, which will not only allow for examination of quickness-related variables like RFD-SF in both forces but will permit investigation of the indices of force coordination during these movements.

This new, ecologically valid assessment of both coordination and quickness will have numerous applications. The results will provide clarity to the ways in which the CNS controls movement while also allowing for analysis of the effects of aging and neurological diseases on motor control. Furthermore, it could become a diagnostic tool for neurological diseases, and it could assess the outcomes of a given treatment in order to determine the next step to take. Similarly, this method may also be utilized as a progress check for injury rehabilitation purposes. As both a research and diagnostic tool, this method will serve as a basis for even further exploration of the neuromuscular system.

1.4 Aims and Hypotheses

This study aims to develop a novel method of assessing hand function and neuromuscular quickness and to standardize this technique by testing multiple pulse directions and surface frictions to vary the conditions of force production.

Specific Aim 1: To explore the coordination between GF and LF during rapid isometric force pulses to varying submaximal levels.

Hypothesis 1: Coordination between GF and LF during rapid isometric pulses will be high in healthy, young individuals.

Specific Aim 2: To assess the quickness of GF and LF in rapid, isometric object manipulation tasks performed to varying submaximal force levels.

Hypothesis 2: The variables of quickness will be similar in GF and LF, and these values will be high in healthy, young adults.

Specific Aim 3: To assess the effects of force direction (i.e. pulses up and pulses down) on GF and LF coordination and quickness.

Hypothesis 3.1: The values of GF-LF variables will be the same for both pulse directions.

Hypothesis 3.2: The values of quickness variables will be the same for both pulse directions.

Specific Aim 4*:* To assess the effects of surface friction on GF and LF coordination and quickness.

Hypothesis 4.1: GF and LF will be more highly coordinated on the rougher surface.

Hypothesis 4.2: GF will be produced faster on the smoother surface.

Chapter 2

Manuscript

2.1 Abstract

Throughout daily life, it is necessary to handle and control innumerable objects, requiring precise regulation of hand function. To ensure that an object is effectively manipulated, an individual must apply a grip force (GF) perpendicular to the object's surface to overcome load force (LF), which acts tangential to the surface to counteract the object's weight and inertia. Previous studies have shown an elaborate coordination between GF and LF in a variety of object manipulation tasks in healthy populations. This kinetic analysis is clinically significant because the GF-LF coordination is shown to deteriorate in aging and neurologically impaired populations. Here we explored the coordination between GF and LF and their neuromuscular quickness values in rapid force production tasks, which represent conditions where one has to grasp externally fixed objects to avoid falling. The aims of this study were to develop a clinically meaningful measurement technique to assess GF-LF coordination and neuromuscular quickness simultaneously, to standardize this method, and to assess its reliability. The GF-LF coordination values (GF/LF ratio, cross-correlation (r_{max}), and time lag) matched with those reported in previous studies, while the neuromuscular quickness variables (rate of force development-scaling factor (RFD-SF), R^2, intercept, and half-relaxation) revealed that GF is consistently slower and takes a longer time to relax than LF. To further standardize the propose measurement technique, we varied LF direction (e.g. pulling up and pushing down) and found a higher cross-correlation between GF and LF in the more ecologically valid downward direction than those in the upward direction. Furthermore, a

concurrent validity assessment revealed that a selection of twenty pulses were reliable for analysis. Results support this protocol as a simple, non-invasive measurement technique to quantify force coordination and neuromuscular quickness in a clinical setting.

2.2 Introduction

Both hand function and neuromuscular quickness are important qualities of motor function that could be adversely affected by the aging process or the presence of a neurological disease such as multiple sclerosis (MS), stroke, or Parkinson's disease. As a consequence, it becomes difficult to manipulate everyday objects or to generate forces quickly, both of which decrease the patient's quality of life and makes him or her susceptible to falls and other related injuries. Although some qualitative and quantitative assessment tools have been used to make clinical decisions, they are limited in their ability to detect small changes in hand function and to explain how the central nervous system (CNS) controls this function.

Recently, a simple kinetic analysis of holding a vertically oriented object has shown to be a robust clinical evaluation of hand function [2]–[4]. Within this model, an individual must ensure that an object – whether it be fixed or free to move – is effectively manipulated by applying a grip force (GF) perpendicular to the object's surface to overcome the load force (LF), which acts tangential to the surface to counteract the object's weight and inertia. The force coordination during object manipulation has been studied through two indices; GF scaling and GF-LF coupling. GF scaling, calculated with either the average or peak values, is seen in the relatively low and stable GF/LF ratio. Observed even when changing LF, this is an indication of precise force coordination [5], [7], [14]. The other index of force coordination, GF-LF coupling, involves the maximum cross-correlation coefficient (r_{max}) and the corresponding time lag calculated from the waveforms of GF and LF. Commonly, studies have found that larger r_{max} values and lag times close to zero represent instances of high force coupling [8], [19]–[21].

This elaborate force coordination is readily seen in young, healthy individuals, even in situations with varying object surface friction [5] and LF range and its oscillation frequency [21], [48]. However, it could be negatively influenced by the aging process or neurological impairments. For instance, aging populations demonstrate increased GF and larger relative safety margins [30] while neurological populations (e.g. multiple sclerosis [3], stroke [2], Parkinson's disease [49]) commonly have disrupted force coordination, usually reflected through a high GF scaling and low GF-LF coupling (see [50] for review). Therefore, the assessment of GF-LF coordination has recently been considered as a promising method for the development of a clinical test of hand function, and efforts have been made to standardize this technique and to better understand how the CNS controls force coordination [4].

Another critical quality of movement, neuromuscular quickness, is described as a submaximal muscular effort used to complete a movement in the shortest necessary time [37]. The ability to generate a quick submaximal force (e.g. jerking a steering wheel to avoid an accident or thrusting out a hand to grab an externally fixed object after tripping) could affect one's quality of life and is also a key factor in fall prevention in aging and neurological populations. Therefore, efforts have been made to quantify this function.

Kinetic analysis is traditionally utilized for neuromuscular quickness assessments because it both allows for analysis of the force production and clarifies the methods of control utilized by the CNS [23], [37]. In this method, subjects are evaluated through submaximal force production tasks under the instructions to produce them as quickly as possible at varying amplitudes. However, subjects are instructed not to aim for precision in amplitude selection, because it has been shown that the CNS will alter the rise time of the force in order to accurately select a chosen amplitude [23].

Using these brief force pulses, the peak value of force pulse (PF) and its corresponding rate of force development (RFD) is found [38]. Previous findings by Freund [38] and Wierzbicka [41], [42] have shown that PF and RFD have a strong linear relationship. The slope of the regression line drawn to this relationship– the amount the RFD must adjust to match the contraction amplitude– is called the RFD scaling factor (RFD-SF) and may be used as an index of neuromuscular quickness [37], [43]. The R-squared (R^2) value calculated from this regression equation reveals the robustness of the calculated RFD-SF, while the intercept of the regression line could provide quickness information when it is used with RFD-SF. A kinetic study by Bellumori et al. performed on healthy young adults revealed a high RFD-SF along with R^2 values close to one, both of which were highly reliable and generally comparable among muscle groups [37]. Another important variable that could be extracted from the pulses is half-relaxation time, defined as the time it takes for the force to reduce to half of its peak value. This passive release of the forces is prolonged in aging and correlates well with a patient's clinical status [51], [52].

Similar to the indices of GF-LF coordination, the indices of neuromuscular quickness are sensitive the effects of neurological impairments on motor function. For example, it has been found that elderly adults display a decline in the maximal motor unit discharge frequency, and therefore exhibit slower RFD and longer rise times [43], [44].

Furthermore, studies have shown that when subjects are asked to perform rapid, isometric forces as quickly as possible, those with Parkinson's disease have reduced rates of force development and take a longer, more variable time to reach peak force (i.e. lower RFD-SF and R^2) [42], [47]. Also in stroke patients, the RFD-SF and R^2 values were lower in their paretic compared to their non-paretic sides [53]. These findings indicate that kinetic assessment of brief force pulses could be a useful technique for testing neuromuscular function in clinical settings.

Because neurological impairments have a measureable effect on the described variables, the main aim of this study was to develop a noninvasive, clinically useful measurement technique that quantifies force coordination and neuromuscular quickness simultaneously. To further standardize the proposed technique, we also used two different LF directions (i.e. pushing down or pulling up) to generate brief force pulses. The last aim was to determine the number of trials required to ensure a high reliability. A previous study using similar tasks showed day-to-day reliability in fifty pulses [37]. To test the reliability of our measure, we oversampled pulses and computed the indices of GF-LF coordination and neuromuscular quickness from randomly selected subsets of twenty to eighty pulses. We hypothesize that the indices of GF-LF coordination and neuromuscular quickness obtained from brief force pulses will be high and similar to the values observed in the previous literature. We also hypothesized that the indices of neuromuscular quickness obtained from GF and LF will be similar.

