Effect of Age, Elbow Muscle Co-contraction Level, and Elbow Moment Loading Characteristics on Elbow Angle Positional Variability in Postural Tasks

by

Mark T. Gordon

A dissertation submitted in partial fulfillment
Of the requirements for the degree of
Doctor of Philosophy
(Mechanical Engineering)
In The University of Michigan
2012

Doctoral Committee:

Professor James A. Ashton-Miller, Chair
Associate Professor Richard E. Hughes
Associate Professor Riann M. Palmieri-Smith
Assistant Professor Kathleen H. Sienko

Acknowledgements

I would like to take this opportunity to thank everyone who has helped me throughout my graduate school years. First I would like to thank God, without whom this would not have been possible. He has kept me centered throughout my graduate school years and carried me through the times of doubt and desperation.

Secondly, I'd like to thank my family. I'll start with my wife Catherine whom I married two years before my defense (to the day). I love you very much. Thank you for sticking with me, encouraging me, and being there for me in general. Next I'd like to thank my parents Joyce and Todd Gordon for their support. They have been incredibly encouraging while not being overbearing. Someday I hope to learn how to do that as well. I'd like to thank my sister Hilary. She has been there for me through the years when I needed to talk to someone and has always brought a grounded perspective to my situations. Thank you to my extended family: both natural and in-law. Being able to take a step back from graduate school and realize what is really important has served me well over the past years. Also a special thanks to my sister-in-law and illustrator Becky Bartlett who drew many of the figures in the thesis.

I'd like to thank everyone in the Biomechanics Research Lab. In particular my advisor Dr. Ashton-Miller has guided me through the graduate school process. The various members of the BRL have provided everything from technical and physical help setting up my experiments to distractions and good friendships.

I'd like to thank my committee starting with Dr. Hughes who not only provided critical input, but also took the time to meet with me to resolve the issues including the derivations. I'd like to thank Dr. Sienko and Dr. Palmieri-Smith who also contributed to my research by bringing up issues at my preliminary exam that needed to be addressed.

I'd like to thank the friends that I made over my graduate school years. In an attempt to not leave people out I'll list groups and let you decide where you fit: Graduate Christian Fellowship, various Bible studies, Huron Hills Church, Lindquist dinners, North Campus Lunches, GYP potluck, ultimate frisbee, The 407, and Nixon House.

I'd like to thank the students that I taught in both ME250 and ME211. I truly enjoyed the chance to get away from my research and interact with you. Your encouragement helped me become a better instructor as well as a better graduate student. I wish you the best in your future studies.

Lastly, thank you to everyone who proofread parts of the thesis: Todd Gordon, Hilary Gordon, Catherine Gordon, David Lipps, Becky Matz, Josh Matz, Carrie Shumaker, Jennifer Steiner, Rich Steiner, Katherine Swart, Chris Wahl, and my writing instructor Dr. Hildinger.

Due to my limited space and even more limited brain capacity I'm sure that I have left people out. Please know that I am deeply appreciative for everything that everyone did to help me out during my graduate school career. While I am looking forward to the next chapter in life, I will also miss this one and the great people that filled it.

This research was supported by the University of Michigan Mechanical Engineering Departmental Fellowship and NIH Biogerontology Training Grant (NIA Training Grant AG000114).

Table of Contents

List of Figures

Figure

List of Tables

Table

List of Appendices

Abstract

Effect of Age, Elbow Muscle Co-contraction Level, and Elbow Moment Loading Characteristics on Elbow Angle Positional Variability in Postural Tasks

by

Mark T. Gordon

Chair: James A. Ashton-Miller

Hand positional variability during a task confounds many activities of daily living, such as inserting a key into a lock, pouring hot water into a cup, and eating with a spoon. Current dogma suggests that older adults use increased co-contraction to attenuate hand positional variability; however, this has not been proven. This thesis proposes that elbow muscle co-contraction significantly affects elbow angle positional variability (PV), which affects hand positional variability, and that older adults can reduce their PV by decreasing their co-contraction level.

Variable muscle stiffness was incorporated into the Signal Dependent Noise Theory to model the effect of elbow muscle co-contraction on predicted PV during a quasistatic elbow flexion task. The results demonstrated an optimal level of co-contraction that minimized the positional variability for both variable and constant loading paradigms.

The first experiment tested the effect of greater than natural elbow muscle co-contraction levels on PV. Fourteen younger and 14 older healthy adults resisted both constant and variable external elbow extension moments while co-contracting at different levels. Increasing co-contraction was found to increase positional variability for both younger and older adults ($p < 0.005$). Older adults naturally co-contracted at higher levels and had lower positional variability than younger adults
($p < 0.005$).

The second experiment tested the effect of below-natural elbow muscle co-contraction levels on PV in 14 younger and 14 older healthy adults. The effect of age was not significant in predicting positional variability at the lowered co-contraction level.

In the third experiment, 12 younger and 14 older healthy adults poured water through different sized container openings. Older subjects co-contracted at higher levels and had lower PV than younger adults. While co-contraction increased with decreasing target size, increased co-contraction within a pouring target size increased PV ($p < 0.05$) and decreased pouring accuracy ($p < 0.001$).

We conclude that increasing elbow muscle co-contraction increases PV and hinders the performance of tasks, such as pouring, that require low positional variability. This counters the current belief that high co-contraction levels are beneficial for task performance, and suggests that people who struggle with these tasks may benefit from reduced co-contraction.

Chapter 1

Introduction

1.1 Motivation

The inability to perform activities of daily living (ADLs), including eating, dressing, and grooming, prevents many older adults from living independently and forces them to seek in-home care, move to assisted living, or move to a nursing home. In a nursing home, elderly residents who can no longer perform one or more of these ADLs require one or more extra hours of assistance per day than those who remain independent (Williams et al., 1994). Not only is this inconvenient for the resident and costly for the family and society (Buttar et al., 2001), but requiring such assistance may have negative consequences for a person's feelings of self-efficacy. Whether buttoning a shirt, pouring hot liquid into a cup, or inserting a key into a lock, hand steadiness is required to accomplish the task without assistance. A central question addressed in this thesis is "what determines the steadiness of the hand during such tasks?"

1.2 Background

1.2.1 Background for Working Hypotheses

An emerging theory in the field of muscle activation prediction is the Signal Dependent Noise (SDN) Theory (Harris and Wolpert, 1998). The theory assumes that the muscle activation patterns used to perform tasks are those that minimize the endpoint positional variability of a limb. This positional variability is defined as the trial-to-trial variability of the hand position at the end of a movement. Positional variability is thought to be the result of cellular and motor noise, which cause fluctuations in muscle force (Jones et al., 2002 and Moritz et al., 2005). These force fluctuations lead to positional variability which confounds many ADLs.

The present research extends SDN Theory from dynamic tasks to include quasistatic tasks. Positional variability *during* the quasistatic task was chosen to replace endpoint positional variability as the outcome metric because we believe the positional variability during a task is more important to the performance of a quasistatic task. The sources and amount of muscle force variability are similar for quasistatic and dynamic tasks; however, the muscle forces resist an external load during a quasistatic task instead of generating a motion.

The conceptual model for the present research is shown in Figure 1.1. An internal model of the task is used to develop the desired angular position, velocity, and acceleration. For a quasistatic task these are the current angular position, zero velocity, and zero acceleration. The differences between the desired and current states are utilized by the Torque Calculator to determine the desired torque. The Muscle Activation Calculator sets the muscle activations to generate the desired torque while minimizing the positional variability of the joint. It also monitors the position of the joint over time to minimize joint positional variability (shown as SD in Figure 1.1). The muscle activation signals are sent to the muscles, but error is introduced from cellular and motor noise. The muscles generate force and stiffness based on the corrupted activation level. These muscle forces and stiffnesses combine with the external load and joint body mechanics to

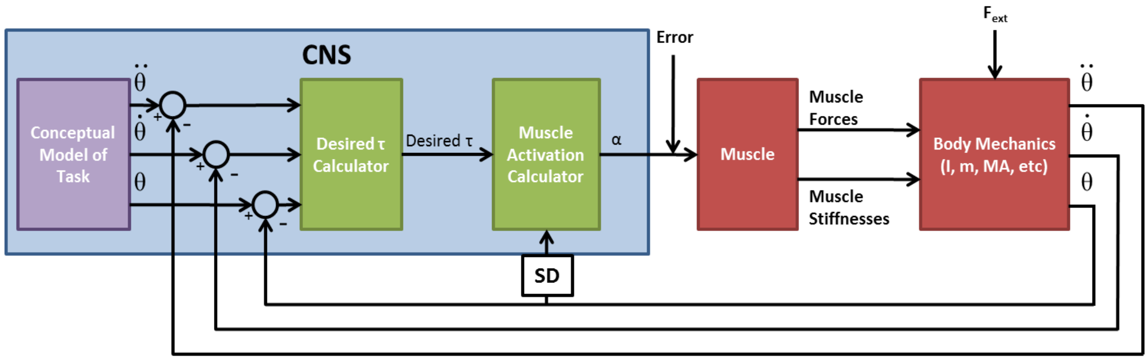


Figure 1.1: Conceptual model incorporating variable muscle stiffness into SDN Theory.

generate the actual joint position, velocity, and acceleration. This model is novel because it combines variable muscle stiffness and the variability of muscle force in the minimization of positional variability during a task.

Deterministic muscle activation prediction schemes often have difficulty predicting muscle co-contraction (which is defined in section 1.2.2.1 *Co-contraction Increases Muscle Activation*, but can be thought of as the simultaneous activation of agonist and antagonist muscles about a joint). Many of the schemes used to predict muscle activation patterns minimize the sum of the muscle stresses to a certain power (e.g., Crowninshield and Brand, 1981) or the maximum muscle intensity (e.g., Hughes et al., 1995). However, these schemes lead to the elimination of co-contraction since co-contraction increases the sum of the muscle stresses or forces about one or more joints (see section 1.2.2.1 Co-contraction Increases Muscle Activation below). More recent stochastic muscle activation prediction theories do predict muscle co-contraction. For example, Monte Carlo techniques have been used in combination with electromyographic (EMG) data to predict co-contraction (e.g., Chang et al., 2000). An alternative approach identifies muscle synergies to predict co-contraction (Torres-Oviedo and Ting, 2007). However, the latter theory lacks a physiological explanation and must develop subject-specific models using EMG data, which makes the theory difficult to validate. Other approaches have considered muscle stiffness in helping to assure postural stability, such as for the spine (e.g., Cholewicki et al., 1997 and McGill et al., 2003). To be successful, SDN Theory must overcome the past difficulties of correctly predicting co-contraction patterns to better predict muscle activation synergies.

For SDN Theory to successfully predict co-contraction patterns, co-contraction must act to minimize positional variability. However, since subjects do not co-contract maximally during most ADLs, SDN Theory suggests that minimal positional variability of the limb is caused by a nonzero, non-maximal co-contraction level (Figure 1.2). This supposition led to our first working hypothesis:

WH-1) *There exists an optimal level of co-contraction that minimizes positional variability during a quasistatic upper extremity task.*

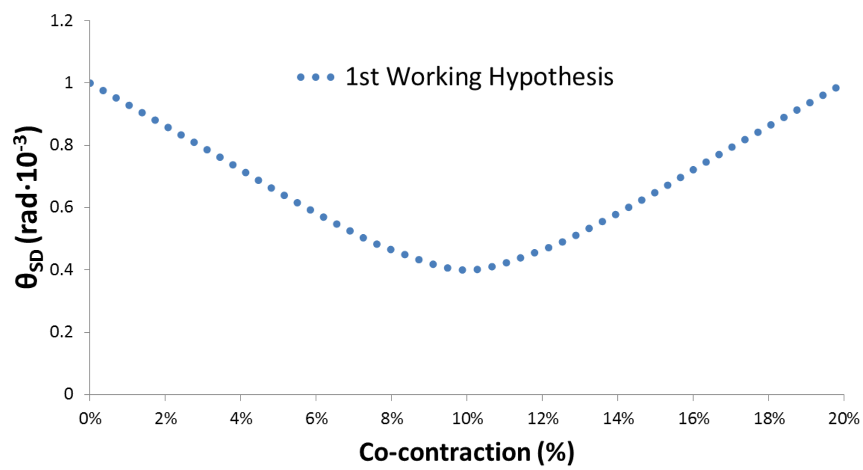


Figure 1.2: Hypothetical positional variability versus co-contraction curve showing a co-contraction level that minimizes positional variability.

Older adults exhibit higher levels of co-contraction than younger adults during dynamic tasks. For example, a review shows increased co-contraction in leg, finger, and elbow muscles with age (Hortobagyi and DeVita, 2006). In addition, Seidler-Dorbin et al.

(1998) studied the elbow extensor muscles during a movement task and found that older adults activated antagonist muscles 30% more than younger adults. However some of the antagonist activation employed during the movement task was used to attenuate angular momentum, which is not necessarily evidence of co-contraction but could simply reflect the triphasic pattern of muscle activation (Cooke and Brown, 1990). The present study addresses this complexity by studying quasistatic tasks.

Seidler-Dorbin et al. (1998) also examined trial-to-trial path variability and created a model to determine the effect of the additional antagonist activation on endpoint path variability. Their model suggests that the additional antagonist activity seen in older adults decreases trial-to-trial path variability. The present research examines the effect of additional co-contraction during quasistatic tasks on positional variability *during* the task rather than *between* trials. This is an important distinction because positional variability *during* a task directly affects the success of the task.

The effect of aging on elbow muscle co-contraction level has also been studied in quasistatic tasks. Valour and Pousson (2003) did not find a significant age difference in co-contraction during an isometric elbow flexion task at five loading levels. However the older adults for this study were recruited from a fitness club and therefore are not representative of all older adults. When sedentary older men were compared to younger men they were found to co-contract 5% more during maximal elbow flexion and extension efforts (Klein et al., 2001). This research expands on these studies by testing the elbow co-contraction levels of healthy older adults during a sub-maximal quasistatic task.

The findings of these previous studies led us to believe that older adults do not co-contract at a level that minimizes their positional variability. The hypothetical positional variability versus elbow muscle co-contraction level curve for younger adults is shown by the blue curve in Figure 1.3. The blue circle represents the “natural” co-contraction level for younger adults that minimizes positional variability. Older adults exhibit increased elbow muscle co-contraction levels (Klein et al., 2001) and increased positional variability (Ranganathan et al., 2001) as indicated by the yellow triangle. The two theories developed in this research are that the curve is shifted for older adults (shown in green), or the curve for older adults is the same as for younger adults but older adults co-contract more than the optimal level. The latter theory suggests that older adults can decrease positional variability by decreasing muscle co-contraction level (moving towards the blue circle). This leads to the second working hypothesis:

WH-2) *Older adults can lower positional variability by decreasing their co-contraction level.*

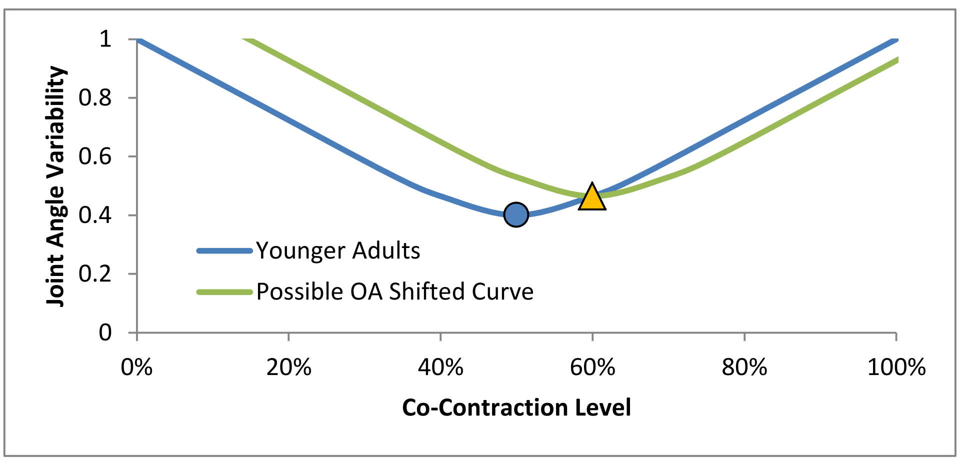


Figure 1.3: Hypothetical results demonstrating the theories for the increase in elbow co-contraction level and positional variability in older adults (yellow triangle): 1) a shifted positional variability versus co-contraction curve for older adults (green) and 2) incorrect co-contraction level on the same curve as younger adults (blue).