2.3 Methods

2.3.1 Subjects. 13 healthy, right-handed adults between the ages of 21 and 31 (five female and eight male, aged 23.2±2.9 years) were recruited. Prior to entering the study, all participants were asked to read and sign an institutionally approved consent form approved by the IRB of Rowan University, and the study was conducted in accordance with the Declaration of Helsinki.

2.3.2 Device. The equipment used in this study (see Fig. 1a) consisted of an instrumented grasping fixture and an adjustable base, which were custom built to resemble those used in previous studies [6], [54]. The grasping fixture was comprised of two parallel plates joined by a single-axis load cell (WMC-50, Interface Inc., USA) and fixed vertically to a tri-axial force transducer (Mini40, ATI Industrial Automation, USA). The handheld portion was fully fixed to the adjustable base, which was secured to a desk. The base height was set for each participant so that, when standing in front of the device, his or her elbow was flexed to 90 degrees while grasping the vertical plates.

The single axis load cell recorded the horizontal compression force (F_C) produced by the subject's precision grip against the two vertical plates, while the tri-axial transducer recorded net forces produced in the F_Y direction. Both of these forces were utilized to calculate the GF, or the average force exerted against both sides of the hand held device. The tri-axial transducer also recorded the horizontal (F_X) and vertical (F_Z) forces, which were used to calculate LF. Figure 1 depicts the positioning of the precision grip as well as the equations used to calculate these variables.

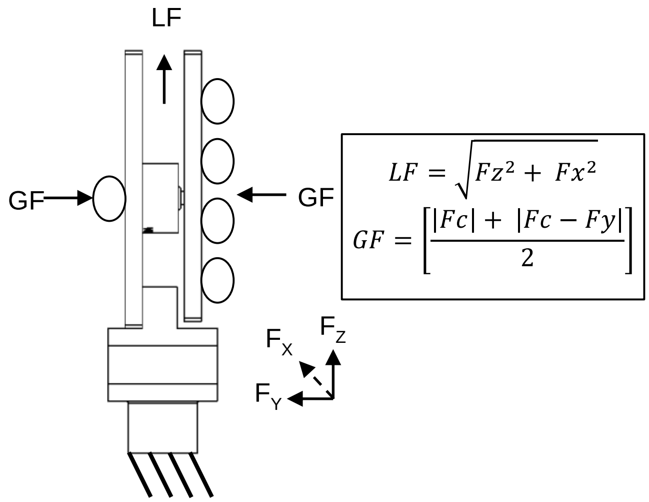

Figure 1. Schematic illustration of the handheld portion of the experimental device with a simplified view of the instructed finger placement. The single-axis force sensor measured the compression force, F_C, while the tri-axial transducer measured F_X, F_Y, and F_Z. These forces were used in the equations shown to calculate the grip force, GF, and the load force, LF.

2.3.3 Procedure. All data for each subject was taken in a single session lasting approximately one hour. Upon arrival, subjects were asked to wash and dry their hands to ensure no natural or artificial contaminants (i.e. sweat or lotion) were present to affect skin frictional properties.

The main study then began with tests of the maximal voluntary contractions (MVC) of GF (GF_{max}) and LF (LF_{max}), independently. The MVC was taken as the largest of three maximal contractions for each force, completed with sixty seconds of rest after each trial [37]. The GF_{max} trial was performed by gripping the parallel force plates as hard as possible using a precision grip. The LF_{max} was tested in both directions by having the subjects either push down or pull up on the device using a precision grip. Both GF_{max} and LF_{max} were also measured at the end of the study (only in the direction of the first set of LF pulses) to determine if the GF and LF producing muscles had fatigued.

Subjects were then asked to generate rapid isometric contractions (i.e. pulses) of varying amplitudes by either pushing down or pulling up on the externally fixed device. Visual feedback in the form of a line graph on a computer monitor displayed pulse amplitudes as a percentage of the subject's MVC in real time (Figure 2). Four red lines appeared on the screen to represent 20, 40, 60, and 80% of their maximum LF. The area between 20% and 40% represented the small area, 40% to 60% was the medium area, and 60% to 80% was the large area. Each subject received identical instruction to "produce forces as quickly as possible without aiming for specific magnitudes." The timing between each pulse was cued by a metronome set at approximately two-second intervals. It is important to note that subjects were told not to target specific force levels, as this has been shown to slow the rate of force production [55]. Following the introduction, subjects were allowed to practice until they were comfortable with performing the instructed discrete force pulses at various amplitudes.

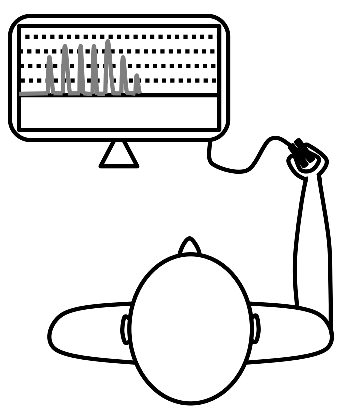

Figure 2. Illustration of experimental conditions: the subject exerted LF (normal to paper) upon the experimental device, and feedback was output on the computer monitor.

Once the subjects were ready, they performed three separate trials of approximately thirty pulses each. The set of trials was repeated twice to analyze both pulse directions: pulling up and pushing down[1]. They were block randomized among subjects. Within each trial, pulses were performed as blocks of five pulses at each of three approximate amplitudes, referred to as “small,” “medium,” and “large.” The order of pulse amplitudes was randomized between trials. This resulted in about ninety pulses spread over the subject’s force range for each LF direction. This number of pulses was selected because it provides a high day-to-day reliability for the selected variables of neuromuscular quickness [37]. In order to avoid muscle fatigue, a short period of rest (approximately 30 seconds) was given after each trial, while a longer rest was granted between each LF force direction.

2.3.4 Data acquisition and reduction. The signals from the two transducers were sent to a National Instruments 16-bit data acquisition board (NI PCI-6224, Austin, Texas, USA). The data was read by a custom-made LabVIEW routine (National Instruments Corp., Austin, Texas, USA), which sampled the data at 200 Hz. Data analysis was performed with another customized LabVIEW routine. The sampled raw data was filtered using a 2^{nd} order low-pass Butterworth filter with a cutoff frequency of 10 Hz. The derivative of each force was calculated using the central difference method and then filtered with a 2^{nd} order low-pass Butterworth filter with a 10 Hz cutoff frequency to reduce the noise due to derivation. The waveforms representing GF and LF and their derivatives (Figures 3-5) were then used to automatically place cursors depicting force

[1] The study performed four sets of trials for two directions (LF up and down) and two frictional conditions (high and low). Only the high frictional condition was reported in the manuscript, but the low frictional condition results may be found in Appendix A.

onset, peak force, half-relaxation, and end. Each force onset was determined to be the time at which the derivative of the force reached 10% of its maximum values [56]. Time to peak was calculated as the time between force onset and peak force. The half-relaxation time was calculated as the point at which the peak force reduced to half of its maximum value during relaxation phase. The average rate of forces was calculated by dividing the peak force by its time to peak [38]. This is used during the calculations of the variables of neuromuscular quickness (see Data Analysis section below). An experienced researcher then visually inspected each pulse for proper cursor placement before saving the values for further analysis. Abnormal pulses, including those with significantly longer times to peak or those with obvious multiple peaks, were removed. In all, the removed pulses totaled less than 2% of the whole body of data from the study.

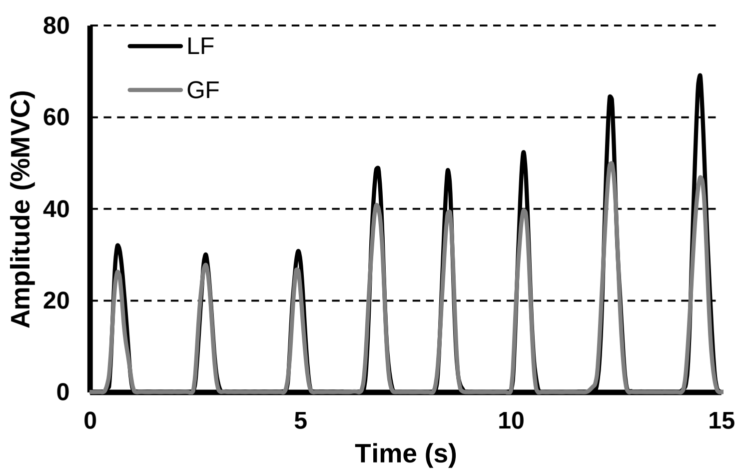

Figure 3. Plot of a series of pulses showing LF (black) and GF (grey) as a percentage of their maximum voluntary contractions

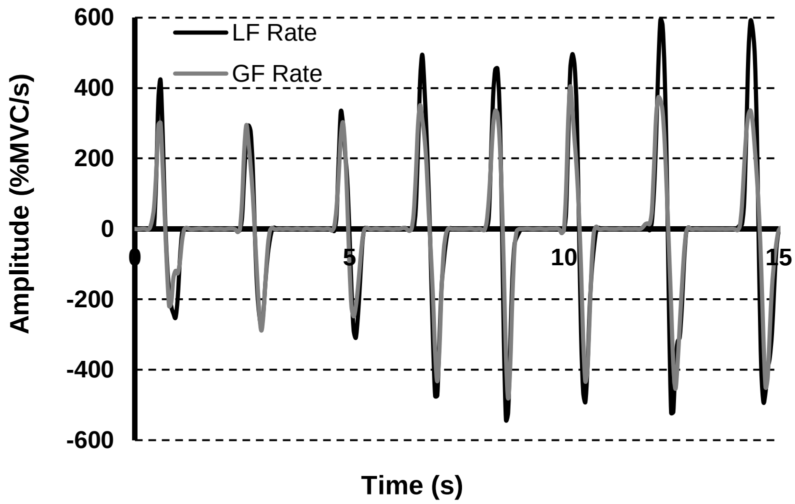

Figure 4. Plot of the LF and GF rates (black and grey, respectively) as a function of the percent of the subject's maximum voluntary contraction per second.

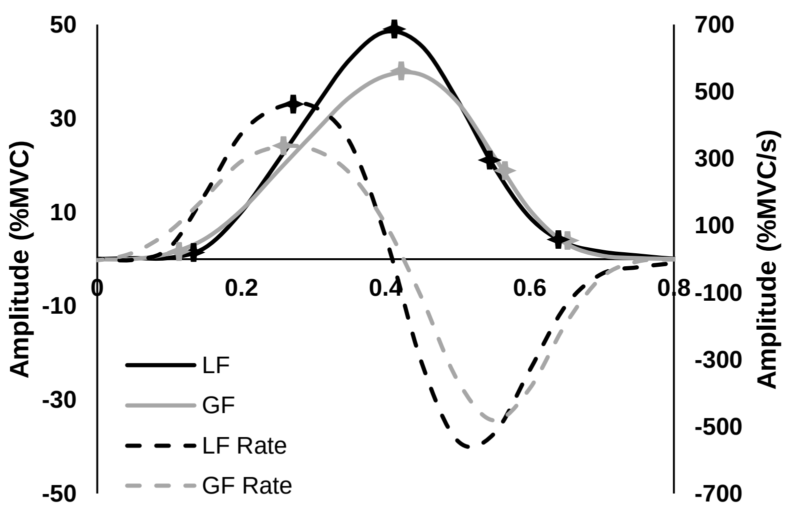

Figure 5. Singular pulse wherein the forces are displayed in %MVC and their rates are shown in %MVC/s (LF is black, GF is grey). The markers show the force onset, peak, half-relaxation, and end.

2.3.5 Data analysis. In line with previous studies, we assessed GF-LF coordination through GF scaling and GF coupling. GF scaling was assessed with the

GF/LF ratio, calculated as the average of the ratio of peak values of GF and LF in each pulse. GF coupling was assessed through the maximum cross-correlation (r_{max}) between GF and LF and the corresponding time lag between the two forces. As r_{max} was not normally distributed, its z-transformed values were used to calculate the averages to be used in the statistical analyses. This was presented as median r_{max} (see Figure 8 in the Results Section).

With respect to the variables of neuromuscular quickness of both LF and GF, we plotted each force pulse's peak (in %MVC) against its rate of force development (calculated by dividing the peak force with its time to peak (%MVC/s)) for each of the directions. We then computed the linear regression line for this relationship (Figure 6). The linear regression parameters (slope, R^2, and intercept) were extracted. The slope of the regression line was used to quantify the amount to which RFD scales with the amplitude of the contraction and was referred to as rate of force development scaling factor (RFD-SF). R^2 was also used here to analyze the robustness of RFD-SF as a controlled variable of the neuromuscular system [37]. Unlike the previous research, we also analyzed intercept. Together with RFD-SF, it can be informative regarding the quickness of individual forces. Finally, we studied the half-relaxation time of both forces because it was shown to be an important variable that correlates strongly with a patient's clinical status [51].

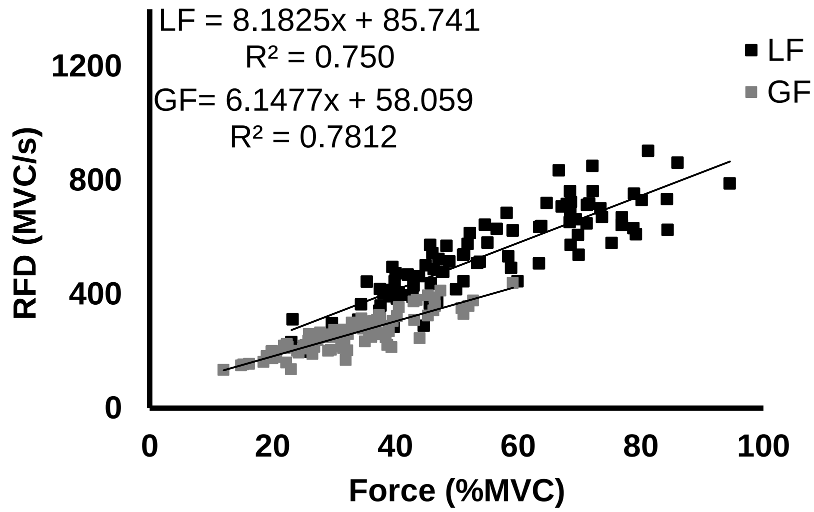

Figure 6. Plot of the forces versus their rates of force development gives a linear regression line, wherein the slope is the RFD-SF.

To test the effects of force direction (up vs down) on each dependent variable of coordination and quickness, we used paired samples t-tests and 2x2 repeated measures (RM) ANOVAs, respectively. For the tests, force was taken as either GF or LF and direction was either pulling up or pushing down. The p values were set to 0.05 and statistical tests were performed with SPSS 21. The Fischer z-transform was utilized to normalize the data for r and R^2. To determine the minimum of number of pulses required to report a reliable result, we run intra-class correlation ($ICC_{3,1}$) analyses on each of the selected coordination and quickness variables calculated from random samples of 20, 40, 60, and 80 pulses.

2.4 Results

Among the total 2,435 pulses collected from all subjects in both force directions, we excluded 37 (1.52%) during visual inspection[2]. Excluded pulses are those that have

[2] Including the low frictional condition, at total of 4,866 pulses were collected, and we excluded 75 (1.54%).

clear double LF peaks or that lasted more than mean+2SD of the remaining pulses. To test whether our experimental protocol led to fatigue, we collected GF_{max} and LF_{max} prior to data collection (average of 125.6 N and 89.52 N, respectively) and compared them to those collected following the completion of the experiment (131.7 N and 90.29 N for GF_{max} and LF_{max}, respectively) by using a paired samples t-tests. No significance was found for either force ($p>0.05$), indicating that the subjects did not experience fatigue throughout the trials.

2.4.1 GF-LF coordination. GF-LF coordination obtained from quick pulses was assessed through GF scaling (GF/LF ratio obtained from absolute peak force values) and GF-LF coupling (maximum cross-correlation coefficient, r_{max}, and the corresponding time lags). A paired samples t-test performed on the GF/LF ratio indicated no significant difference between the up and down directions ($t=0.97$, $p>0.05$; Figure 7). Regarding GF-LF coupling, results of the paired sample t-tests conducted on z-transformed values of r_{max} indicated a higher value in up than in down direction ($t=4.801$, $p<0.001$*);* Figure 8). Regarding time lags, results indicated similar values ($t=1.66$, $p>0.05$) in both directions. Because the average lags were close to zero (up: 2.8±4.6 ms; down: 0.69±2.0 ms), we did not include them in a figure.

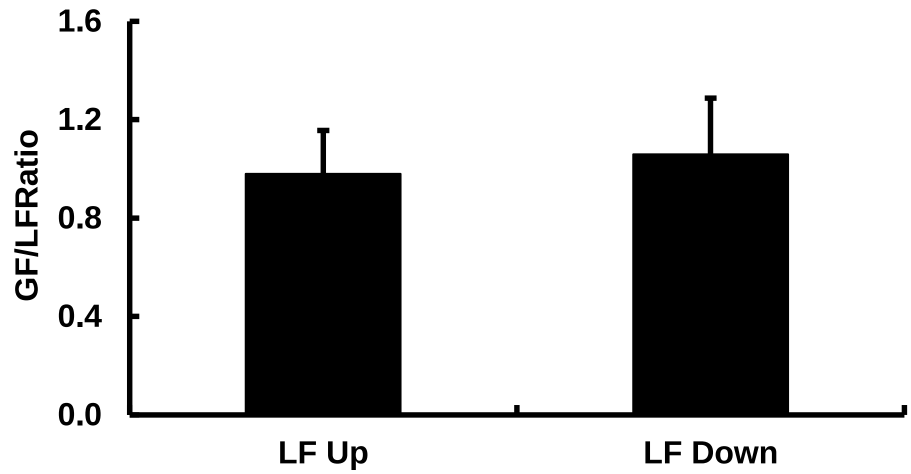

Figure 7. First force coordination variable: GF/LF ratio averaged across participants for both LF directions (pulling up and pushing down). *p<0.001