1.2.2 Background for Chapter 2

1.2.2.1 Co-contraction Increases Muscle Activation

Consider a 2-D model of the upper and lower arm, including the main elbow muscles, in the parasagittal plane (Figure 1.4). In a weight holding task, the antagonist muscles (shown in red) generate moments in the same direction as the external load. While co-contraction has been defined in many ways, for the purpose of this research, co-contraction is defined as the moment generated by the antagonist muscles divided by the maximal moment the antagonist muscles can generate. Therefore, increased co-contraction increases antagonist muscle activation. The moment generated by antagonist muscle activation must be balanced by increased activation from agonist muscles. Higher co-contraction indicates increased antagonist and agonist muscle activation. Since this research mainly investigates the effect of relative co-contraction levels, most of the findings are applicable for all definitions of co-contraction.

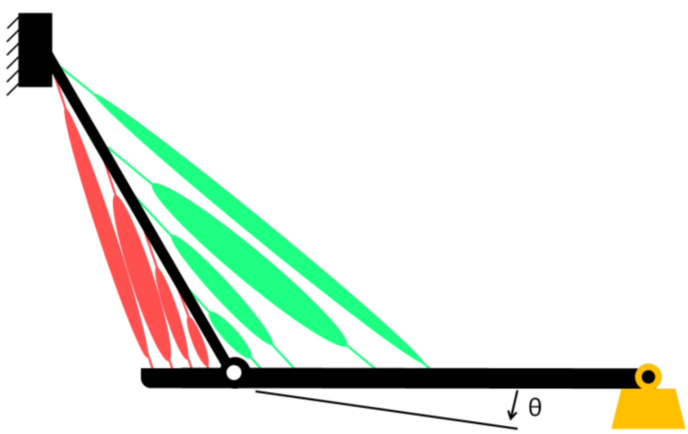


Figure 1.4: Simplified 2-d elbow model in the parasagittal plane. The weight represents an external moment generated about the elbow. The agonist and antagonist muscles are shown in green and red, respectively.

1.2.2.2 Muscle Stiffness Decreases Endpoint Positional Variability

The elbow flexor and extensor muscles, such as those modeled in Figure 1.4, are responsible for the majority of the upper arm stiffness (Perreault et al., 2001). In their review, Faisal and Wolpert (2008) posit that "movement variability decreases overall because the positive stabilizing effect of enhanced stiffness exceeds the negative effects of the increased force variability of the individual muscles." The increasing muscle activation increases muscle and joint stiffness. For example, Sinkjaer et al. (1988) measured the intrinsic stiffness of active muscles about the ankle. When the muscles were activated by electrical stimulation, so as not to activate antagonist muscles, muscle stiffness was found to increase. Other studies have shown that the increased muscle activation caused by co-contraction increased muscle stiffness (Lee and Ashton-Miller, 2011 and Selen et al., 2006).

1.2.2.3 Muscle Force Variability Increases Positional Variability

The force variability described by Faisal and Wolpert (2008) is derived from variation in individual muscle forces (Jones et al., 2002 and Moritz et al., 2005). Muscle force variability causes moment variability and therefore joint angle acceleration. Hamilton et al. (2004) studied the correlation between muscle force and muscle force variability in young adults. The standard deviation of the muscle force increased linearly with the external force suggesting that increased muscle activation leads to increased positional variability. Since co-contraction increases muscle activity, it also increases force variability. The effect of co-contraction on positional variability will be investigated by testing the primary hypothesis:

$H_{2.1}$: *Elbow muscle co-contraction level significantly affects elbow angle variability of healthy younger adults performing a quasistatic task.*

1.2.2.4 Applying SDN Theory to Quasistatic Tasks

Adapting and applying SDN Theory to quasistatic tasks would lead us to predict that co-contraction at lower levels decreases positional variability by increasing stiffness (Figure 1.2). While increasing muscle activation increases force variability, the increase in stiffness may compensate to decrease the endpoint positional variability. This relationship could explain why the natural levels of co-contraction observed in healthy subjects are nonzero. However, according to SDN Theory, co-contraction higher than this natural state must increase positional variability. The increased muscle force variability would cause the positional variability to rise if the additional joint stiffness does not increase enough to compensate for the force variability. This theoretically leads one to expect a co-contraction level that minimizes positional variability, which is investigated by testing the secondary hypothesis:

$H_{2.2}$: *There exists a non-zero, non-maximal elbow co-contraction level that minimizes elbow angle variability in a quasistatic task.*

1.2.3 Background for Chapter 3

1.2.3.1 Loading Paradigms

Natural co-contraction is affected by the elbow loading conditions. For example, Stokes et al. (2000) studied the effect of loading condition on the activation of a subject's elbow muscles. Subjects' upper arms were placed on a table with their elbows flexed at 80°. Their elbows were loaded either by a horizontal or vertical load. For identical elbow moments, muscle activations during the vertical loading were higher than during the horizontal loading. One way that the vertical load is different from the horizontal load is that the vertical load applies an elbow moment that varies with elbow angle. If the elbow moment generated by the subject is less than the external moment, the elbow will extend. This will further increase the moment demand on the elbow. The inverse is true when the subject generates an elbow moment greater than the external load. This will be referred to as the destabilizing loading paradigm because of its propensity to amplify errors. Vertical loading also loads the forearm in axial compression. Axial loading could cause increased muscle activation to balance the axial load or to stabilize the elbow. This research uses a variable horizontal load to produce the destabilizing loading paradigm, mimicking the variable moment created by the vertical loading. This simplifies the system by removing the axial load.

In the absence of the axial load, SDN Theory suggests that the cause of increased muscle activation is to minimize positional variability. One could posit that the effect of force variability should be magnified for vertical loads. The displacement caused by the force variability creates a larger force due to the variable moment loading. The increased effect of force variability may demand greater stiffness about the elbow. This leads to the following hypotheses:

$H_{3.1}$: *Optimal/natural elbow co-contraction is higher for a destabilizing loading paradigm than a stabilizing loading paradigm*.

$H_{3.2}$: *Increasing elbow muscle co-contraction above the natural level will result in higher elbow angle positional variability.*

1.2.4 Background for Chapter 4

As discussed in the background for Chapter 2, the SDN Theory suggests that increasing co-contraction above the natural level will increase positional variability. However, the current belief is that the increased co-contraction seen in older adults (Hortobagyi and DeVita, 2006; Seidler-Dorbin et al., 1998; and Klein et al., 2001) decreases positional variability (Faisal et al., 2008). Chapter 4 addresses this contradiction by testing the primary hypothesis:

$H_{4.1}$: *Increased elbow muscle co-contraction above the natural level will increase elbow angle positional variability in older adults*.

Ranganathan et al. (2001) used a pin test to investigate precision pinch postural steadiness during a task requiring positional control. Subjects were asked to hold a 1 mm pin in a 2 mm diameter hole and the number of times the pin touched the edge of the hole was recorded. Older adults touched the edge six times more often than younger adults, thereby demonstrating increased positional variability of the hand and forearm. Smith et al. (1999) tested the effect of age on hand functionality by recording the time required for younger and older adults to remove a nut from various shaped rods. For subjects under 60 years of age, the time required to remove the nut did not significantly vary with age. However, for adults over 60 years of age the time to remove the nut increased linearly with age. Here we test the positional variability of older adults while resisting an external elbow moment.

As discussed previously, Klein et al. (2001) found that older adults co-contract more than younger adults during maximal isometric contractions. Similar results are expected from the present research, which fills the knowledge gap in the literature left by

Klein et al. not studying sub-maximal tasks. We do so by studying responses to sub-maximal loading and varying loading paradigms:

$H_{4.2}$: *Older adults will exhibit higher natural co-contraction levels and higher positional variability compared to younger adults*.

1.2.5 Background for Chapter 5

Valour and Paulson (2003) did not find any co-contraction difference between fit older adults and younger adults. This suggests that high performing older adults co-contract at the same levels as younger adults. $\mathbf{H_{4.1}}$ and $\mathbf{H_{4.2}}$ state that older adults have higher positional variability and co-contraction than younger adults. $\mathbf{H_{2.1}}$ states that more co-contraction than usual results in higher positional variability. This suggests the source of increased positional variability in older adults may be their increased co-contraction (represented by the yellow triangle on the blue line in Figure 1.3).

However, SDN Theory suggests that the shift in co-contraction seen in older adults is due to a shift in the optimal co-contraction level (represented by the green line in Figure 1.3). This leads us to test the novel hypothesis:

$H_{5.1}$: *Reduced levels of elbow muscle co-contraction increase elbow angle positional variability in younger and older adults for a destabilizing loading paradigm.*

If the hypothesis is supported, then the SDN Theory is applicable to younger and older adults and the source of the shift in co-contraction versus positional variability curves can be investigated. If decreasing the co-contraction level decreases the elbow angle positional variability in both younger and older adults, then SDN Theory is proven to not be applicable for quasistatic tasks. However, if the elbow angle positional variability increases for younger adults and decreases for older adults, this would suggest that aging shifts the natural co-contraction level away from its optimal level. If this is true, occupational therapists can train older adults to reduce their level of co-contraction when performing tasks requiring low positional variability.

1.2.6 Background for Chapter 6

Selen et al. (2006) tested the effect of changing accuracy demands (target sizes) on co-contraction in the elbow. The subject's forearm was constrained to rotate about the elbow and subjects were asked to rotate their elbow to hit a target. The target size was varied to represent accuracy demands. The rotational stiffness of the elbow, tested by an external perturbation, increased with increasing accuracy demand. This suggests that as the target size decreases, elbow co-contraction increases.

Selen et al. (2006) also tested reaching movement trajectory variability to determine how similar the paths were between trials. As the arm approached the target, they found the trial-to-trial variability of the hand trajectory to decrease as target size decreased. However, this is different than the positional variability *during* a task. If co-contraction is increased above normal levels, SDN Theory states that the positional variability will increase. This suggests that subjects do not reduce positional variability with decreasing target size. One possible explanation is that a person's natural response is to increase stiffness when facing a difficult task. While this would decrease the effect of external perturbations, it may not minimize positional variability. This leads to the following primary and secondary hypotheses:

$H_{6.1}$: *Increased elbow muscle co-contraction will result in increased elbow angle positional variability during a task demanding positional accuracy (pouring water).*

$H_{6.2}$: *Elbow angle positional variability and elbow muscle co-contraction will increase as the accuracy demand increases (size of the container opening decreases)*.

Klein et al. (2001) found increased co-contraction for older adults during maximal isometric voluntary contractions compared to younger adults. Similar results are expected during a quasistatic pouring task utilizing both the elbow and wrist muscles. As discussed above, Ranganathan et al. (2001) found that older adults have higher positional variability during a pin task than younger adults. The pin task did not require the elbow to resist an external load. While the subject's arm is loaded with a teapot in this experiment, similar results are expected:

$H_{6.3}$: *Elbow muscle co-contraction and elbow angle positional variability will be higher for older adults than younger adults for an accuracy demanding task regardless of the accuracy level demanded*.

1.3 References

Buttar, A., Blaum, C., & Fries, B. (2001). Clinical characteristics and six-month outcomes of nursing home residents with low activities of daily living dependency. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 56(5), M292-M297.

Chang, Y., Hughes, R. E., Su, F., Itoi, E., & An, K. (2000). Prediction of muscle force involved in shoulder internal rotation. Journal of Shoulder and Elbow Surgery, 9(3), 188-195.

Cholewicki, J., Panjabi, M. M., & Khachatryan, A. (1997). Stabilizing function of trunk flexor-extensor muscles around a neutral spine posture. Spine, 22(19), 2207-2212.

Cooke, J. D., & Brown, S. H. (1990). Movement-related phasic muscle activation. II. generation and functional role of the triphasic pattern. Journal of Neurophysiology, 63(3), 465-472.

Crowninshield, R. D., & Brand, R. A. (1981). A physiologically based criterion of muscle force prediction in locomotion. Journal of Biomechanics, 14(11), 793-801.

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Hamilton, A. F., Jones, K. E., & Wolpert, D. M. (2004). The scaling of motor noise with muscle strength and motor unit number in humans. Experimental Brain Research., 157(4), 417-430.

Harris, C. M., & Wolpert, D. M. (1998). Signal-dependent noise determines motor planning. Nature, 394(6695), 780-784.

Hortobagyi, T., & Devita, P. (2006). Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. Exercise and Sport Sciences Reviews, 34(1), 29-35.

Hughes, R. E., & Chaffin, D. B. (1995). The effect of strict muscle stress limits on abdominal muscle force predictions for combined torsion and extension loadings. Journal of Biomechanics, 28(5), 527-533.

Jones, K. E., Hamilton, A. F. C., & Wolpert, D. M. (2002). Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology, 88(3), 1533.

Klein, C. S., Rice, C. L., & Marsh, G. D. (2001). Normalized force, activation, and coactivation in the arm muscles of young and old men. Journal of Applied Physiology, 91(3), 1341-1349.

Lee, Y., & Ashton-Miller, J. (2011). The effects of gender, level of co-contraction, and initial angle on elbow extensor muscle stiffness and damping under a step increase in elbow flexion moment. Annals of Biomedical Engineering, 39(10), 2542-2549.

McGill, S. M., Grenier, S., Kavcic, N., & Cholewicki, J. (2003). Coordination of muscle activity to assure stability of the lumbar spine. Journal of Electromyography and Kinesiology, 13(4), 353-359.

Moritz, C. T., Barry, B. K., Pascoe, M. A., & Enoka, R. M. (2005). Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle. Journal of Neurophysiology, 93(5), 2449-2459.

Perreault, E. J., Kirsch, R. F., & Crago, P. E. (2001). Effects of voluntary force generation on the elastic components of endpoint stiffness. Experimental Brain Research, 141(3), 312-323.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. (2001). Effects of aging on hand function. Journal of the American Geriatrics Society, 49(11), 1478-1484.

Seidler-Dobrin, R. D., He, J., & Stelmach, G. E. (1998). Coactivation to reduce variability in the elderly. Motor Control, 2(4), 314-330.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieën, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Sinkjaer, T., Toft, E., Andreassen, S., & Hornemann, B. C. (1988). Muscle stiffness in human ankle dorsiflexors: Intrinsic and reflex components. Journal of Neurophysiology, 60(3), 1110-1121.

Smith, C. D., Umberger, G. H., Manning, E. L., Slevin, J. T., Wekstein, D. R., Schmitt, F. A., et al. (1999). Critical decline in fine motor hand movements in human aging. Neurology, 53(7), 1458-1461.

Stokes, I. A. F., & Gardner-Morse, M. G. (2000). Strategies used to stabilize the elbow joint challenged by inverted pendulum loading. Journal of Biomechanics, 33(6), 737-743.

Torres-Oviedo, G., & Ting, L. H. (2007). Muscle synergies characterizing human postural responses. Journal of Neurophysiology, 98(4), 2144-2156.

Valour, D., & Pousson, M. (2003). Compliance changes of the series elastic component of elbow flexor muscles with age in humans. Pflugers Archiv : European Journal of Physiology, 445(6), 721-727.

Williams, B. C., Fries, B. E., Foley, W. J., Schneider, D., & Gavazzi, M. (1994). Activities of daily living and costs in nursing homes. Health Care Financing Review, 15(4), 117.