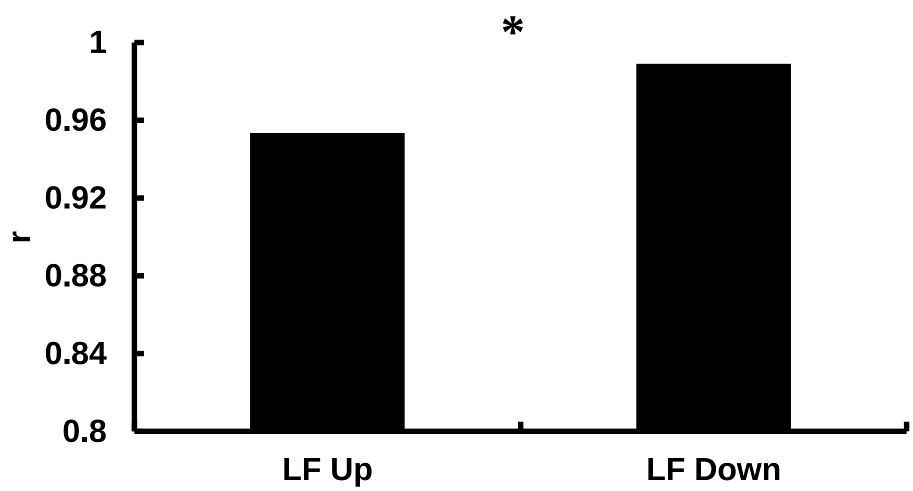

Figure 8. Second force coordination variable: median cross-correlation coefficient, r, for both LF directions. *p<0.001

2.4.2 Neuromuscular quickness. The studied variables of include those obtained from the regression line drawn between peak force and its rate (i.e. RFD-SF, R^2, and intercept) and the half-relaxation values of both LF and GF in both up and down directions. A 2 (direction: up vs down) x2 (force: LF vs GF) repeated measures (RM) ANOVA performed on RFD-SF revealed significant main effect of force ($F(1,12)=14.08$, $p<0.01$) but not a main effect of direction ($F(1,12)=1.73$, $p>0.05$; Figure 9). Also found was a significant direction and force interaction ($F(1,12) = 8.74$, $p<0.05$). When the pairwise comparisons were performed, $RFD\text{-}SF_{LF}$ was higher than $RFD\text{-}SF_{GF}$ ($p<0.001$) in the up direction while they were similar in down direction ($p>0.05$). Regarding forces, $RFD\text{-}SF_{LF}$ in up was higher than the $RFD\text{-}SF_{LF}$ in down ($p<0.05$) while $RFD\text{-}SF_{GF}$ was similar in both directions ($p>0.05$). In general, results revealed a higher RFD-SF for LF than for GF, and this difference was larger in pulses produced in the up direction. Another RM ANOVA performed on the z-transformed R^2 revealed neither the main effects of direction ($F(1,12)= 0.01$, $p>0.05$), force ($F(1,12)= 3.90$, $p>0.05$), nor their interaction ($F(1,12)= 0.7$, $p>0.05$; Figure 10). This indicated highly stable RFD-SFs in both directions for both GF and LF. Regarding the intercept, results revealed neither the main effect of direction ($F(1,12)=1.945$, $p>0.05$) nor the direction and force interaction ($F(1,12)= 0.004$, $p>0.5$). However, there was a main effect of force $F(1,12)=6.064$, $p<0.05$) indicating a higher intercept for LF than those obtained from GF (Figure 11). This indicates that LF is always faster than GF, even at lower force amplitudes. Finally, half-relaxation time did not reveal any main effect of direction ($F(1,12)=1.43$, $p>0.05$) or force-direction interaction ($F(1,12)=0.6$, $p>0.05$), but it did have a significant main effect of force, with $F(1,12)=5.32$, $p<0.05$. This indicated longer half-relaxation for GF in both directions (Figure 12).

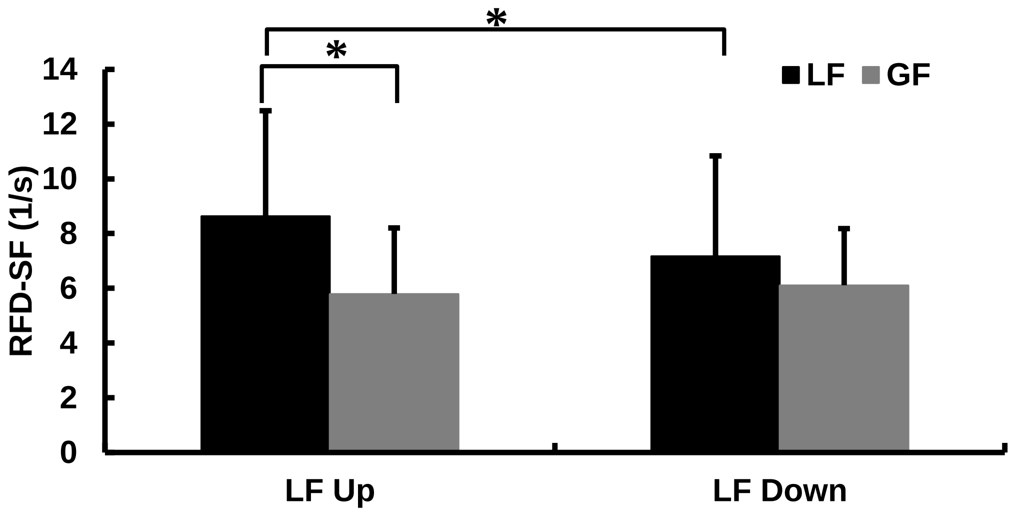

Figure 9. Average subject data depicting the neuromuscular quickness variable of RFD-SF for GF and LF obtained from pulses produced in the LF up and down directions. *p<0.05.

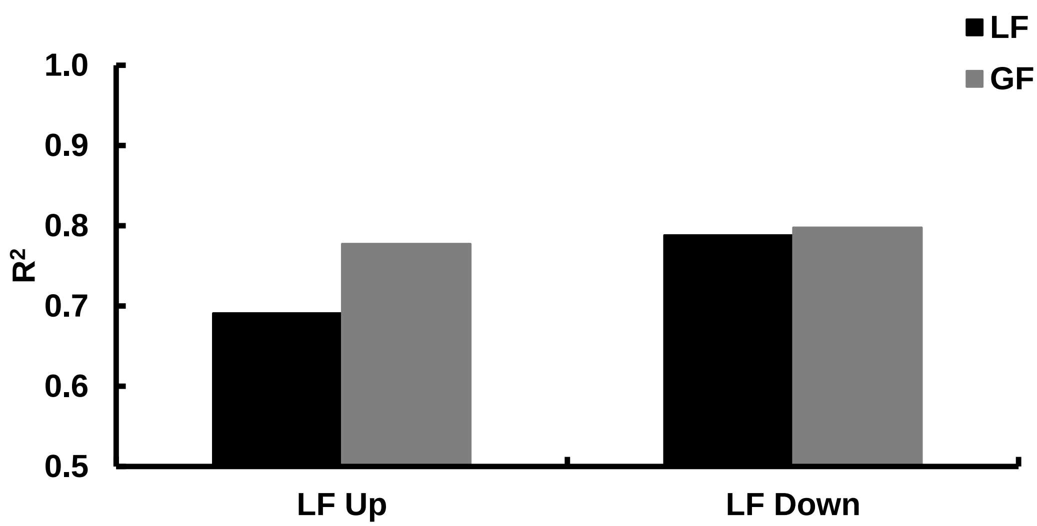

Figure 10. Average subject data depicting the neuromuscular quickness variable R^2 for GF and LF obtained from pulses produced in the LF up and down directions.

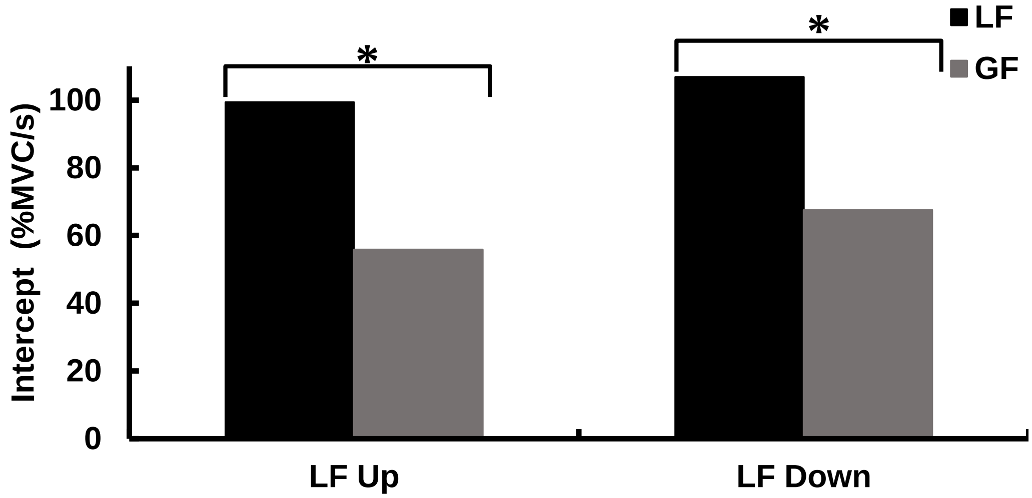

Figure 11. Average subject data depicting the intercepts of the linear regression lines for GF and LF obtained from pulses produced in the LF up and down directions. *p<0.05.