Chapter 2

Effect of Co-contraction on Elbow Angle Positional Variability during a Quasistatic Task: A Theoretical and Experimental Study

2.1 Abstract

Upper extremity steadiness, the inverse of positional variability, is important for performing many activities of daily living such as eating with a spoon, inserting a key into a lock, and pouring hot water into a cup. The effect of elbow muscle co-contraction on upper extremity steadiness during a quasistatic task is unknown. This knowledge gap has been addressed in this chapter by augmenting the Signal Dependent Noise Theory with variable muscle stiffness. This augmented theory was used to produce a simulation to test the primary hypothesis that elbow muscle co-contraction level significantly affects elbow angle variability. The simulation included the physiological properties of four agonist and four antagonist elbow muscles in a planar elbow joint model. To validate the simulation, 10 young healthy adults were asked to resist external moments requiring 20%, 40%, and 60% of their maximal voluntary elbow flexion efforts while co-contracting their elbow muscles at three levels from natural to maximal co-contraction. The simulation results demonstrated that co-contraction level affects elbow angle variability, and a co-contraction level was found that minimizes elbow angle variability. The experimental results largely supported the primary hypothesis by demonstrating significantly increased elbow angle variability with increased elbow muscle co-contraction above the natural co-contraction level ($p < 0.001$). We conclude that healthy

younger adults cannot decrease elbow angle positional variability by increasing elbow muscle co-contraction above natural levels.

2.2 Introduction

Upper extremity steadiness is important for activities of daily living (ADLs) requiring precise positional control. Examples of such tasks include pouring a cup of hot water without spilling or inserting a key into a lock. A recognized source of unsteadiness is the stochastic noise on each muscle's contractile force (Jones et al., 2002 and Moritz et al., 2005). However, the effect of muscle co-contraction, the co-activation of agonist muscles with antagonist muscles, on steadiness is not fully understood. Increased co-contraction is associated with lower trial-to-trial variability of the position at the end of a movement for individual subjects (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006). However, no association was found between co-contraction and acceleration variability between subjects (Burnett et al., 2000 and Laidlaw et al., 2002). The effect of co-contraction on an individual subject's positional variability *during* a single task has not been studied.

Emerging theories in motor planning (Harris and Wolpert, 1998) and muscle force prediction (Haruno and Wolpert, 2005) assume that the governing principle of muscle activation is the minimization of the trial-to-trial variability of the hand position at the end of a movement. This positional variability is thought to result from variability in the force generated by each muscle (Jones et al., 2002 and Moritz et al., 2005), giving rise to the Signal Dependent Noise (SDN) Theory (Haruno and Wolpert). The model proposed by Haruno and Wolpert does not predict co-contraction for a quasistatic task because it uses pure force actuators which do not increase joint stiffness with increasing muscle activation.

The goal of the present research, therefore, was to modify the Signal Dependent Noise Theory to be applicable to quasistatic tasks. Then, we wanted to determine whether co-contraction steadies the elbow or if it is a source of additional joint angle variability. Therefore, we tested the primary hypothesis, stated in Chapter 1 as $H_{2.1}$: *Elbow muscle co-contraction level significantly affects elbow angle variability of healthy younger adults performing a quasistatic task.*

Co-contraction increases muscle activation, which is known to increase muscle tensile stiffness and damping (e.g., Sinkjaer et al., 1988; Blanpied and Smidt, 1993; and Lee and Ashton-Miller, 2011). Therefore, it seems reasonable to expect an inverse relationship between co-contraction and elbow angle variability. However, the increased muscle activation level also causes greater muscle force variability (Graves et al., 2000), which suggests an increase in elbow angle variability leading to a positive relationship between co-contraction and elbow angle variability. It is unclear which of these factors is dominant in influencing the co-contraction versus elbow angle variability relationship. Furthermore, it is possible that the dominant factor at low co-contraction levels is the increased stiffness, while at higher co-contraction levels the muscle force variability is dominant. Therefore, we tested a secondary hypothesis, $H_{2.2}$: *There exists a non-zero, non-maximal elbow co-contraction level that minimizes elbow angle variability in a quasistatic task.* If the relationship between co-contraction and positional variability was better understood, occupational therapists could use this knowledge to assist patients in performing ADLs that require low positional variability.

2.3 Methods

2.3.1 Simulation

A planar model of the elbow joint (Figure 2.1) was developed to test the hypotheses. The shoulder joint was considered fixed and the elbow joint was modeled as a frictionless pin joint. The forearm was modeled as a single rigid body attached to the elbow joint. Eight muscles were included in the model: the four major agonist muscles that flex the elbow joint (long head biceps, short head biceps, brachialis, and brachioradialis) as well as the four major antagonist muscles that extend the elbow joint (long head triceps, medial head triceps, lateral head triceps, and anconeous). An external moment, represented as an external weight in Figure 2.1, was applied to the elbow. The representation of the muscles were simplified by assuming they only exerted moments in flexion or extension: any pronation and supination moments they generated were assumed to be equilibrated by other arm muscles that would not greatly affect the flexion and extension moments. While the simplifications of a planar model may result in

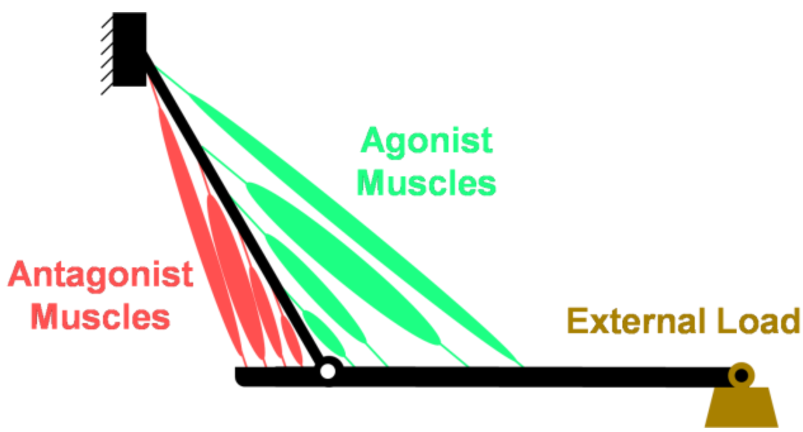


Figure 2.1: Simulation model shown with four agonist muscles and four antagonist muscles with different moment arms and cross sectional areas.

inaccuracies, they allow for a simple definition of elbow angle, agonist muscles, and antagonist muscles.

Physiological cross-sectional area, moment arms, and force length curves of the eight muscles were taken from the literature (Holzbaur et al., 2005 and Murray et al., 2000). The muscle force variability for each muscle was calculated as a function of the muscle's maximum voluntary torque and activation level as described below (Hamilton et al., 2004).

Elbow flexor and extensor muscles are responsible for the majority of the stiffness in the upper arm (Perreault et al., 2001). While most studies fit a linear relationship between activation and active stiffness, the data are unconvincing for low activation levels (Sinkjaer et al., 1988). Sinkjaer et al. state that the passive tension in the study was negligible, yet the intrinsic stiffness regression line has a non-zero stiffness intercept. The measurements at the lowest torque levels (below approximately 2.5 Nm) all fall below the stiffness regression line suggesting a possible downward curve. The data from Blanpied and Smidt (1993) also exhibit a non-zero stiffness intercept for their stiffness curve despite subtracting out the passive stiffness after measuring the stiffness at rest. This suggests a non-linear relationship exists between stiffness and torque for muscle activation below 15%.

Therefore, the muscle tensile stiffness for each muscle was estimated by fitting a sixth order polynomial to the muscle stiffness and activation data from Sinkjaer et al. (1988) resulting in $MuscleStiffness = -25.8 \cdot x^6 + 100.4 \cdot x^5 - 153.2 \cdot x^4 + 115.2 \cdot x^3 - 44.2 \cdot x^2 + 9.4 \cdot x$ where x is the muscle activation percentage. The muscle stiffness was then scaled by the cross sectional area of the muscle. The damping property of muscle was not included in this model due to the quasistatic nature of the research.

2.3.1.1 Moment Variability

Moment variability (σ_M) can be calculated from the muscle properties force coefficient of variation (CV), physiological cross sectional area (PCSA), specific force (SF), moment arm length (MA), and muscle activation level (α). Graves et al. (2000) found the force variability of a muscle to be determined by a constant coefficient of variation for younger adults. This force variability (σ_F) is translated into moment variability through the moment arm of the muscle (equations 1-5). In the equations below, Force is the mean force produced by the muscle, and F_{max} is the maximal force the muscle can produce. This moment variability is necessary to determine the positional variability the muscle will cause in the attached bone.

$$\sigma_M = MA \cdot \sigma_F \qquad (1)$$

$$\sigma_F = CV \cdot Force \qquad (2)$$

$$Force = F_{max} \cdot \alpha \qquad (3)$$

$$F_{max} = PCSA \cdot SF \qquad (4)$$

$$\sigma_M = CV \cdot PCSA \cdot SF \cdot MA \cdot \alpha \quad (5)$$

Using equation 5 we can calculate the moment variability of a system of muscles assuming that the muscle variabilities are independent from each other (Appendix 1). When adding the variability of *n* redundant muscles, equation 5 yields

$$\sigma_{M,Total} = \sqrt{\sum_{i=1}^{n} \sigma_{M,i}{}^2} = \sqrt{\sum_{i=1}^{n} (CV_i \cdot PCSA_i \cdot SF_i \cdot MA_i \cdot \alpha_i)^2} \qquad (6)$$

The moment variability of the muscle system can then be used to calculate the positional variability as shown in 2.3.1.3.

2.3.1.2 Co-activation

Co-activation represents a force sharing between redundant muscles. In this paper, co-activation is defined as the activation of two or more muscles that generate moments in the same direction. We will use the 1-joint 2-dimensional model in the parasagittal plane from Figure 2.1 to simplify the system. The moment variability produced by co-activation of two agonist muscles is:

$$\sigma_{M,Total} = \sqrt{\sum_{i=1}^{n} \sigma_{M1}^{2} + \sigma_{M2}^{2}} \qquad (7)$$

These moments can be used to balance an external moment on the elbow joint in a quasistatic task.

$$M_{ext} = M_1 + M_2 = F_{Max,1} \cdot \alpha_1 \cdot MA_1 + F_{Max,2} \cdot \alpha_2 \cdot MA_2 \quad (8)$$

Force variability is determined by the coefficient of variation of the muscle and therefore scales with the force and the muscle activation (Graves, et al., 2000).

$$\sigma_{M,Total} = \sqrt{(CV_1 \cdot PCSA_1 \cdot SF_1 \cdot MA_1 \cdot \alpha_1)^2 + (CV_2 \cdot PCSA_2 \cdot SF_2 \cdot MA_2 \cdot \alpha_2)^2} \qquad (9)$$

Note that in equation 9 the only unknown variables for a physiologically known system are the muscle activations (α). Solving for α_2 in equation 8 and substituting into equation 9 yields the relationship between variability and the activation of muscle 1.

$$\sigma_{M,Total} =$$

$$\sqrt{[(CV_1 \cdot PCSA_1 \cdot SF_1 \cdot MA_1)^2 + (CV_2 \cdot PCSA_1 \cdot SF_1 \cdot MA_1)^2]\alpha_1^2 - 2 \cdot CV_2^{\,2} \cdot M_{ext} \cdot PCSA_1 \cdot SF_1 \cdot MA_1 \cdot \alpha_1 + CV_2^{\,2} \cdot M_{ext}^{\,2}} \quad (10)$$

$$= \sqrt{[(CV_1 \cdot M_{Max,1})^2 + (CV_2 \cdot M_{Max,1})^2]\alpha_1^2 - 2 \cdot CV_2^{\,2} \cdot M_{ext} \cdot M_{Max,1} * \alpha_1 + CV_2^{\,2} \cdot M_{ext}^{\,2}} \qquad (11)$$

Taking the derivative with respect to α_1 and setting it equal to 0 solves for the activation that produces the smallest amount of moment variability (equation 12). In the case of two identical muscles where $CV_1 = CV_2$ and, equation 13 results in each muscle taking half of the external load.

$$0 = [(\mathrm{CV}_1 \cdot \mathrm{M}_{\mathrm{Max},1})^2 + (\mathrm{CV}_2 \cdot \mathrm{M}_{\mathrm{Max},1})^2]\alpha_1 - \mathrm{CV}_2{}^2 \cdot M_{ext} \cdot \mathrm{M}_{\mathrm{Max},1} \quad (12)$$

$$\alpha_1 = \frac{\mathrm{CV}_2{}^2 \cdot M_{ext}}{(\mathrm{CV}_1{}^2 + \mathrm{CV}_2{}^2) \cdot \mathrm{M}_{\mathrm{Max},1}} \quad (13)$$

$$\alpha_1 = \frac{\mathrm{CV}_2{}^2 \cdot M_{ext}}{(\mathrm{CV}_1{}^2 + \mathrm{CV}_2{}^2) \cdot \mathrm{PCSA}_1 \cdot \mathrm{SF}_1 \cdot \mathrm{MA}_1} \quad (14)$$

These equations demonstrate which muscle properties affect the optimal activation. Equation 14 suggests that the coefficient of variation is the only property of muscle 2 that affects the optimal activation of muscle 1. The right hand side of Equation 14 shows that the maximal moment producible by muscle 1 and the coefficient of variation on each muscle determines the optimal activation level of muscle 1.

The muscle force coefficient of variation is inversely proportional to the maximal moment the muscle can produce (Hamilton et al., 2004). Minimizing moment variability therefore favors higher loads in muscles that are larger and/or have larger moment arms over muscles that are smaller and/or have smaller moment arms. Increased moment variability produces increased positional variability and therefore should be minimized according to the SDN Theory. Therefore, the SDN theory should predict muscles that can produce higher moments to carry higher loads than weaker muscles.

2.3.1.3 Co-contraction

Co-contraction is the activation of antagonist muscles in combination with agonist muscles. In this simple model, antagonist muscles extend the elbow joint while agonist muscles flex the joint. Co-contraction introduces another form of redundancy to the system. Not only do agonist muscles share the external load, but the moment generated by antagonist muscles must be balanced by increasing the activation in the agonist muscles. Since moment variability increases with muscle activation (equation 5), additional agonist and antagonist muscle activation caused by co-contraction increases the moment variability of the joint. This yields a model that favors low activation and therefore no co-contraction to minimize the moment variability of the joint.

However, most tasks demand decreased positional variability rather than decreased moment variability. One way to couple moment and positional variability is to include muscle stiffness. While muscle stiffness is typically ignored or held constant, it has great importance to positional variability. Higher joint stiffness decreases the moment variability's effect on positional variability. To approximate the relative variability seen in a quasistatic task, we divided the joint moment variability by the joint rotational stiffness generated by the active muscles. This allowed us to solve for the muscle activations that generate the least positional variability. Equations 15 and 16 show the objective function for one agonist and one antagonist muscle where c = [9.4, −44.2, 115.2, −153.2, 100.4, −25.8].

$$\sigma^2_{\mathrm{Position,Total}} = \frac{{\sigma_{\mathrm{M,Agon}}}^2 + {\sigma_{\mathrm{M,Antag}}}^2}{k_{\mathrm{Agon}} + k_{\mathrm{Antag}}} \quad (15)$$

$$= \frac{{\sigma_{\mathrm{M,Agon}}}^2 + {\sigma_{\mathrm{M,Antag}}}^2}{\left(\sum_{n=1}^{6} c_n \cdot \alpha_{\mathrm{Agon}}\right) \cdot \mathrm{MA}_{\mathrm{Agon}} \cdot \mathrm{PCSA}_{\mathrm{Agon}} + \left(\sum_{n=1}^{6} c_n \cdot \alpha_{\mathrm{Antag}}\right) \cdot \mathrm{MA}_{\mathrm{Antag}} \cdot \mathrm{PCSA}_{\mathrm{Antag}}} \quad (16)$$

While this does not yield the actual positional variability, it does take the correct form. Equations 15 and 16 increase positional variability with increasing moment variability and decrease positional variability with increasing muscle stiffness. Equation 16 allows us to approximate the optimal co-contraction level without knowing the actual positional variability. We also verified this approximation using a dynamic Monte Carlo simulation (see section 2.3.1.4) which supports this methodology. This simplification allowed us to investigate more complex muscle arrangements such as those about the elbow.

2.3.1.4 Monte Carlo Simulation

The Monte Carlo simulation was used to support the use of a static solution to a quasistatic situation as discussed at the end of section 2.3.1.3. The physical properties of the forearm such as mass, length, and inertia of the model were that of a 50th percentile male. The muscle moment variations were generated using the CV as calculated in Equation 5. The muscle moments were generated using the corresponding CV and were added together to generate a net joint torque. This torque was then used with the moment of inertia, joint stiffness, and external force to calculate the dynamic movement of the arm. The simulation was run for 1000 seconds with muscle forces being generated at 30 Hz. The data were filtered at 0.5 Hz to reduce the effect of positional drift.

The Monte Carlo simulation was not used to determine the optimal activation because the size of the system led to high computation time. While we were able to simulate the positional variability for a given set of muscle activations, solving for the optimal activations themselves using the Monte Carlo simulation would be difficult due to computation time.

2.3.2 Experimental Methods

2.3.2.1 Subjects

Ten healthy right-handed younger adults (6 male and 4 female) volunteered to take part in this institutional review board approved experiment and signed a written statement of informed consent. The subjects were between the ages of 24 and 29 years, right hand dominant, and screened for the absence of arm, shoulder, or spine injuries. They were also screened to be free of any neurological disabilities or chronic musculoskeletal conditions.

2.3.2.2 Kinematic Data

All subjects were instrumented with triads of infrared markers to track the kinematics of the forearm and upper arm. The markers were operated at 1 kHz and a single Certus Optotrak camera system (Northern Digital Inc., Waterloo, Canada) was used to collect the kinematic data. Since subjects increase their co-contraction level when concentrating on positional accuracy (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006), subjects were instructed to not focus on maintaining the starting position. However, this led to positional drift during the trial which would mask the positional variability during the task. The positional drift was removed by filtering the data in Matlab through a third order high pass filter with a cutoff frequency of 4 Hz. Kinematic data were collected at 1 kHz on a central computer.

2.3.2.3 Force Measurement

A wrist cuff was placed around the subject's wrist to transmit the force from a cable to the subject's arm. The forearm was allowed to rotate freely inside the wrist cuff;

this prevented the wrist cuff from exerting a supination or pronation moment on the forearm. A custom strain gauge was attached between the cable and the wrist cuff to measure the force on the wrist. Force data were collected at 1 kHz on a central computer.

2.3.2.4 EMG Data

The electromyographic (EMG) activation levels for the aforementioned eight muscles were measured using a Myosystem 2000 system (Noraxon U.S.A. Inc., Scottsdale, Arizona) with blue dot surface electrode spacing of 2 cm aligned along the muscle fibers. The EMG data were collected at 1 kHz on a central computer using a 12 bit analog-to-digital converter.

2.3.2.5 Maximal Force Trial

The subject's upper arm and elbow were placed on a horizontal surface at shoulder height (Figure 2.2). The cable from the wrist cuff was attached to a static structure mounted on the wall. The subject was asked to flex his or her elbow to 80° and pull against the wrist cuff. Next, the subject was instructed to increase his or her force over three seconds, hold their maximal force for two seconds, and then allow the force to decrease over two seconds. The maximal force was taken to be the highest average over a 0.25-second period. This test was repeated up to five times until two measurements were within 5% of each other. The same test was repeated for the elbow extensor muscles. No subject repeated either trial more than four times.

2.3.2.6 Trials

During each trial the wrist cuff cable was attached to a ¾-inch diameter air cylinder (Bimba Manufacturing, University Park, Illinois) that exerted elbow extension loads of 20%, 40%, or 60% of the subject’s maximal effort. An air cylinder was used to eliminate the effect of varying inertia between loading levels. For the subject’s safety, a regulator limited the maximal air supply pressure to 40 psi.

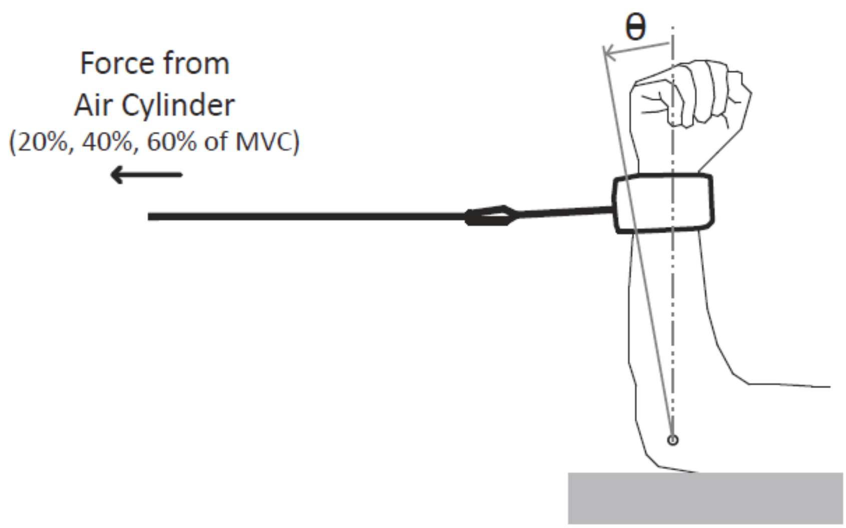


Figure 2.2: Diagram of the experimental setup showing the orientation of the subject’s arm and the measure of the elbow angle used to calculate the elbow angle positional variability.

Next, subjects were asked to co-contract at one of three specified muscle activation levels. The muscle co-contraction levels were designated as ‘natural’, ‘medium’, and ‘maximal’. Natural was defined as the level of co-contraction that the subject naturally uses when performing the task. Maximal was the maximal co-contraction the subject could produce while performing the task. To learn how to co-contract at a medium level, subjects were provided with visual biofeedback of their root-mean-square (RMS) triceps muscle EMG at the beginning of the session. Subjects practiced generating co-contraction that was approximately 50% between natural and maximal. This visual feedback was removed for the actual trials so the subjects would not fixate on controlling the muscle activation level. The co-contraction levels were randomized during the trials.

Subjects were instructed that they need not focus on maintaining the starting position. Before each trial, the subjects were told the co-contraction level for the trial. The subject then flexed their elbow approximately 10° and co-contracted at the specified level for five seconds. Kinematic, force, and EMG data were recorded during these trials. Each subject was asked to perform the trials at the three muscle co-activation levels and the three loading levels. Each force/co-contraction combination was repeated three times. Subjects were told to release the load any time they felt uncomfortable.

2.3.2.7 Statistical Analysis

MiniTab (State College, Pennsylvania) was used to develop a general linear model of the data. The model blocked on subjects to eliminate subject-dependent factors such as height and strength. This resulted in a model of the form $Variability = \beta_0 + \beta_1 x_1 + \beta_2 x_2$ where β_0 is a constant, $\beta_1$is the coefficient for force, x_1 is the force, β_2 is the coefficient for co-contraction, and x_2 is the co-contraction level. A p-value of 0.05 or less was considered significant in testing the hypotheses.

2.4 Results

2.4.1 Simulation Results

Elbow muscle co-contraction level was found to significantly affect elbow angle positional variability (θ_{SD}) thereby supporting the primary hypothesis. An elbow muscle co-contraction level was found that minimized elbow angle variability (marked with ★ in Figure 2.3) supporting the secondary hypothesis. Elbow angle variability increased monotonically from the optimal co-contraction level for both higher and lower co-contraction levels. Higher external moments were found to lead to higher levels of elbow angle positional variability and increasing optimal co-contraction levels (Figure 2.3). The co-contraction corresponding to minimal joint angle variability was verified with the Monte Carlo simulation (Table 2.1). The joint angle variability magnitudes from the two simulations were normalized for this comparison.

Table 2.1: Normalized positional variability results from both the Monte Carlo Simulation and the simulation using the static approximation. These results indicate that the simulation using the static approximation predicts the same co-contraction level that minimizes positional variability as the Monte Carlo simulation.

	Normalized Positional Variability Results Generated with a 5 Nm External Moment	
Co-contraction	Simulation	Monte Carlo Simulation
0.5%	0.843	0.811
1%	0.807	0.800
2%	0.766	0.762
3%	0.750	0.747
4%	0.749	0.751
5%	0.757	0.780
6%	0.772	0.781
7%	0.791	0.806
8%	0.813	0.842
9%	0.838	0.857
10%	0.864	0.878
11%	0.891	0.879
12%	0.918	0.935
13%	0.946	0.992
14%	0.973	0.993
15%	1.000	1.000

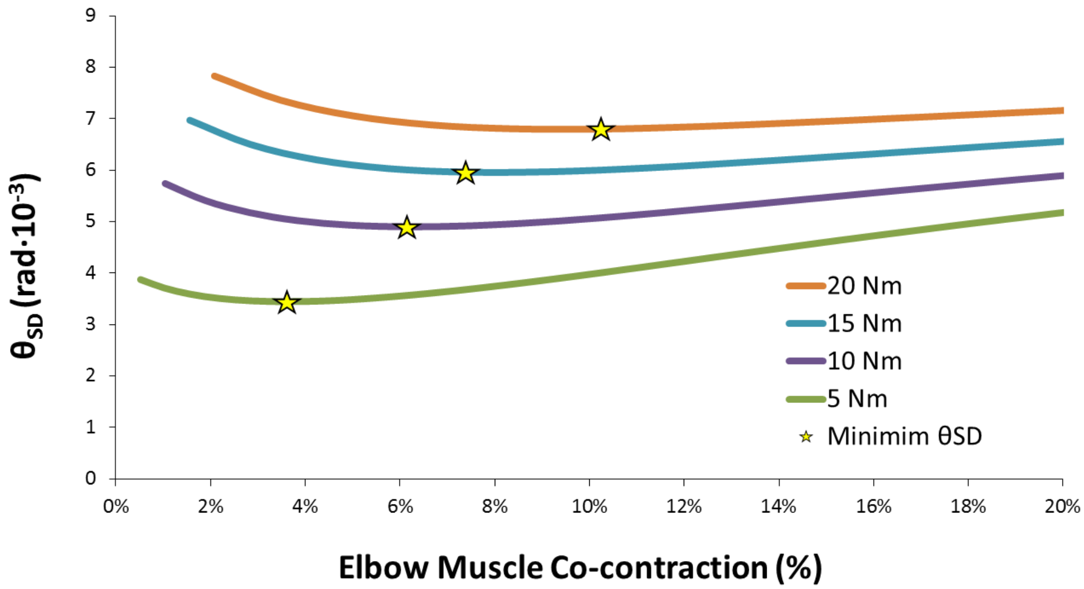


Figure 2.3: Simulation results showing elbow angle positional variability is strongly affected by both elbow muscle co-contraction level and external loading level. The co-contraction level yielding the lowest positional variability for each external load is represented by ★.

2.4.2 Experimental Results

The experiment investigated the effect of co-contraction levels from the ‘normal’ level (approximately 5-10%) to ‘maximal’ co-contraction. Over this range, nine out of the ten subjects showed a positive correlation between co-contraction level and elbow angle variability while the other subject showed a weak negative correlation (Figure 2.4).

Trial data grouped by verbally instructed co-contraction level show that, while the co-contraction level varied between subjects, elbow angle variability did increase with co-contraction level (Figure 2.5). At the 20% MVC level, positional variability increased approximately three-fold between natural and maximal co-contraction. Statistical analysis accounting for subject differences found that elbow angle variability was correlated with both co-contraction level ($p < 0.001$) and force ($p < 0.05$, Table 2.2). This result supports the primary hypothesis and corroborates the simulation results (Figure 2.3). The effect of fatigue and the interaction of co-contraction and force were not found to be significant.

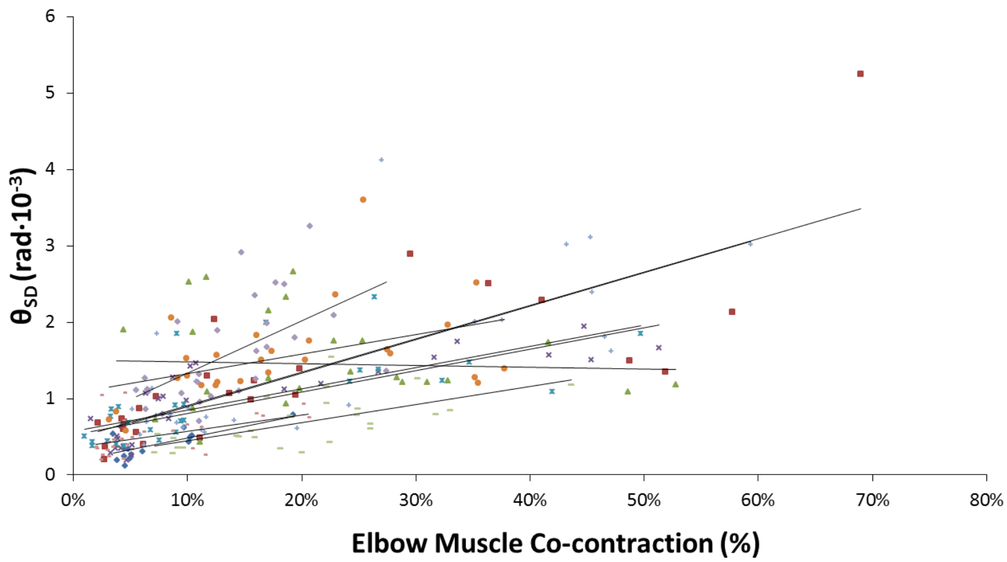


Figure 2.4: Scatter plot showing experimental data from 10 healthy young subjects. The linear regression line for each subject indicates increased elbow positional variability with increased co-contraction in 9 of the 10 subjects.

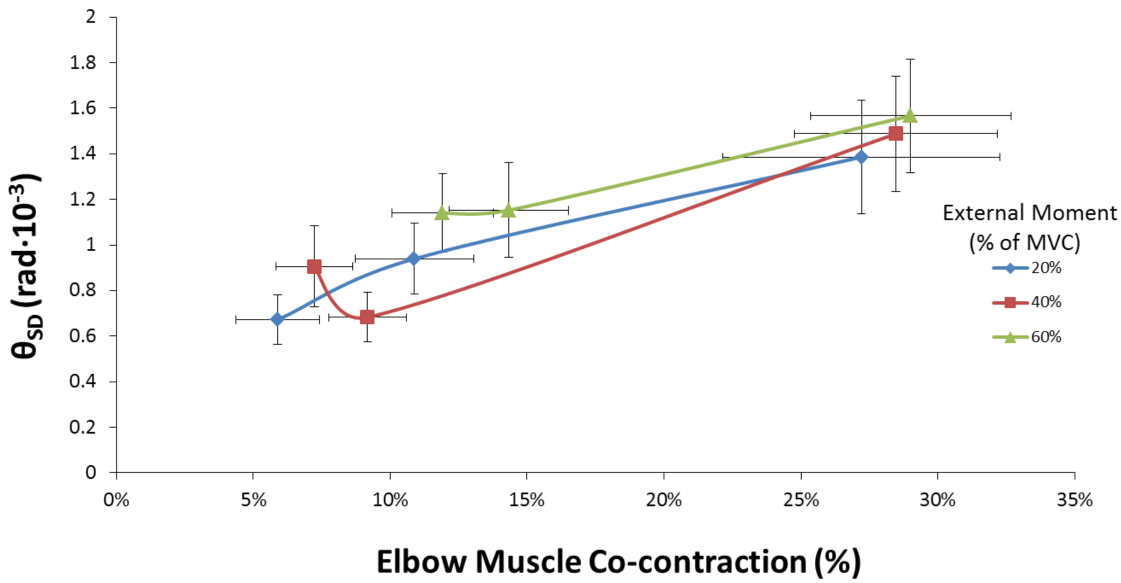


Figure 2.5: Experimental data for the three different instructed elbow muscle co-contraction levels (natural, medium, and maximal) shown the grouped mean and standard error bars. Elbow muscle co-contraction level significantly raised elbow angle positional variability ($p < 0.005$) supporting the findings of the simulation (Figure 2.3) for above natural co-contraction levels.