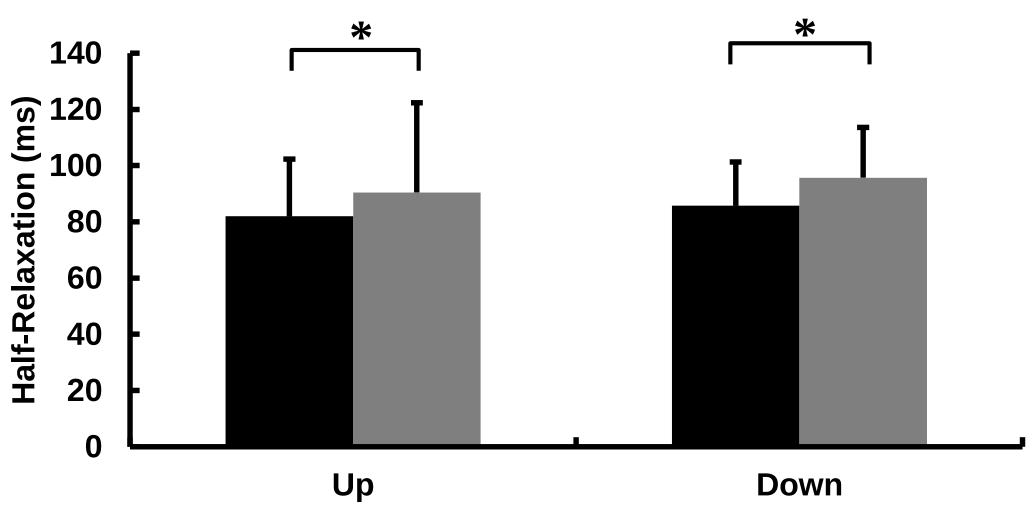

Figure 12. Average subject data depicting the neuromuscular quickness variable of half-relaxation for GF and LF obtained from pulses produced in the LF up and down directions. *p<0.05.

2.4.3 Number of pulses required. Among all the generated pulses, 20, 40, 60, and 80 pulses were randomly selected, and an ICC analysis was run for all the variables in both force directions. Results indicated excellent ICC scores (i.e. ICC>0.9) for every variable of GF-LF coordination and neuromuscular quickness (Table 1).

Table 1

Intra-class correlation coefficient ($ICC_{3,1}$) with 95% confidence interval, its F value, standard error of the measure (SEM%) as the percentage of mean, and mean (standard deviation) from randomly selected 20, 40, 60, and 80 pulses.

Variable		ICC(3,1) (95%CI)	F value	SEM%	Mean (stdev)
GF/LF ratio	Up	0.995 (0.987-0.998)	729.8*	0.051	0.98 (0.01)
	Down	0.994 (0.987-0.997)	699.2*	0.045	1.05 (0.01)
r	Up	0.992 (0.981-0.997)	486.3*	0.052	1.84 (0.01)
	Down	0.993 (0.983-0.998)	549.9*	0.048	2.47 (0.01)
$RFDSF_{LF}$ (1/s)	Up	0.963 (0.917-0.987)	104.7*	0.645	8.50 (0.29)
	Down	0.981 (0.956-0.993)	205.5*	0.148	7.20 (0.07)
$RFDSF_{GF}$ (1/s)	Up	0.967 (0.926-0.989)	119.1*	0.267	5.73 (0.08)
	Down	0.961 (0.913-0.986)	99.7*	0.176	6.19 (0.06)
R^2_{LF}	Up	0.961 (0.913-0.986)	99.9*	0.532	1.08 (0.03)
	Down	0.901 (0.790-0.964)	37.3*	1.419	1.12 (0.05)
R^2_{GF}	Up	0.949 (0.887-0.969)	75.2*	0.482	1.20 (0.03)
	Down	0.902 (0.793-0.965)	38.0*	0.885	1.23 (0.04)
$Intercept_{LF}$ (%MVC/s)	Up	0.920 (0.828-0.972)	47.1*	3.859	99.33 (13.6)
	Down	0.944 (0.876-0.980)	68.1*	0.899	106.87 (4.1)
$Intercept_{GF}$ (%MVC/s)	Up	0.937 (0.863-0.978)	60.8*	1.569	55.83 (3.5)
	Down	0.937 (0.891-0.978)	60.1*	1.025	67.51 (2.8)
$Half\text{-}relax_{LF}$ (ms)	Up	0.993 (0.985-0.998)	608.7*	0.084	81.93 (20.4)
	Down	0.989 (0.975-0.996)	371.6*	0.051	85.7 (15.6)
$Half\text{-}relax_{GF}$ (ms)	Up	0.992 (0.982-0.997)	523.1*	0.066	90.42 (31.9)
	Down	0.986 (0.968-0.995)	287.3*	0.002	95.57 (17.2)

*$p<0.001$. Note that cross correlation (r) and R^2 are represented as z-transformed values.

2.5 Discussion

The purpose of this study was to develop a measurement technique that uses a static object manipulation task to assess both hand function and neuromuscular quickness simultaneously. Our results indicate that (1) the proposed measurement technique is capable of extracting the variables of GF-LF coordination and neuromuscular quickness simultaneously (2) the indices of GF-LF coordination could be higher in LF pulses produced in down direction as compared to those produced in up direction (3) the indices of neuromuscular quickness of LF is higher than those of GF (4) the selected variable of GF-LF coordination and neuromuscular quickness obtained from randomly selected 20 pulses are highly reliable.

2.5.1 Coordination. No other study has observed GF-LF coordination in quick force pulses, but the above results show elaborate force coordination in both directions with GF/LF ratios and cross-correlation values comparable to previous studies. This includes such publications as those that varied surface friction [5], LF force direction (bidirectionality) [19], and LF frequency and range [21]. The relatively low, stable GF/LF ratio is consistent with the findings of Johansson and Westling, who noted that the force ratio was constant among individuals and suggested that regulated by the CNS during manual manipulation [7]. The pulses are performed so fast that feedback components such as surface friction do not have time to affect the force magnitudes. The CNS therefore maintains a relatively constant ratio to allow for faster force production instead of acting more slowly to determine the best ratio for each specific movement. Furthermore, the correlation values were very high, and there were no significant time lags, indicating predictive feed-forward control mechanisms governing brief force production tasks [48]. This desirable GF-LF coupling further supports the findings of Johansson and Westling, who found that during times of parallel force expression, no systematic time lag disturbed the balance between the two forces [7]. We can therefore say that utilizing a method of quick force pulses results in the desired elaborate coordination.

Though the cross-correlation values are high in both LF directions, it is significantly larger in the down direction. The observed differences between directions could be due to the role of skin receptors in coordination tasks [7]. It is possible that the change in LF direction alters the pattern of skin receptor afferent firing, which could result in use of a separate, distinct muscle synergy [19], [57]. Also, the observed higher GF-LF coupling in LF pulses produced in down direction could indicate that this direction alone should be utilized for future studies. This is a logical choice, as the task is seen often in such scenarios as grasping a cane or reaching out to clutch a handrail to avoid a fall. Because the correlation is higher, and considering the downward direction is more ecologically valid, it is possible that only the pulses produced downward direction may be used in future studies, thereby shortening the testing time and reducing the likelihood of fatigue.

2.5.2 Quickness. Along with the GF-LF coordination variables, we also successfully extracted those related to neuromuscular quickness. The RFD-SF_{LF} values found in our study were similar to those reported in Bellumori et al.'s, in which participants performed isometric force pulses by pushing down on a grasped instrumented stick [37]. Please note that in their study of elbow extensors, although GF was applied to produce LF pulses, only LF was pulses were analyzed. A separate study by Wierzbicka et al. studied the LF from elbow flexors in such a way that GF was not involved [42]. While our results were generally similar to this publication, our $RFD\text{-}SF_{LF}$ values were lower. However, while those studies only measured LF, our technique collected data on GF as well. This allowed for comparison between the two forces. Surprisingly, we found that $RFD\text{-}SF_{LF}$ was generally greater than $RFD\text{-}SF_{GF}$ and the LF intercept was higher than that of GF indicating a quicker LF production than GF production. Therefore, it is possible that GF may limit the quickness of LF because GF is unable to reach its maximum in the time it takes to complete a pulse. This could explain the differences in $RFD\text{-}SF_{LF}$ values between this study and that of Wierzbicka et al., as GF was not utilized in their study and, therefore, could not limit the rate of force production [42]. Future studies assessing the indices of quickness for LF and GF when they act individually is required to determine the effect of each force on the other.