Table 2.2: General linear model for elbow angle positional variability suggests that force and co-contraction effects were significant ($n = 10$, $p < 0.005$).

	Coefficient	p-value
Constant	0.026	<0.001
Force	0.022	0.025
Co-contraction	0.18	<0.001

2.5 Discussion

Elbow muscle co-contraction level was found to affect elbow angle variability during this quasistatic elbow flexion task, supporting the primary hypothesis. Both the simulation and the experimental results indicate that high co-contraction levels increased elbow angle variability.

The simulation predicted an optimal value of co-contraction that minimizes elbow angle positional variability. Increasing or decreasing the co-contraction level from the optimal level monotonically increased positional variability. This supports the second hypothesis. This may have implications in rehabilitation aimed at improving steadiness if this finding is corroborated by others.

The simulation corroborates previous simulations for low co-contraction levels (Selen et al., 2005) by predicting decreasing position variability with increasing co-contraction. However, the simulation presented here disagrees with the simulation results from Selen et al. for high levels of co-contraction. That model continues predicting lower positional variability for higher levels of co-contraction. While that model agrees with the current dogma which assumes that co-contraction at any level decreases positional variability (Faisal et al., 2008 and Selen et al., 2005), both the simulation and experimental results presented in this study contradict that dogma. This could be because Selen's experiments did not control co-contraction levels, but rather examined natural activation levels associated with accuracy constraints. Also the positional variability in Selen's experiment was measured as the trial-to-trial variability in the final position of the arm. The present study tested the effect of increasing co-contraction in individual subjects on the variability *during* the trial; this allows us to predict the effect that increasing a subject's co-contraction level will have on their elbow angle variability.

One novel feature of the simulation is that muscle stiffness was assumed to vary with muscle activation. We believe this is the first time that variable muscle stiffness has been implemented in SDN theory. This allows for a better estimate of the variability in elbow angle, as well as a means of predicting co-contraction levels required for quasistatic tasks. We believe this is also the first application of SDN theory to a quasistatic task and the first to show the effect of co-contraction level on elbow angle positional variability.

While most studies fit a linear relationship between muscle activation and active muscle stiffness, the data is unconvincing for low activation levels (Sinkjaer et al., 1988 and Blanpied and Smidt, 1993) as discussed in section 2.3.1. If muscle stiffness changed linearly with muscle activation, the antagonist activation in this study would either always increase or decrease the positional variability. This would lead the optimal activation of the antagonist muscles to be either inactivated or maximally activated. Approximating the stiffness as a non-linear function allows the simulation to predict co-contraction which is naturally seen during tasks.

One limitation of the study is that we could not determine the optimal level of co-contraction experimentally. We were only able to investigate co-contraction levels equal to and higher than the subject's normal co-contraction level. However, the co-contraction levels observed when subjects naturally co-contract were in the same range as the optimal co-contraction predicted by the simulation.

A second limitation is the planar nature of the model. Some of the muscle activity observed experimentally was likely required to equilibrate out-of-plane moments. Furthermore, the simulation only investigated the effect of the elbow muscles on the elbow joint even though many of the elbow muscles also cross the shoulder. The shoulder moments created by the biarticular muscles were assumed to be balanced by other muscle groups that would not significantly impact the results. By expanding the model to three dimensions and multiple joints one should be able to learn more about the relationship between co-contraction and elbow angle variability.

Some older adults struggle to perform everyday tasks requiring fine motor skills. These include the precise movements required for buttoning shirts, tying shoes, feeding themselves, using phones, or inserting a key in a lock. For some, the performance of such tasks is frustrated by a lack of hand steadiness. Older adults have less hand steadiness than younger adults (Ranganathan et al., 2001), as well as higher amounts of co-contraction (Laidlaw et al., 2002). The present study suggests that the additional co-contraction seen in older adults is not a compensatory mechanism to decrease elbow angle variability. Instead, these results suggest that the co-contraction may be a source of the elbow angle variability itself. If corroborated by others, this insight might lead to a change in the way occupational therapists aid older adults in performing everyday tasks requiring precision.

2.6 Conclusions

The hypothesis that elbow muscle co-contraction affects elbow angle variability was supported. The simulation predicted a non-zero, non-maximal co-contraction level that minimized elbow angle variability. The experimental results supported this finding by showing that co-contraction above normal levels led to increased elbow angle variability. While the subjects were unable to decrease their co-contraction below natural levels, their natural co-contraction level was in the same range as the optimal co-contraction level shown by the simulation.

2.7 References

Blanpied, P., & Smidt, G. L. (1993). The difference in stiffness of the active plantarflexors between young and elderly human females. Journal of Gerontology, 48(2), M58.

Burnett, R. A., Laidlaw, D. H., & Enoka, R. M. (2000). Coactivation of the antagonist muscle does not covary with steadiness in old adults. Journal of Applied Physiology, 89(1), 61.

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Graves, A. E., Kornatz, K. W., & Enoka, R. M. (2000). Older adults use a unique strategy to lift inertial loads with the elbow flexor muscles. Journal of Neurophysiology, 83(4), 2030.

Hamilton, A. F., Jones, K. E., & Wolpert, D. M. (2004). The scaling of motor noise with muscle strength and motor unit number in humans. Experimental Brain Research., 157(4), 417-430.

Harris, C. M., & Wolpert, D. M. (1998). Signal-dependent noise determines motor planning. Nature, 394(6695), 780-784.

Haruno, M., & Wolpert, D. M. (2005). Optimal control of redundant muscles in step-tracking wrist movements. Journal of Neurophysiology, 94(6), 4244-4255.

Holzbaur, K. R. S., Murray, W. M., & Delp, S. L. (2005). A model of the upper extremity for simulating musculoskeletal surgery and analyzing neuromuscular control. Annals of Biomedical Engineering, 33(6), 829-840.

Jones, K. E., Hamilton, A. F. C., & Wolpert, D. M. (2002). Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology, 88(3), 1533.

Laidlaw, D. H., Hunter, S. K., & Enoka, R. M. (2002). Nonuniform activation of the agonist muscle does not covary with index finger acceleration in old adults. Journal of Applied Physiology, 93(4), 1400-1410.

Lee, Y., & Ashton-Miller, J. (2011). The effects of gender, level of co-contraction, and initial angle on elbow extensor muscle stiffness and damping under a step increase in elbow flexion moment. Annals of Biomedical Engineering, 39(10), 2542-2549.

Moritz, C. T., Barry, B. K., Pascoe, M. A., & Enoka, R. M. (2005). Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle. Journal of Neurophysiology, 93(5), 2449-2459.

Murray, W. M., Buchanan, T. S., & Delp, S. L. (2000). The isometric functional capacity of muscles that cross the elbow. Journal of Biomechanics, 33(8), 943-952.

Osu, R., Kamimura, N., Iwasaki, H., Nakano, E., Harris, C. M., Wada, Y., & Kawato, M. (2004). Optimal impedance control for task achievement in the presence of signal-dependent noise. Journal of Neurophysiology, 92(2), 1199-1215.

Perreault, E. J., Kirsch, R. F., & Crago, P. E. (2001). Effects of voluntary force generation on the elastic components of endpoint stiffness. Experimental Brain Research, 141(3), 312-323.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. (2001). Effects of aging on hand function. Journal of the American Geriatrics Society, 49(11), 1478-1484.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2005). Can co-activation reduce kinematic variability? A simulation study. Biological Cybernetics, 93(5), 373-381.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieën, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Sinkjaer, T., Toft, E., Andreassen, S., & Hornemann, B. C. (1988). Muscle stiffness in human ankle dorsiflexors: Intrinsic and reflex components. Journal of Neurophysiology, 60(3), 1110-1121.

Chapter 3

Effect of Elbow Moment Loading Paradigm on Natural Elbow Muscle Co-contraction Level and Elbow Angle Positional Variability

3.1 Abstract

For the same external quasi-static sagittal plane moment applied to a near-upright forearm, subjects exhibit higher levels of elbow muscle activation, or elbow muscle co-contraction, when the moment is applied by a downward force on the wrist in comparison to a horizontal force. The cause of this increased elbow muscle co-contraction is unknown. This study investigates the underlying cause of this increased co-contraction by eliminating confounding effects, such as the axial load on the forearm. The Variable-Stiffness Signal Dependent Noise (SDN) Theory is applied to a simulation of the elbow joint bent at 90° of flexion with the forearm held vertically. The elbow was loaded using three different loading paradigms: destabilizing, stabilizing, and constant. The destabilizing loading paradigm is a variable elbow extension moment that increases with decreasing elbow flexion angle. The stabilizing loading paradigm is a variable elbow extension moment that decreases with decreasing elbow flexion angle. The final loading paradigm is a constant loading which is invariant of elbow flexion angle. The simulation predicts an optimal elbow muscle co-contraction level that minimizes elbow angle positional variability. The optimal co-contraction level is highest for the destabilizing loading paradigm, intermediate for the constant loading paradigm, and lowest for the

stabilizing loading paradigm. The same results were found for the elbow angle positional variability.

Fourteen healthy younger adults were tested experimentally using the same loading conditions. The experimental results found increased positional variability for co-contraction above natural levels for each loading condition ($p < 0.005$), supporting the simulation for high levels of elbow muscle co-contraction. Natural co-contraction levels for the destabilizing and constant loading paradigms were found to be 9% higher than for the stabilizing loading condition ($p < 0.05$). The positional variability for the natural co-contraction trials was not found to be significantly different between the loading paradigms. We conclude that loading paradigm, not just loading magnitude, effects natural elbow muscle co-contraction levels and that increased elbow muscle co-contraction leads to increased elbow angle positional variability.

3.2 Introduction

Positional variability of the hand, the inverse of positional steadiness, is required for many activities of daily living, such as eating with a spoon and inserting a key into a lock. Variations in the position of the hand are due in part to variation in elbow angle caused by the stochastic noise in muscle contractile force (Galganski et al., 1993; Jones et al., 2002; and Moritz et al., 2005). Increasing muscle co-contraction has been observed with increased accuracy demands (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006). Because of this, investigators believe that increased co-contraction is used as a compensatory mechanism to decrease positional variability of the forearm (Faisal et al., 2008 and Selen et al., 2005). However, no relation was found between co-contraction and acceleration among subjects (Burnett et al., 2000 and

Laidlaw et al., 2002). The relationship between co-contraction and hand positional variability during a task has not been adequately studied.

The findings from Chapter 2 suggest that increasing co-contraction levels above the natural level in younger adults increases positional variability during an upper extremity quasistatic task of resisting a constant external load. This suggests that while co-contraction naturally increases during positionally demanding tasks (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006), it may be inhibitory rather than compensatory. However, the effect of elbow muscle co-contraction on positional-variability is known only for constant loads applied perpendicularly to the subject's wrist. For instance, the relationship between co-contraction and positional variability for loads having an axial component (parallel to the long axis of the forearm) has not been studied.

While the positional variability during a quasistatic task containing an axial load has not been studied, the natural elbow muscle activation has been studied (Stokes and Gardner-Morse, 2000). Stokes and Gardner-Morse investigated the effect of loading paradigm on elbow muscle activation using an external elbow extension moment. Subjects were seated with their elbows and upper arms resting on a table. Their elbows were positioned at 80° of flexion with their forearm held nearly vertical. The forearm was loaded either horizontally or vertically at the wrist to create an elbow extension moment. The magnitude of the wrist load was adjusted to generate equal elbow extension moments in the two loading paradigms. Subjects exhibited higher muscle activations during vertical loading than during horizontal loading. The researchers proposed that the muscle activity was higher during vertical loading in order to stabilize the forearm under the more challenging load. However, no research has been done to determine why the vertical loading needs to be stabilized.

While vertical and horizontal loading of the forearm can create identical moments about the elbow, they represent vastly different loading paradigms. Vertical loading of the wrist can be viewed as a moment about the elbow and a compressive axial load on the forearm. While there is an axial component to the horizontal loading paradigm, it is significantly smaller and in tension. The vertical loading may result in higher muscle activation for one of these three reasons: 1) to balance the forces produced by the axial load, 2) to provide stability to the elbow joint, or 3) to balance the variable moment about the elbow created by vertical loading. However, these three potential reasons have not been tested to determine whether and why they lead to higher muscle activation.

One theory that predicts elbow muscle activation levels is the Signal Dependent Noise (SDN) Theory (Harris and Wolpert, 1998 and Haruno and Wolpert 2005). SDN Theory supposes that natural muscle activation minimizes positional variability. Since the subjects from the study conducted by Stokes and Gardner-Morse (2000) exhibited higher muscle activation during vertical loading than during horizontal loading, SDN Theory suggests that higher muscle activation was necessary to minimize the positional variability during vertical loading than during horizontal loading.

In their original experiment, Harris and Wolpert (1998) developed the SDN Theory to minimize what they term positional variability. The positional variability used in the study was the variability of the hand position at the end of a movement; I will call this positional variability the endpoint positional variability. Since the theory minimized the endpoint positional variability, the theory was not applicable to the quasistatic task that subjects performed in the study of Stokes and Gardner-Morse (2000). However, in Chapter 2 the SDN Theory was modified to minimize the dynamic positional variability (variability of the hand *during* the task) for quasistatic tasks. This new Variable-Stiffness SDN Theory suggests that the higher natural muscle activation exhibited during the vertical loading paradigm was necessary in order to minimize the dynamic positional variability.

The variable external elbow moment of the vertical loading paradigm of Stokes and Gardner-Morse may amplify the dynamic elbow positional variability and therefore be a cause of increased co-contraction. If the elbow flexion moment generated by the subject is less than the external elbow extension moment, the net moment about the elbow will be an extension moment. This extension moment will cause the elbow to extend (shown by the dashed arm, Figure 3.1), which lengthens the moment arm of the external vertical force (MA_2 to MA_3, Figure 3.1). The increased moment arm will increase the elbow moment generated by the vertical load. This increases the difference between the external moment and the moment generated by the elbow muscles. This larger moment difference produces a larger than normal deviation from the desired position.

Similarly, if the elbow flexion moment generated by the subject is greater than the external elbow extension moment, the net moment about the elbow will be a flexion moment. When the elbow flexion angle increases (dotted arm, Figure 3.1), the moment arm for the external load decreases (MA_2 to MA_1) resulting in a lower external moment generated about the elbow. This again increases the difference between the moment generated by the muscle and the external moment, which would cause a larger than

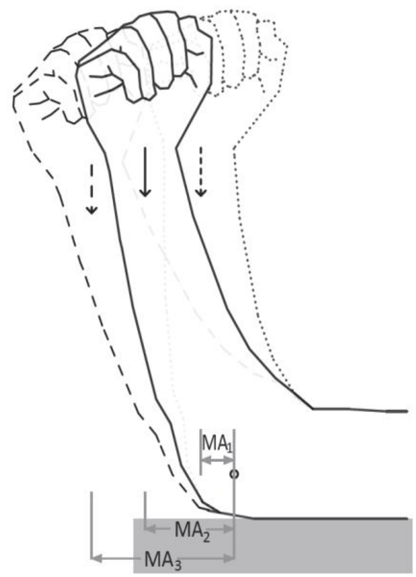


Figure 3.1: The moment created by a vertical load at the wrist decreases with increasing elbow flexion angle due to the change in moment arm about the elbow joint center.

normal deviation from the desired position. To compensate for this increase in positional variability, the optimal level of co-contraction may increase to utilize the increased joint stiffness associated with higher muscle activation.

This working hypothesis for explaining increased co-contraction during vertical loading is tested in this study by comparing three loading paradigms: constant loading, destabilizing loading, and stabilizing loading. Constant loading is a perpendicular load applied to the wrist that creates a near-constant moment about the elbow. This is nearly identical to the horizontal loading paradigm in Stokes and Gardner-Morse (2000). The destabilizing loading paradigm uses a variable horizontal load that increases with decreasing elbow flexion angle to simulate the variable loading of a vertical load. This loading is called a destabilizing loading paradigm due to the increased deviation from the desired position described above for the vertical load. However, loading the elbow with a variable horizontal load produces no axial load on the forearm; therefore, there will be no need for additional muscle activation to stabilize the elbow or balance the axial load. Lastly, the stabilizing loading paradigm has the opposite load versus elbow angle effect from the destabilizing loading paradigm. The stabilizing loading paradigm decreases the difference between the generated elbow moment and the external elbow moment. These loading paradigms were used to test the primary hypothesis presented in Chapter 1 as $H_{3.1}$: *Optimal/natural elbow co-contraction is higher for a destabilizing loading paradigm than a stabilizing loading paradigm*. If this hypothesis is proven true, the difference in co-contraction level can be used to test the effect of lowered co-contraction on positional variability (see section 4.3). If proven false, this would suggest that increased co-contraction is not used during destabilizing tasks. This study also tested the secondary hypothesis $H_{3.2}$: *Increasing elbow muscle co-contraction above the natural level will result in higher elbow angle positional variability.*

3.3 Methods

3.3.1 Modeling Methods

The computer model described in section 2.3.1 was modified to determine the effect of variable loading on optimal co-contraction levels. The same 2D, single joint, eight-muscle model was used as previously described (Figure 2.1). The variability of the muscle moments and the elbow stiffness corresponding to the muscle activations were calculated as discussed in Chapter 2. The relative displacement was then calculated as:

$$\Delta\theta = \frac{Muscle\ Moment\ Variability}{stiffness\ coefficient + external\ moment\ slope}$$

As shown in Figure 3.2, the external moment slope is -4% of the maximal voluntary contractile (MVC) force per degree of elbow flexion angle for the destabilizing loading paradigm, 0 for the constant loading paradigm, and 4% MVC/degree of elbow flexion angle for the stabilizing loading paradigm. The external moment slope in the denominator accounts for the changing external moment when the elbow angle is displaced. This model was used to reduce the complexity and to create a basic model that could be used to understand the effect of elbow muscle activation on elbow angle positional variability.

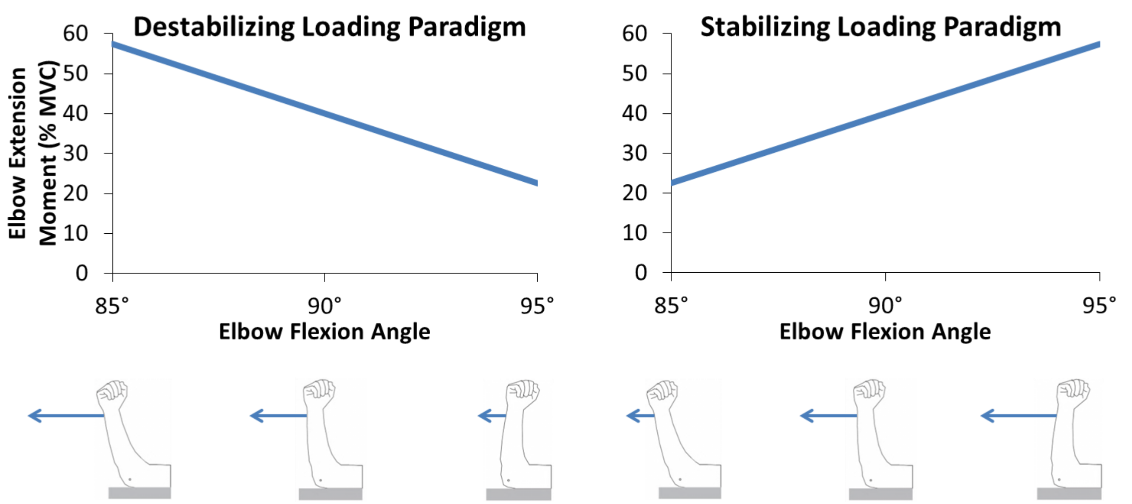


Figure 3.2: The relationship between elbow loading and elbow flexion angle for the destabilizing loading paradigm and stabilizing loading paradigm. The destabilizing load was designed to mimic the elbow position and moment relationship of a wrist vertically loaded with a flexion angle of 80°. The elbow angle was shifted 10° for this experiment to allow subjects to start in a more natural position.

3.3.2 Experimental Methods

3.3.2.1 Subjects

Fourteen right-handed, healthy younger adults (7 male and 7 female) between 21 and 30 years of age (mean age = 25 years) signed a statement of consent to take part in this institutional review board-approved study. Subjects were screened for previous injuries to the arm, shoulder, and spine, neurological disabilities, and musculoskeletal conditions.

3.3.2.2 Experimental Apparatus

The experimental apparatus consisted of a set of seven constant force springs, a retaining rod, a plate, and a padded table (Figure 3.3). The springs were mounted on individual axles attached to a custom-made bracket. The axles were held in place by two mounting plates that were attached to the wall. The constant force springs varied in strength from 1 lb to 24 lbs. This allowed the total force to be varied in 1-pound increments between 1 lb and 72 lbs.

Cables were attached to the springs to transfer the force to the subject. The springs were coupled together using a single metal ring. A cable was attached to the metal ring and the loading plate. The loading plate was a solid piece of aluminum mounted on an axle. The plate had three loading holes for attaching the cable to create the loading paradigms described in section 3.3.2.3 (Figure 3.3). The subject's loading cable was attached to the "Constant Loading / Subject Hole" on one end and the subject's wrist cuff on the other. The forearm was allowed to rotate within the wrist cuff to prevent a pronation or supination moment from being exerted on the forearm. Each subject placed his or her elbow on the padded table. The chair was adjusted to orient the subject's shoulder slightly higher than the padded table. A curtain was placed between the subject and the apparatus. This prevented the subject from focusing on the apparatus during the trials. To ensure consistency, an "X" was placed on the curtain for the subjects to focus on during the trial.

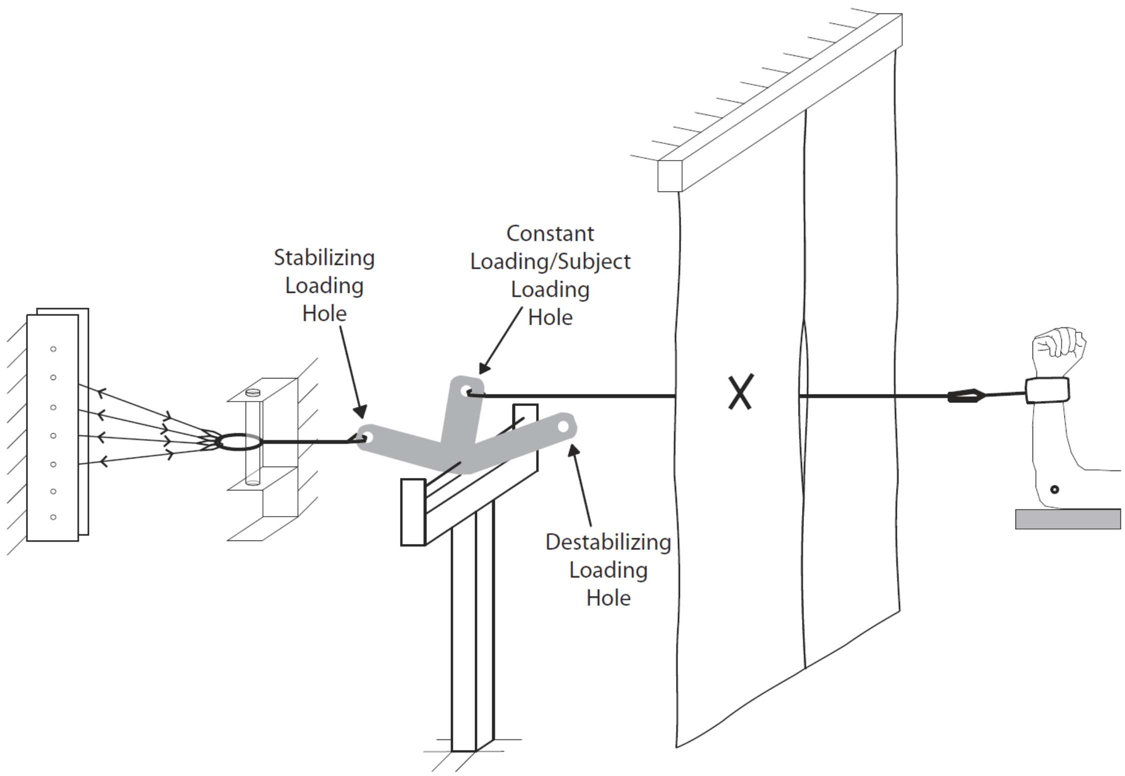


Figure 3.3: Diagram of the experimental setup for the trials. The spring set was attached to the wall and contained seven constant force springs. The springs selected for the trial were connected to a metal ring in order to combine them into a single force. A retaining rod was put through the metal ring to hold the spring force before the trial began and to give an audible cue for the subject. The loading plate created the three loading paradigms due to the changing moment arms of the plate as it rotates about its axle. A curtain was placed between the subject and the loading apparatus. An "X" was placed on the curtain to help the subject focus on the task rather than other distractions. A wrist cuff was placed on the subject's wrist in order to apply the elbow extension moment to the arm.

Before each trial began, the springs were preloaded and a retaining rod was placed through the metal ring to hold it in place. At the beginning of the trial, the subjects increased their elbow flexion moment until the retaining rod fell. The retaining rod created an audible cue for the subjects. This cue informed the subjects that the trial had begun and they should maintain a constant elbow moment (described below in section 3.3.2.7).

3.3.2.3 External Force

The external force was generated using constant force springs as described above. The spring cable attached to the loading plate in order to create the desired loading paradigms of destabilizing, constant, and stabilizing. As indicated in Figure 3.3, the loading cable was attached to one of the three holes on the plate. For the destabilizing load, if the subject's elbow was at 100° of flexion, the hole was aligned horizontally with the plate mounting hole, generating no moment on the plate and thus no force on the subject's wrist. The force increased from 0% of the subject's maximal voluntary contraction (MVC) when the elbow was at 100° of flexion to 80% of the subject's MVC at 80° of flexion (Figure 3.2). For constant loading, the force was transmitted to the subject's wrist without any modification. For the stabilizing load, the force increased as the elbow flexion angle increased (Figure 3.2).

The force was measured using an axial force transducer (ATI Industrial Automation, Apex, North Carolina). The force transducer was attached to the subject's wrist cuff to measure the elbow-moment-generating force on the subject. The force data were collected at a frequency of 1 kHz on a central computer.

3.3.2.4 Kinematic Data

The kinematic data of the arm were recorded during the trial. Infrared markers were placed on the medial side of the upper arm, elbow, and forearm of the subject. A Certus Optotrak (Northern Digital Inc., Waterloo, Canada) camera system was used to record the position of the markers and calculate the elbow angle. The kinematic data were collected at a frequency of 1 kHz on a central computer. The data were filtered in Matlab using a third order high pass Butterworth filter with a cutoff frequency of 4 Hz. The filtering allowed the effect of positional drift to be removed from the data. Removing the positional drift allowed the positional variability to be measured without the subject focusing on the position of his or her arm.

3.2.2.5 EMG Data

The electromyographic (EMG) signal from four elbow flexor muscles (long head biceps, short head biceps, brachialis, and brachioradialis) and four elbow extensor muscles (medial head triceps, long head triceps, short head triceps, and anconeous) were collected using blue dot surface electrodes with 2-cm spacing along the muscle fiber direction. The signal was collected by a Myosystem 2000 (Noraxon U.S.A. Inc., Scottsdale, Arizona) and recorded at 1 kHz on a central computer.

3.2.2.6 Maximal Voluntary Force Trials

Subjects were asked to perform maximal force trials to determine their maximal voluntary contraction (MVC) force. The wrist cuff was attached to the subjects, and they placed their elbows, bent at 90°, on the padded table with their forearm held vertical. A cable attached the cuff to the wall to provide resistance. Subjects were instructed to increase their force over the first three seconds of the trial and then hold their maximal force for two seconds. The researcher told the subjects when to begin and then counted "one … two … three … hold … hold … relax." Each subject's MVC was taken to be the highest average over a one-quarter of a second period during the trial. The trials were repeated until two trials produced forces within 10% of each other. No subject repeated the maximal voluntary force trials more than five times.

3.2.2.7 Elbow Angle Variability Trials

The elbow angle variability trials consisted of the three loading paradigms and three co-contraction levels. The three levels of co-contraction were designated as natural, medium, and maximal. Natural co-contraction was defined as the amount of co-contraction the subject would naturally use when resisting the load. Maximal co-contraction was defined as the maximum amount of co-contraction a subject could generate while resisting the load. In order to generate an appropriate medium co-contraction level, subjects were shown a real-time graph of their muscle activation. Subjects were allowed to practice generating the three levels of co-contraction. Once the subject was able to generate the desired co-contraction levels, the visual feedback was removed.

The order of the loading paradigms was chosen randomly. At the beginning of each new loading paradigm, subjects were allowed to conduct practice trials to familiarize themselves with the loading paradigm. Instructed co-contraction levels were randomly chosen within each loading paradigm. The natural co-contraction level was repeated five times, while the maximal and natural trials were each repeated three times.

At the beginning of each trial, the subject was instructed as to which co-contraction level to generate, asked if he or she was ready, and then told to begin. The subject increased their elbow flexion moment until they heard the audible cue from the dropping retaining rod (Figure 3.4). Once the audible cue was heard, the subject was to maintain a constant elbow flexion moment for the remainder of the trial. The spring force was changed for each subject to allow the retaining rod to fall at 40% of the subject's MVC. Subjects were instructed to increase their force at the same rate for each trial regardless of loading paradigm or co-contraction level. This created the same pre-trial loading for each loading paradigm. Since the task demand affects a subject's co-contraction level (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006), subjects were not asked to maintain their exact starting position. While this allowed for positional drift during the trial, the drift was removed by filtering the kinematic data as described in section 3.2.2.4.

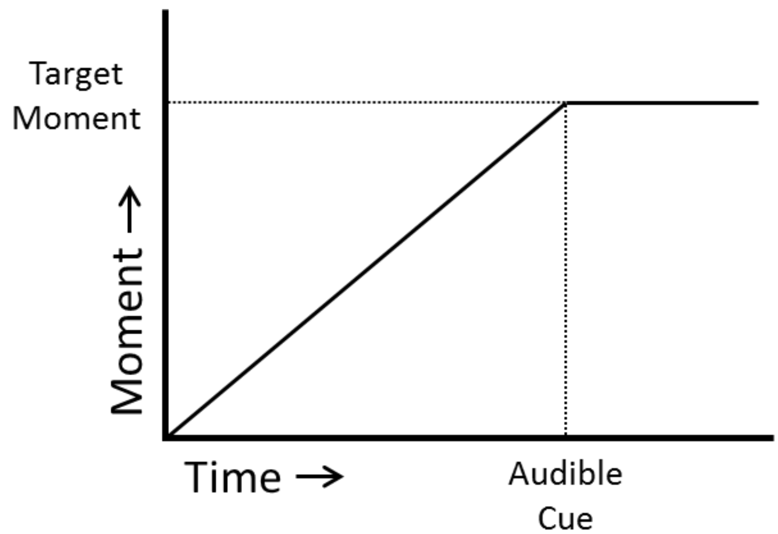


Figure 3.4: Idealized loading of the subject's elbow for the beginning of each trial. Subjects increased their elbow flexion moments until they heard the audible cue from the retaining rod. Subjects were instructed to maintain the elbow flexion moment level once they heard the audible cue.