Previous findings by Freund [38] and Wierzbicka [41], [42] have shown that PF and RFD have a strong positive linear relationship. The ability to create linear regression lines with high R^2 values show that this is the case for our testing within 20-80% of the subjects' MVCs. Furthermore, the consistently high R^2 values may indicate an invariant control strategy utilized by the CNS in controlling brief force production tasks. This evaluation could be valuable for comparisons to neurologically impaired populations, as elderly populations have lower RFD-SF and R^2 values [43], while Parkinsonian patients have exhibited difficulty in producing rapid contractions and reduced R^2 [42], [56].

Regarding the half-relaxation times, we found that the values of GF were consistently higher than those of LF. This indicates the use of a safe strategy, wherein the subject maintains a reduced rate of GF relaxation on the object to avoid slippage during load reduction. This will be a useful variable to assess when looking to compare healthy to impaired populations, as muscle relaxation has already been shown to slow in aging and Parkinson's disease [51], [52]. Moreover, Corcos et al. noted that, of the assessed quickness variables, relaxation was the most sensitive to disease severity in Parkinson's disease, likely due to the patients' inability to switch off agonist muscle activity [51]. Its ability to correlate highly with a patient's clinical status makes relaxation time invaluable as a diagnostic tool.

2.5.3 Reliability. Bellumori et al. showed that the use of rapid force pulses had a high day-to-day reliability for 25 pulses in elbow extensors and about 50 pulses in general [37]. We oversampled the number of pulses during data collections and run a reliability analyses among randomly selected 20, 40, 60, and 80 pulses. We found that a random selection of 20 pulses provided highly reliable results as compared to 80 pulses. We suggest that 20 brief pulses performed at various submaximal levels between 20-80% MVC is adequate to extract indices of GF-LF coordination and neuromuscular quickness reliably. The ability to reduce the number of pulses per testing session would greatly increase the viability of this method for use in a clinical setting, as it would decrease testing time, cost, and patient fatigue.

2.6 Conclusion

This study, which is the first to combine GF-LF coordination and neuromuscular quickness into the same technique, successfully extracted all desired variables. While studies have examined neuromuscular quickness before, they limited themselves to studying only LF and failed to note the effect of each force on the other [37], [42]. This technique is therefore more robust than its predecessors because it not only combines two assessments, it studies them more deeply by measuring both LF and GF. Also, because a comparison of the initial and final maximum force values showed no fatigue, the trial outcomes are readily acceptable. Further study is needed to note the effects of aging and neurological disease on these variables under the discussed conditions.

Chapter 3

Overall Conclusions and Future Work

The study undertaken in this thesis was successful in its goal of evaluating GF-LF coordination and neuromuscular quickness within the same measurement for healthy, young adults. We evaluated all indices of force coordination and neuromuscular quickness for two frictional and two directional conditions, and unlike previous studies, both GF and LF were quantified. A comparison of the high and low frictional conditions showed little difference between the two. The differences elicited no novel findings or insight into the mechanisms of the central nervous system. Therefore, we chose to focus on the high frictional condition alone, as verbal feedback from the subjects indicated that they were more comfortable manipulating that surface and higher frictions require lower GF, so fatigue is minimized. For these reasons, it is suggested that future studies utilize only one frictional condition, and that condition should have a relatively high coefficient of friction.

Utilizing our chosen frictional condition, GF-LF coordination was found to be elaborate in both directions, as similar G/L ratios, high correlation values, and low time lags indicated that the CNS maintains a high coordination. These values are similar to previous studies, so our method of obtaining coordination variables via quick force pulses is valid. Furthermore, as the downward direction shows higher correlation values and is more ecologically valid for instances of falls (i.e. grasping a cane), this direction should be utilized in future studies.

The neuromuscular quickness variables were similarly well-acquired from the quick force pulses. In a surprising find, the RFD-SF and intercepts were consistently higher for LF than for GF. These results indicated a faster LF production than GF production, which raised the question of whether GF is a limiting factor in the movement. Further testing, wherein GF and LF are tested independently of each other, is needed to determine their individual quickness values. This study would be beneficial because it could reveal which force would benefit more from training. For instance, if GF is found to be slower, then the hand muscles could be trained to move more rapidly, thereby allowing the entire arm movement to be quicker. Furthermore, the high RFD-SF and intercept values and the insignificant effect of direction on the R^2 values indicated high stability and an invariant control strategy. Longer half-relaxation values in GF also signified a safe strategy of load reduction prior to releasing the object.

Using this collected data, a reliability analysis showed that the testing method is highly reliable with as little as twenty pulses. Thus, future studies may be conducted with significantly fewer pulses per trial, thereby limiting the possibility of subject fatigue.

Now that the baseline data has been collected for young, healthy individuals, it is necessary to broaden the scope of the population data. It is suggested that such groups as multiple sclerosis and Parkinson's disease patients, as well as older adults, should be recruited. The comparison of variables between the healthy and impaired populations will both give insight into the effects of neurological impairments and affirm the validity of this test in perceiving those differences. The ability of this technique to quantify the variations in these coordination and quickness variables will make it an invaluable diagnostic and assessment tool in rehabilitative settings.

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Appendix A

Device Design Documentation

Purpose

The system was designed to measure both the grip force and the load force exerted by a subject during object manipulation tasks. The design for the device utilized in this study was based off of those previously used in studies by Jaric et al., [58], Krishnan and Jaric [59], and Uygur et al. [60]. It was built to be used for a variety of tasks and conditions including single and bimanual manipulations in both the vertical and horizontal directions.

Requirements

The device had to be:

- Able to measure data for both the load (vertical) force and the grip (horizontal compressive) force.
- Compatible with data acquisition equipment.
- Height adjustable to work for every subject.
- Usable for multiple experimental setups (i.e. free vs fixed object manipulations).

Overview

The system consists of two main device components and a sensor system, including:

- A height adjustable base
- A handheld device
- A force sensor, amplifier, and data acquisition circuit

The assembly (Figure 13) consists of two parts, the grasping fixture and the adjustable base. The grasping fixture contains two parallel plates, on which the subjects place the pads of their fingers in a precision grip. While a power grip utilizing the entire hand would be more applicable to the situations we are trying to mimic, the sensor array necessary to accurately measure it would be far more complex and costly. The grasping fixture is fixed to the height adjustable base by a single screw, making it a simple task to remove the fixture for free object manipulation tasks. For fixed tasks, however, the handheld portion can be secured either vertically or horizontally to the base. The base was designed for two handheld fixtures to be attached at the same time, allowing for the study of bimanual tasks. The base also has a telescoping vertical structure, which allows it to raise or lower to match the height of the subject's elbow.

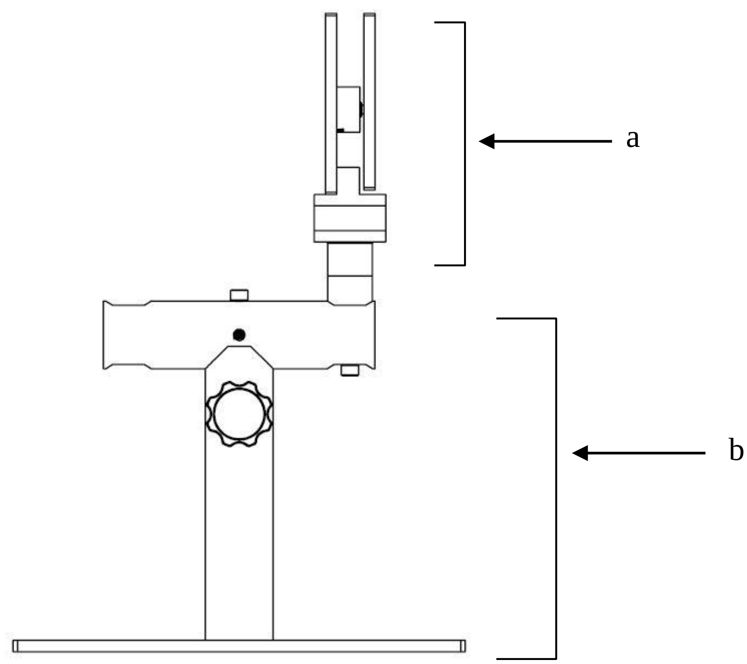

Figure 13. Assembly drawing of the test device showing (a) the grasping fixture and (b) the adjustable base.

A diagram of the sensor system is shown below (*Figure 14*). The handheld grasping fixture contains the two transducers. The WMC-50 load cell (Interface Inc., U.S.A.) measures the compressive grip force while the Mini40 tri-axial force transducer

(ATI Industrial Automation, U.S.A.) collects the vertical load and eccentric force data. These sensors send the signal through their individual amplifiers, which perform some smoothing and prepare the data for the data acquisition (DAQ) equipment (NI PCI-6220, National Instruments, U.S.A). The DAQ connects to the computer, and then inputs the data to a LabVIEW (2013) VI, which both outputs the waveforms and saves it for analysis.