3.2.2.8 Statistical Analysis

Statistical analyses were performed using Minitab (State College, Pennsylvania). A general linear model was used to analyze the data while treating subjects as a random effect to remove the differences between subjects. The loading paradigm was used as a categorical variable to predict subjects' natural co-contraction levels using the model $Co\text{-}contraction = \beta_0 + \beta_1 x_1 + \beta_2 x_2 + \beta_3 x_3 + \beta_4 x_3$ where β_0 was a constant, β_1 was a coefficient for the loading paradigm, x_1 was the loading paradigm, β_2 was a coefficient for the force, x_2 was the force, β_3 was a coefficient for the trial number, x_3 was the trial number, and β_3 was a coefficient for the trial number and subject interaction. The least squares means of the natural co-contraction levels were calculated. A Tukey post-hoc analysis was performed to compare the natural co-contraction levels during the loading paradigms. This method was used to test the primary hypothesis $H_{3.1}$: *Optimal/natural elbow co-contraction is higher for a destabilizing loading moment than a stabilizing loading moment*.

The general linear model was also used to test the secondary hypothesis $H_{3.2}$: *Increasing elbow muscle co-contraction above the natural level will result in higher elbow angle positional variability*. In this analysis, the co-contraction level was used to predict the level of positional variability using the model $Positional\ Variability = \beta_0 + \beta_1 x_1 + \beta_2 x_2 + \beta_3 x_1$ where β_0 is a constant, $\beta_1$is the coefficient for force, x_1 is the force, β_2 is the coefficient for co-contraction, x_2 is the co-contraction level, and β_3 is the coefficient for the subject*force interaction. The effect was considered significant if the p-value was less than 0.05. A Tukey post-hoc analysis was performed with the positional variability to test the difference between the loading paradigms.

3.4 Results

3.4.1 Simulation Results

The simulation results predict an optimal level of co-contraction that minimizes positional variability (θ_{SD}) when the elbow is exposed to a variable loading paradigm, such as the destabilizing loading described earlier (Figure 3.4). The optimal level of co-contraction was found to be higher for the destabilizing loading paradigm than for the constant loading paradigm (shown in blue and red respectively in Figure 3.5). Similarly, an optimal co-contraction level was found for the stabilizing load. The optimal co-contraction level was lower for the stabilizing loading paradigm than for the destabilizing loading paradigm.

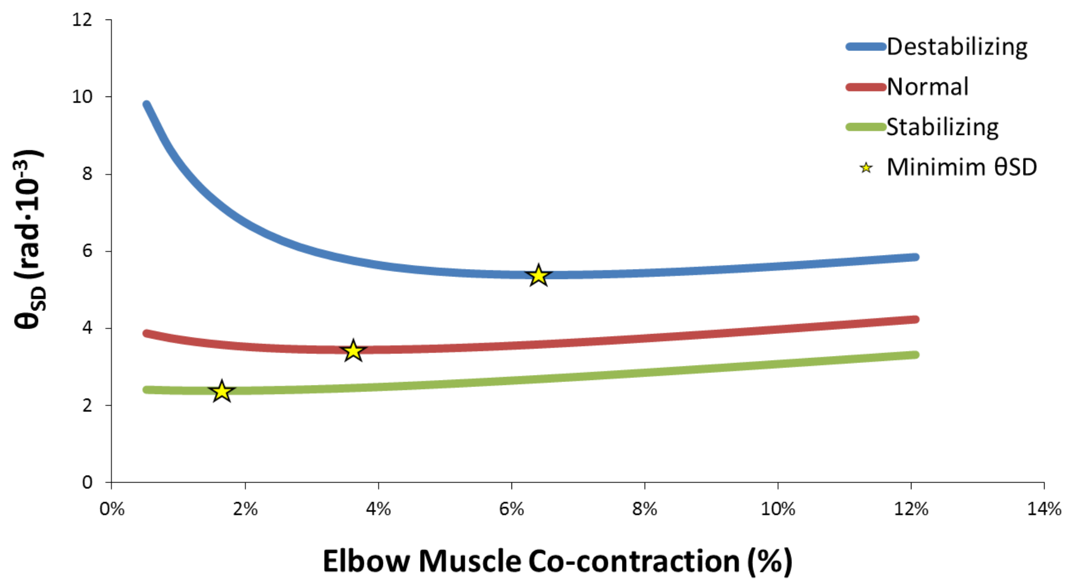


Figure 3.5: Simulation results for the three elbow moment loading paradigms found optimal levels of co-contraction that minimized elbow angle variability for all three loading paradigms.

The simulation suggests elbow angle positional variability is higher during destabilizing loading than during constant loading for a given co-contraction level (Figure 3.5). Also, when comparing the minimal positional variability for each loading paradigm, the positional variability was higher during destabilizing loading than during constant loading. Again, the opposite was true for stabilizing loading: positional variability was lower for a given co-contraction level and for the minimal positional variability.

3.4.2 Experimental Results

The scatterplot for subjects' co-contraction at and above their natural level during the destabilizing, constant, and stabilizing loading paradigms are shown in Figure 3.6. A basic linear regression suggests that all 14 subjects exhibited higher levels of elbow angle positional variability when subjects increased their elbow muscle co-contraction level for both the destabilizing and constant loading paradigms. Thirteen of the 14 subjects increased their elbow angle positional variability with increased elbow muscle co-contraction during the stabilizing loading paradigm trials. Co-contraction above the subject's natural level was found to significantly increase the positional variability for each of the loading paradigms (Table 3.1). This finding generalizes the results from Chapter 2 to variable loading paradigms. There was no statistical difference in the coefficient of the co-contraction effect for the three loading paradigms.

Table 3.1: The general linear model for elbow angle positional variability suggests that elbow muscle co-contraction significantly affects elbow angle positional variability.

Loading Condition	Variable	Coefficient	SE Coefficient	p-value
Destabilizing	Subject	NA	NA	0.01
	Subject*Force	NA	NA	0.01
	Force	-2.55	1.30	0.05
	Co-contraction	3.88	0.48	0.00
Constant	Subject	NA	NA	0.00
	Subject*Force	NA	NA	0.00
	Force	-4.11	1.32	0.00
	Co-contraction	3.93	0.36	0.00
Stabilizing	Subject	NA	NA	0.00
	Co-contraction	3.29	0.34	0.00

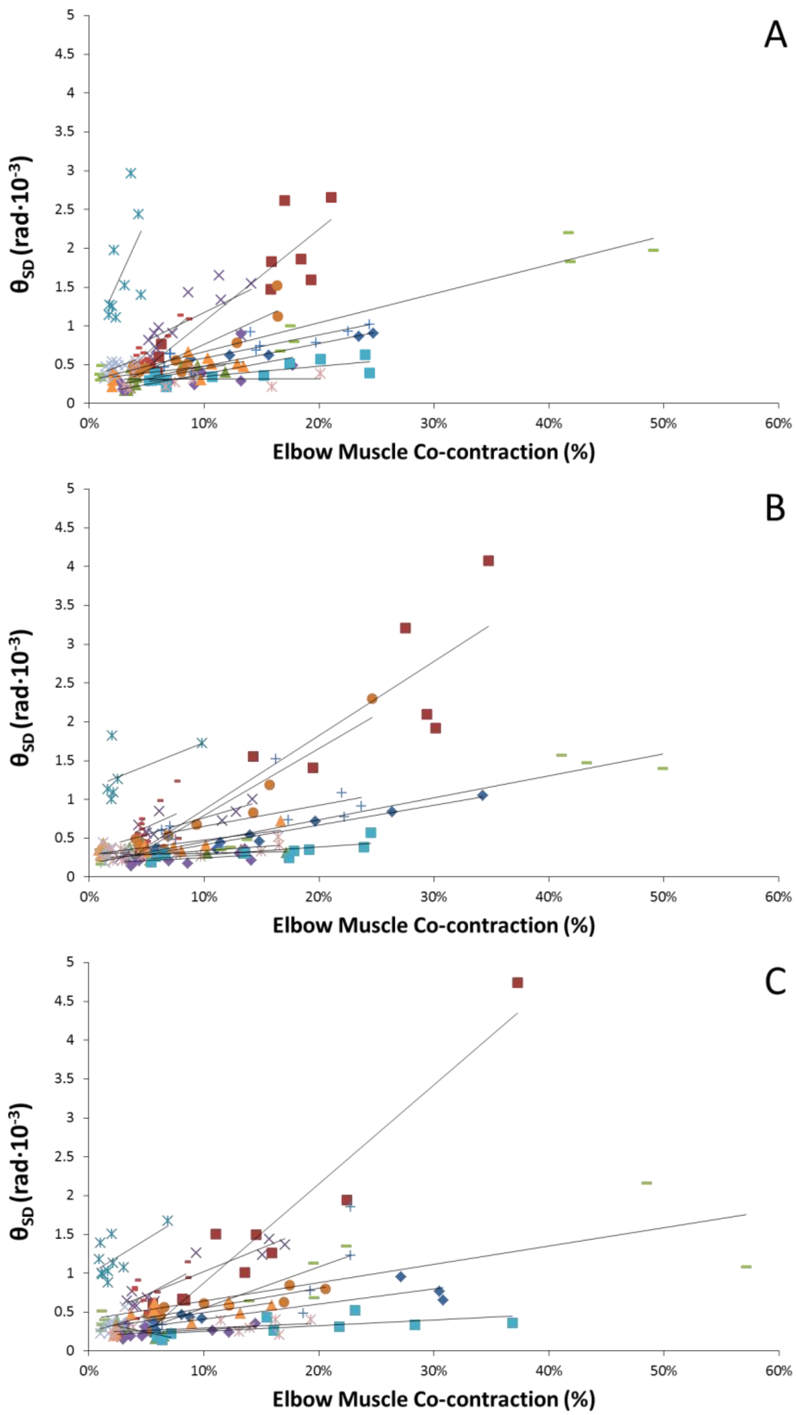


Figure 3.6: Scatter plot showing the relationship between elbow angle positional variability and elbow muscle co-contraction at, and above, natural levels under the A) destabilizing, B) constant, and C) stabilizing loading paradigms. Regression lines indicate the best fit for each subject. Across the three paradigms, the regression line suggests that at least 13 of the 14 subjects increase elbow angle variability with increased elbow muscle co-contraction level.

The loading paradigm was found to be a significant factor in predicting the subject's natural co-contraction level ($p < 0.05$, Table 3.2). The co-contraction level during the stabilizing loading paradigm was found to be significantly higher than for the destabilizing and constant loading paradigms ($p < 0.05$, Figure 3.7).

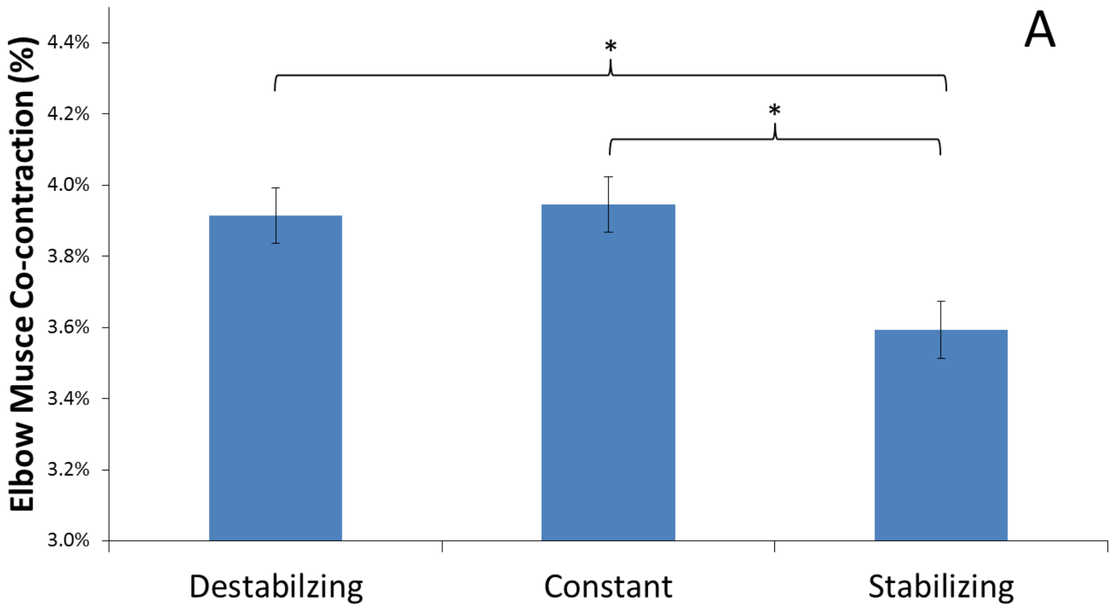
The loading paradigm trended towards, but was not found to be a significant factor in predicting the elbow angle positional variability ($p = 0.1$, Table 3.3). When comparing the least squares means, the elbow angle variability during destabilizing loading paradigm trended towards being higher than during constant loading paradigm ($p < 0.1$, Figure 3.7). Fatigue and learning were accounted for by checking for the effect of trial number on positional variability. Trial number was not found to have a significant effect on elbow angle positional variability.

Table 3.2 The general linear model for natural elbow muscle co-contraction suggests that the loading paradigm is a significant factor ($p = 0.01$) in predicting the natural elbow muscle co-contraction level.

Variable	**p-value**
Subject	0.00
Loading	0.01
Force	0.00
Trial	0.16
Subject*Trial	0.00

Table 3.3: The general linear model suggests that the loading paradigm trends towards, but is not a significant predictor of elbow angle variability ($p = 0.1$).

Variable	**p-value**
Subject	0.00
Loading	0.10
Force	0.20
Subject*Force	0.00



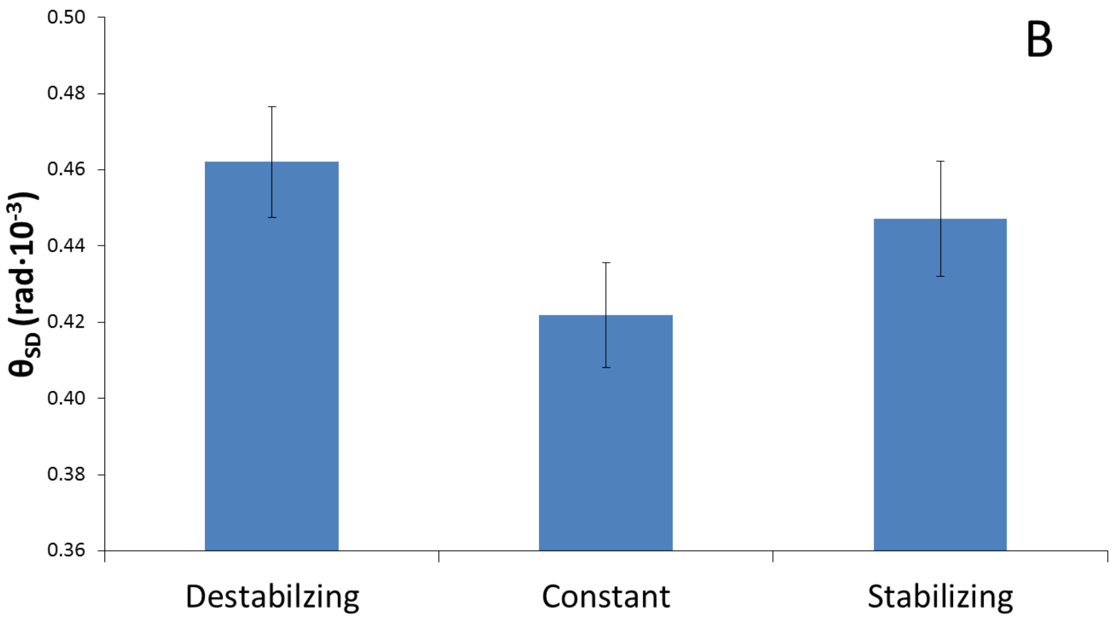


Figure 3.7: Bar graphs showing results for A) elbow muscle co-contraction level and B) elbow angle positional variability for the three different elbow moment loading paradigms. Error bars represent the standard error for the least squares means of each loading paradigm. * indicates $p < 0.05$.