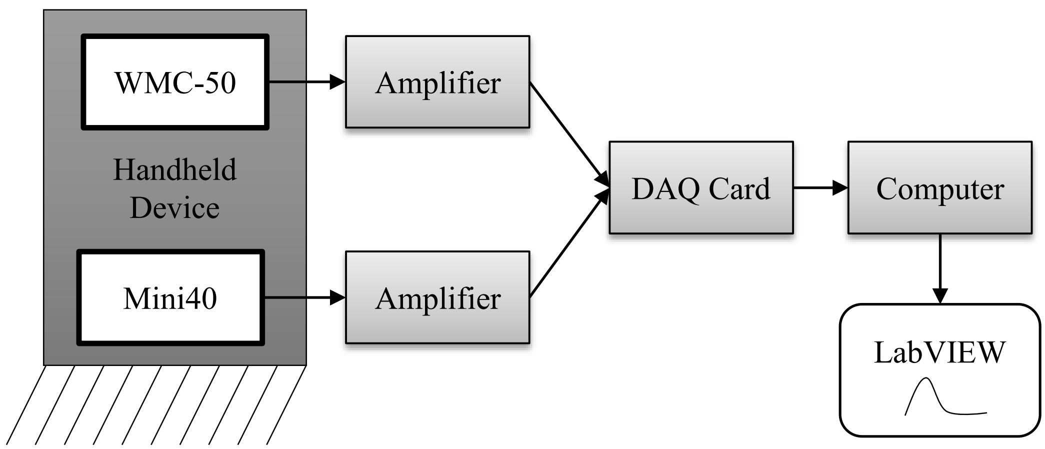

Figure 14. A diagram of the device system, wherein the two force sensors, the WMC-50 and the Mini40, are collect data in the handheld device and send it through amplifiers and a data acquisition (DAQ) card to a LabVIEW program.

Device Modifications

Of major concern during the design of this device was the fact that human subjects do not pull upwards or push downwards perfectly every time. If their posture is not correct during pulse production (i.e. they lean forward or pull sideways), they could produce undesired tensile, compressive, or torsional forces on the grasping fixture. The Mini40 tri-axial force transducer was utilized here because it can measure all three directions (x, y, and z) as well as all three moments. These data are utilized in the calculations for grip force and load force to minimize error, but there was still a concern that an overly large eccentric force (i.e. anything other than the vertical, z, direction) would overtax the sensor and hinder its accuracy.

To reduce the possibility of damage to the sensor, an optional ball-and-socket joint was added to the assembly (Figure 15). The joint was placed between the base and the handheld portion and tightened so that the assembly would stand vertically on its own but would allow the grasped portion to rotate in the presence of an excessive lateral force. To ensure that the handheld portion could be utilized with or without the ball-and-socket joint, an adaptor was created to match the screw threads. This modification allowed for a large range of setup options while also protecting the device from damage.

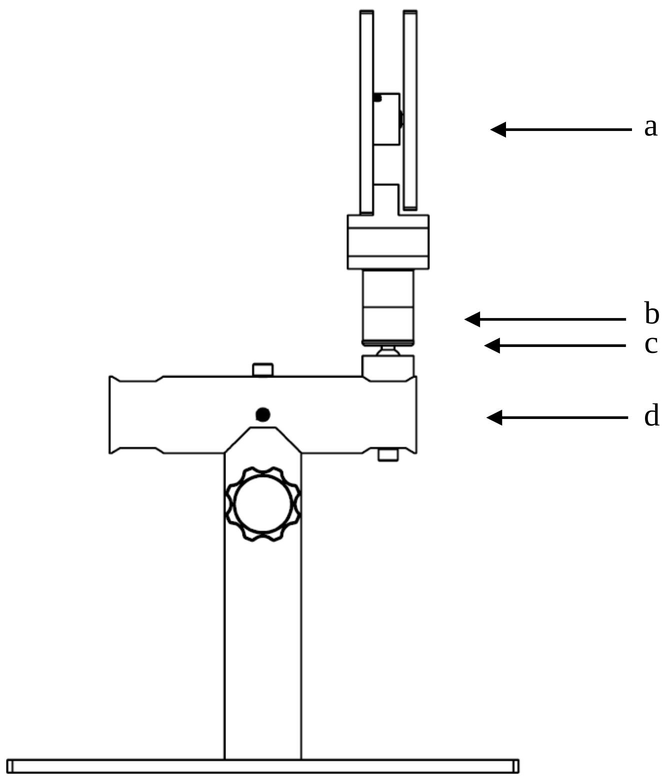

Figure 15. A drawing of the device, including (a) the handheld fixture, (b) the adaptor, (c) the ball-and-socket joint, and (d) the adjustable base. The ball-and-socket joint and adaptor were added to this assembly to reduce the possibility of damage to the tri-axial transducer due to eccentric forces.

Appendix B

Results for the Low Frictional Condition

As discussed in the introduction, this thesis utilized two frictional conditions. The lower frictional condition was not included in the manuscript because what little difference was present between the high and low frictions was of minimal benefit to the understanding of the coordination and quickness measures. High friction was therefore utilized in the manuscript because subjects reported feeling far more comfortable performing the tasks in that condition. For comparison purposes, the results of the low frictional condition are reported below.

Coordination

Utilizing a 2x2 repeated measures ANOVA (direction x friction), the GF/LF ratio revealed significant interaction: $F(1,12)= 8.237$, $p<0.05$. Pairwise comparisons showed that direction had no effect on the ratio ($p>0.05$), but in both directions low friction led to a larger GF/LF ratio than high friction ($p<0.01$ in up and $p<0.05$ in down). This relationship is clear in Figure 16, as both of the low friction bars are higher than those of high friction. This suggests that the surface friction, but not the movement direction, affects the way in which GF and LF work together during object manipulation.

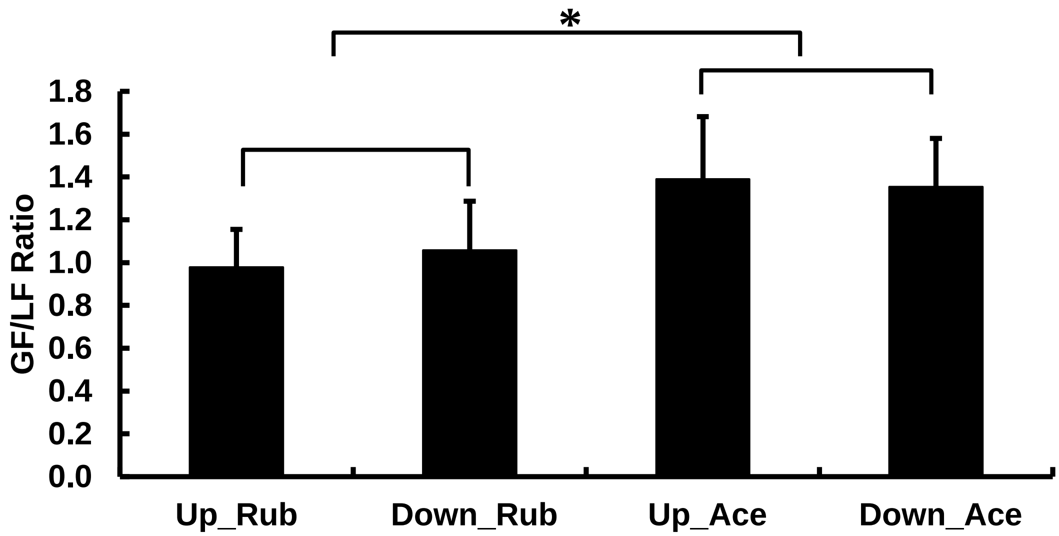

Figure 16. Graph depicting the GF/LF ratio for all four conditions. The higher friction (rubber) conditions clearly required a lower GF/LF ratio than the lower frictional (acetate) conditions.

There was no significant interaction between the correlation (Fisher transformed) values of GF and LF. However, main effects showed that $R_{DOWN}>R_{UP}$ $F(1,12)=21.783$ ($p<0.001$). The graph in Figure 17 supports this finding, as the correlation values are clearly higher in the downward direction for both surfaces. We also analyzed the time lag, which revealed no significant interaction in the ANOVA. Therefore, the CNS must send signals to the muscle groups of both forces at the same time.

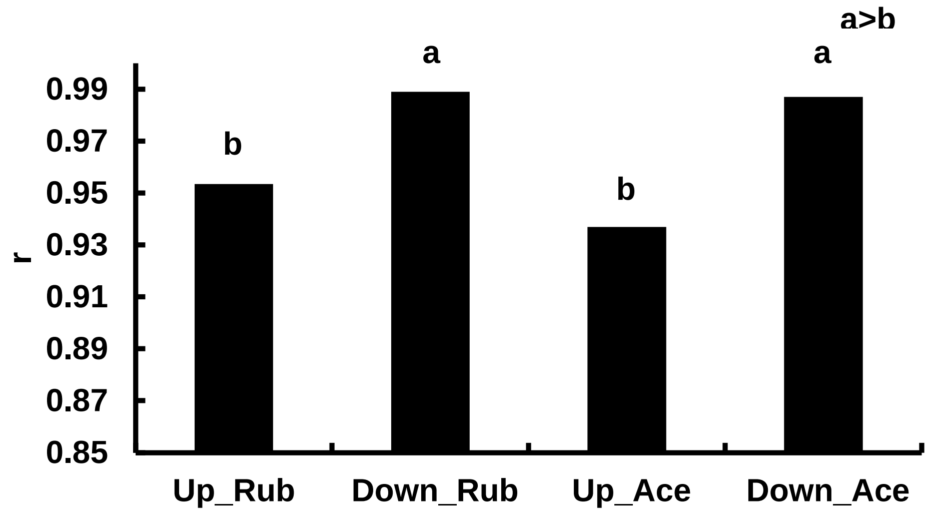

Figure 17. Cross-correlation values for all four conditions.

Quickness

Please note: the RFD-SF and R^2 statistics were calculated differently in this section than what appears in the manuscript. This appendix utilizes the original calculation of the rate, as found with the derivative of the force curve, rather than an average of the values. Therefore, while the statistical relationships are similar for both sections, the F values are differ slightly.

RFD-SF. The 3-way ANOVA showed no significant 3-way interaction. However, the main effect of direction x force was significant, with $F(1,12)=35.781$, $p<0.001$, and the force main effect was significant, with $F(1,12)=31.804$, $p<0.001$ (Figure 18).

A 2 (direction) x2 (force) ANOVA was also performed using the high friction conditions. Results revealed significant interactions, with $F(1, 12) = 27.48$, $p<0.001$. When the pairwise comparisons were performed, in the up direction $RFD\text{-}SF_{LF}$ > $RFD\text{-}SF_{GF}$ ($p<0.001$). In the down direction, $RFD\text{-}SF_{LF}$ was not statistically different from

$RFD\text{-}SF_{GF}$ ($p>0.05$). Furthermore, there was no significant difference for LF, but $RFD\text{-}SF_{DOWN} > RFD\text{-}SF_{UP}$ ($p<0.05$) for GF.

The same 2x2 ANOVA was performed using the low friction conditions, and the results revealed significant interactions: $F(1,12)=24.535$, $p<0.001$. Pairwise comparisons likewise showed that $RFD\text{-}SF_{LF} > RFD\text{-}SF_{GF}$ ($p<0.001$) in the up direction and in the down direction $RFD\text{-}SF_{LF} > RFD\text{-}SF_{GF}$ ($p<0.05$). Further pairwise comparisons illustrated that $RFD\text{-}SF_{UP} > RFD\text{-}SF_{DOWN}$ ($p<0.05$) for LF, and $RFD\text{-}SF_{DOWN} > RFD\text{-}SF_{UP}$ ($p<0.05$) for GF. Given the overall outcome, it can be said that $RFD\text{-}SF_{LF}$ is faster than $RFD\text{-}SF_{GF}$ regardless of direction or friction.

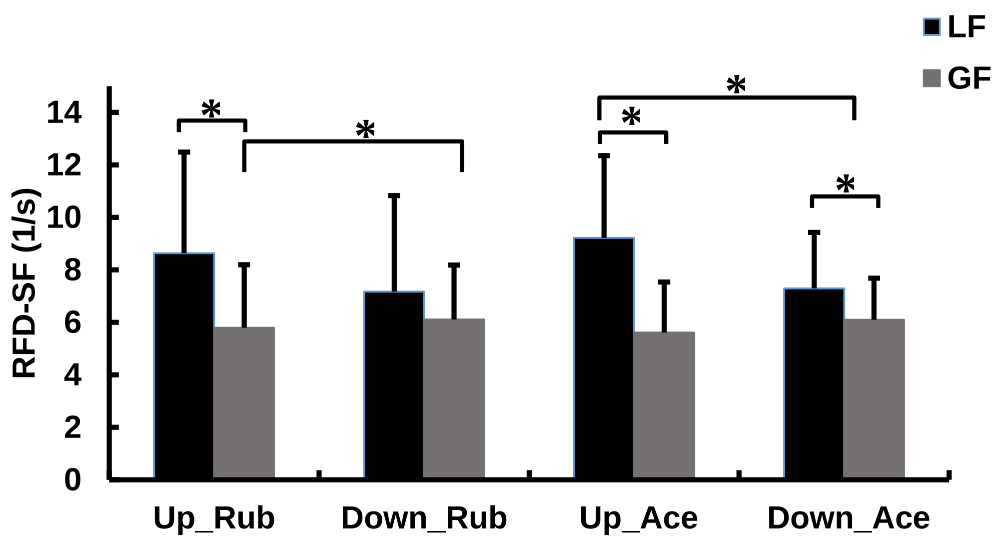

Figure 18. Rate of force development-scaling factor for all four conditions, where LF is black and GF is grey.

R^2. The 3-way ANOVA showed no significant interaction, but the main effect of direction x force did display significance ($p<0.05$).

The 2x2 repeated measures ANOVA for the high frictional conditions revealed no significant interaction, but that for the low frictional conditions did, $F(1,12)=7.950$, $p<0.05$. However, no pairwise comparisons showed significance (Figure 19).

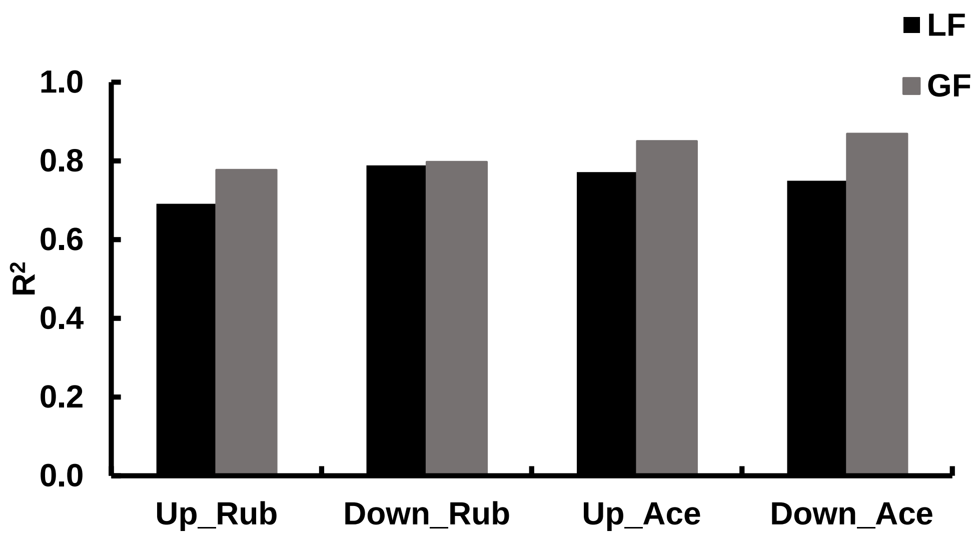

Figure 19. R^2 values for all four conditions, where LF is black and GF is grey.

Half-Relaxation. The 3-way ANOVA showed no significant interaction, but the main effect of force was significant ($F(1,12)=7.146$, $p<0.05$).

Neither the high nor the low friction 2x2 ANOVAs revealed significant interactions, but both had significant main effects of force, with $p<0.05$ for each. However, the graph of half relaxation, Figure 20, shows that LF consistently relaxes faster than GF for all conditions.

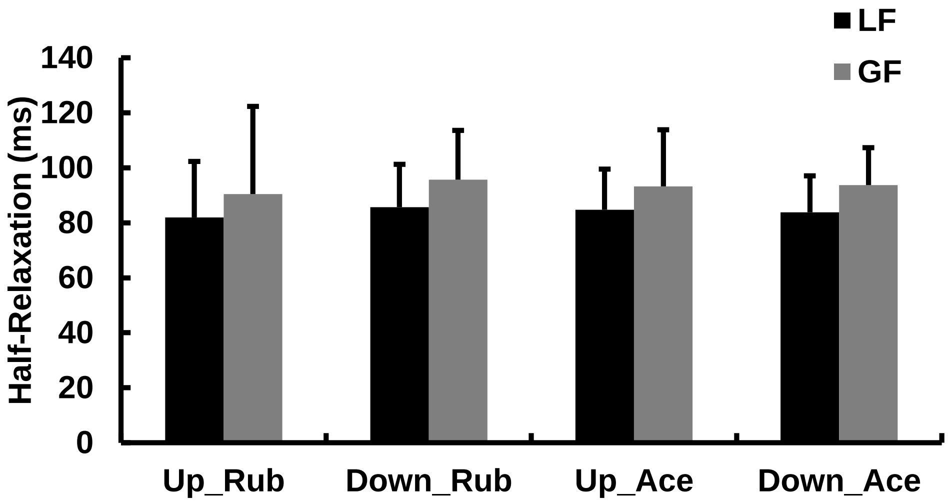

Figure 20. Graph of half-relaxation showing that LF (black) consistently takes less time to relax than GF (grey) for both directions and frictional conditions.