3.5 Discussion

The primary hypothesis $H_{3.1}$ was supported by both the simulation and the experiment (Figure 3.8). The co-contraction level was significantly higher during destabilizing loading paradigm than during the stabilizing paradigm. Also, the co-contraction during the constant loading paradigm was found to be higher than during the stabilizing loading paradigm. Both of these results agree with the Variable-Stiffness SDN Theory. While the co-contraction during constant loading was not lower than the co-contraction during destabilizing loading, the significant results of natural co-contraction levels support the Variable-Stiffness SDN Theory.

The elbow angle variability results from the experimental trials trended towards supporting the Variable-Stiffness SDN Theory. The elbow angle variability during the destabilizing loading trended towards being lower than the variability during the constant loading. However, further study must be done before the theory is fully validated as this finding was not significant.

Furthermore, both the simulation (Figure 3.5) and the experimental data (Figure 3.8) support $H_{3.2}$. Variable-Stiffness SDN Theory states that a subject's natural co-contraction level will produce the lowest positional variability. The experimental results demonstrate that increased elbow muscle co-contraction above the natural co-contraction level increase positional variability, which supports the variable-stiffness SDN Theory and the secondary hypothesis.

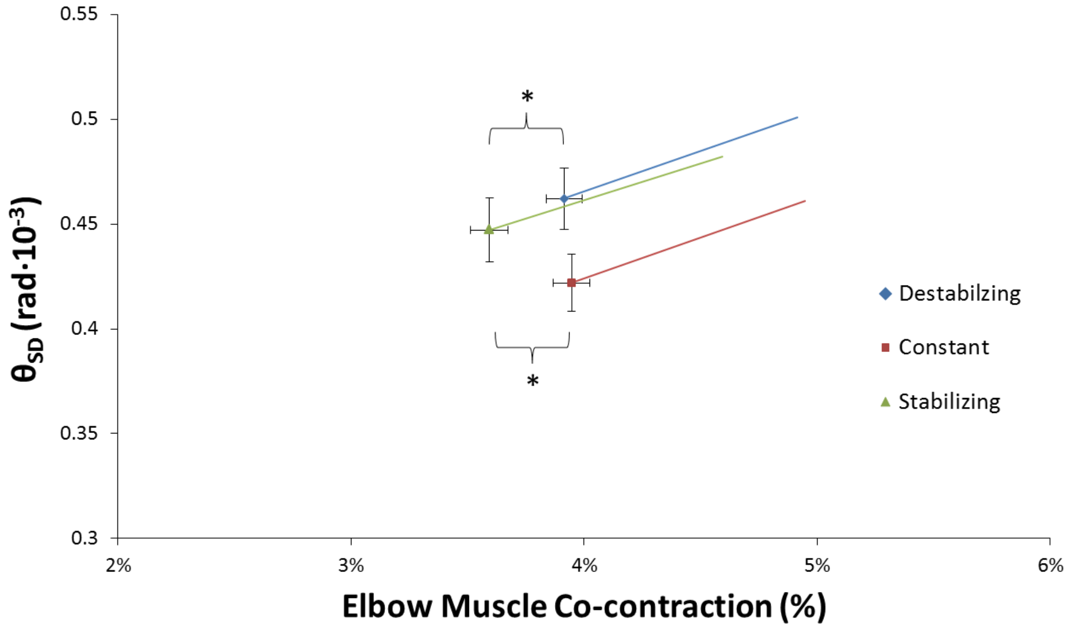


Figure 3.8: Summary of the experimental results showing the least squares means and corresponding standard errors for the natural elbow muscle co-contraction levels and elbow angle positional variability for the three different elbow moment loading paradigms. Positional variability increases with co-contraction above the natural co-contraction level regardless of loading paradigm ($p < 0.005$) as indicated by the positive sloping lines. Subjects co-contracted less during stabilizing loading than during constant loading (* indicates $p < 0.05$) and destabilizing loading ($p < 0.05$). Subjects trended towards higher positional variability during destabilizing loading than during constant loading, but this trend was not significant ($p < 0.1$).

One limitation of the study is that it did not experimentally test the effect of elbow muscle co-contraction on elbow angle positional variability below the natural level. Without experimentally verifying the left side of the curve in Figure 3.5 we cannot conclusively state that there is a non-zero co-contraction value that minimizes positional variability. However, two results from the study do support the model: 1) the ability of the Variable-Stiffness SDN Theory to predict increased co-contraction levels in destabilizing loading paradigms and decreased co-contraction levels in stabilizing loading paradigms and 2) the effect of increased co-contraction above the natural levels on positional variability.

Another limitation of this study was the inertia present in the loading of the elbow, which may reduce the positional variability. The loads were created using a system of springs in order to remove the inertia that weights would introduce; however, this resulted in the use of a loading plate which added a small amount of rotational inertia to the system. In order to minimize the changes in inertia between loading paradigms, the same loading plate was used for each trial. This ensured that the comparisons between the loading paradigms were not influenced by changes in inertia. While the inertia could not be completely removed, it was minimized and kept constant throughout the experiment.

Similarly, the friction in the system may also have reduced the magnitude of the positional variability. Bearings were used to mount the springs to the mounting bracket axels as well as for the plate itself. This reduced, but did not eliminate, the friction in the system.

The loads generated by the springs were assumed to be constant throughout the range of motion. The springs were aligned vertically and joined at the metal ring. When the ring was pulled forward by the subject, the angles from the springs to the ring changed. This is not thought to be a major source of error due to the small range of motion during the trials. To minimize the effect of the varying force load, the force measured during the trials allowed the general linear model to compensate for this loading error.

The loads during these trials are idealized and not likely to be seen in activities of daily living. Most activities of daily living involve greater inertia than was present in this experiment. Also, in a real-life scenario, the load is not likely to be localized to the elbow joint. This is because almost all loading of the elbow also loads the shoulder joint. Since most of the muscles that were measured during this experiment also span the shoulder joint, it is likely that their natural activation level would be different during an activity of daily living. An idealized scenario was used to control for as many factors as possible. Partially isolating the elbow joint minimized the muscle activation associated with the loads on the shoulder.

The simulation was intentionally kept simple to allow for easy manipulation and understanding. This research used a two-dimensional model of the elbow rather than a three-dimensional model. The two-dimensional model neglected the external out-of-plane forces that exist both during the experiment and during activities of daily living. Out-of-plane forces were assumed to not significantly affect the results.

Since a two-dimensional model was used for the simulation, the internal out-of-plane forces caused by the muscles in the model were also neglected. Again, these forces were assumed to have minimal effect on the results of the model. Despite the many simplifications of the model for the simulation, the simulation correctly predicted the effect of increased co-contraction on positional variability as well as many of the effects of variable loading on natural co-contraction levels.

3.6 Conclusions

Above natural co-contraction increases the positional variability of the elbow joint under variable loading for healthy younger adults. The simulation based on the Variable-Stiffness SDN Theory correctly predicted the increase in natural co-contraction level when compared to constant loading. This demonstrates that the Variable-Stiffness SDN Model is a good predictor of co-contraction behavior and suggests that the increased muscle activation present during certain loading paradigms such as the vertical loading from Stokes and Gardner-Morse (2000), may act to decrease positional variability.

3.7 References

Burnett, R. A., Laidlaw, D. H., & Enoka, R. M. (2000). Coactivation of the antagonist muscle does not covary with steadiness in old adults. Journal of Applied Physiology, 89(1), 61.

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Galganski, M. E., Fuglevand, A. J., & Enoka, R. M. (1993). Reduced control of motor output in a human hand muscle of elderly subjects during submaximal contractions. Journal of Neurophysiology, 69(6), 2108-2115.

Gribble, P. L., Mullin, L. I., Cothros, N., & Mattar, A. (2003). Role of cocontraction in arm movement accuracy. Journal of Neurophysiology, 89(5), 2396-2405.

Harris, C. M., & Wolpert, D. M. (1998). Signal-dependent noise determines motor planning. Nature, 394(6695), 780-784.

Haruno, M., & Wolpert, D. M. (2005). Optimal control of redundant muscles in step-tracking wrist movements. Journal of Neurophysiology, 94(6), 4244-4255.

Jones, K. E., Hamilton, A. F. C., & Wolpert, D. M. (2002). Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology, 88(3), 1533.

Laidlaw, D. H., Hunter, S. K., & Enoka, R. M. (2002). Nonuniform activation of the agonist muscle does not covary with index finger acceleration in old adults. Journal of Applied Physiology, 93(4), 1400-1410.

Laursen, B., Jensen, B., & Sjøgaard, G. (1998). Effect of speed and precision demands on human shoulder muscle electromyography during a repetitive task. European Journal of Applied Physiology and Occupational Physiology,78(6), 544-548.

Moritz, C. T., Barry, B. K., Pascoe, M. A., & Enoka, R. M. (2005). Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle. Journal of Neurophysiology, 93(5), 2449-2459.

Osu, R., Kamimura, N., Iwasaki, H., Nakano, E., Harris, C. M., Wada, Y., & Kawato, M. (2004). Optimal impedance control for task achievement in the presence of signal-dependent noise. Journal of Neurophysiology, 92(2), 1199-1215.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2005). Can co-activation reduce kinematic variability? A simulation study. Biological Cybernetics, 93(5), 373-381.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieën, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Stokes, I. A. F., & Gardner-Morse, M. G. (2000). Strategies used to stabilize the elbow joint challenged by inverted pendulum loading. Journal of Biomechanics, 33(6), 737-743.

Chapter 4

Effect of Aging on the Optimal Co-contraction Level Required to Minimize Elbow Angle Positional Variability when Resisting an Elbow Extension Moment

4.1 Abstract

The effect of co-contraction on positional variability in older adults is unknown. Chapter 3 demonstrated that younger adults increase their elbow angle positional variability with increased elbow muscle co-contraction. These results suggest that the higher co-contraction levels naturally seen in older adults are the source of positional variability rather than a mechanism to compensate for positional variability. The current study tested the effect of elbow muscle co-contraction on elbow angle positional variability in older adults during a quasistatic elbow task of resisting an external elbow extension load. Three loading paradigms were used: a destabilizing loading paradigm that increased the elbow extension moment as the elbow flexion angle increased, a constant loading paradigm that did not vary the elbow extension moment with elbow flexion angle, and a stabilizing loading paradigm that decreased the elbow extension moment as the elbow flexion angle increased.

Older adults co-contracted their elbow muscles 32% more on average than younger adults performing the same strength matched task ($p < 0.005$). Older adults were also found to average 53% less elbow angle positional variability than younger adults ($p < 0.005$). Higher natural co-contraction levels were found during the destabilizing

loading trials than during the constant and stabilizing loading trials ($p < 0.005$). Higher positional variability was found during the stabilizing loading trials than during the constant loading trials ($p < 0.01$ and $p < 0.005$ respectively). Elbow angle positional variability was found to increase with increased co-contraction above the natural level for all loading paradigms ($p < 0.005$). We conclude that higher levels of co-contraction in older adults increases, rather than decreases their positional variability.

4.2 Introduction

Hand positional variability, the inverse of positional steadiness, is essential for the performance of self-care tasks such as inserting a key into a lock and eating with a spoon. The hand positional variability exhibited by many older adults inhibits their ability to perform tasks of daily living and care for themselves. If the source of this positional variability were better understood, new interventions for the elderly could be developed to help older adults live independently.

One source of positional variability is the variability in force generated by the muscle contractile elements over time (Galganski et al., 1993; Jones et al., 2002; and Moritz et al., 2005). This muscle force variability generates moment variability about a joint such as the elbow. This elbow moment variability would produce hand positional variability over time.

Older adults may have greater hand positional variability than younger adults. In a study by Ranganathan et al. (2001), subjects were asked to hold a 1 mm diameter pin inside a 2 mm diameter hole. Older adults touched the edge of the hole six times more than younger adults, indicating that older adults were less capable of maintaining a precise position. However, since no muscle activation data were recorded during the trials, no information is known about the muscle activation that may have led to this increased positional variability.

In a study by Smith et al. (1999), adults from 18 to 94 years of age removed a nut from differently shaped rods to test the functionality of the hand and arm. For subjects under 60 years of age, aging did not affect the time required to remove the nut from the rod. However, for subjects over 60 years of age, the amount of time required to remove the nut increased linearly with age. Again, only the timing results were measured. Therefore, the effect of muscle activation and aging is unknown.

Research suggests that older adults co-contract more than younger adults when performing a similar task (Seidler-Dorbin et al., 1998; Klein et al., 2001; and Hortobagyi and DeVita, 2006). This leads some researchers to believe that older adults increase their muscle activation, and therefore co-contraction level, to compensate for their lack of control (Faisal et al., 2008). Higher co-contraction and lower trial-to-trial endpoint variability was found in subjects reaching to smaller targets (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006).

However, it is possible that increasing co-contraction could lead to lower positional variability due to the increased stiffness of the joint (Selen et al., 2006 and Lee and Ashton-Miller, 2011). This joint stiffness would attenuate the effect of the force variability on the joint angle positional variability. Co-contraction has been shown to not affect positional variability when compared across subjects (Burnett et al., 2000). However, the results from Chapters 2-3 suggest that increasing co-contraction levels above natural for younger adults increases positional variability. It is unclear whether this result is true for older adults.

This study aims to address this knowledge gap by testing the primary hypothesis introduced in Chapter 1 as $H_{4.1}$: *Increased elbow muscle co-contraction above the natural level will increase elbow angle positional variability in older adults*. If the hypothesis is supported, then the co-contraction seen in older adults would appear to hinder the performance of precision tasks. But if it is rejected, then this may lead to a practical intervention for older adults. Therefore, the results of this study will help determine whether older adults can improve performance at tasks requiring low positional variability by increasing their co-contraction. The secondary hypothesis is $H_{4.2}$: *Older adults will exhibit higher natural co-contraction levels and higher positional variability compared to younger adults*. Filling these knowledge gaps will aid in understanding how co-contraction affects positional variability in the elderly.

4.3 Methods

4.3.1 Subjects

Fourteen healthy, right-handed, older adults (6 male and 8 female, mean age = 68 years) were recruited for this institutional review board-approved study. Subjects were screened for a history of hand, arm, or shoulder injuries and chronic diseases. Subjects signed an informed consent form prior to the study.

4.3.2 Kinematic Data

Kinematic data during the trials were collected using a Certus Optotrac camera (Northern Digital Inc., Waterloo, Canada). Three infrared markers were placed on the subject’s arm to determine the elbow joint angle. A marker was placed on the medial side of the subject's upper arm, elbow, and forearm. Kinematic data were collected at 1 kHz on a central computer.

4.3.3 Force Data

Force data were collected using an axial force transducer (ATI Industrial Automation, Apex, North Carolina). The force transducer was mounted between the subject's wrist cuff and the loading cable. Force data were collected at 1 kHz on a central computer.

4.3.4 EMG Data

EMG Data were collected using a Myosystem 2000 EMG system (Noraxon U.S.A. Inc., Scottsdale, Arizona). Blue dot style electrodes were applied, with 2 cm spacing along the muscle fibers, to the four main elbow flexor muscles (long head biceps, short head biceps, brachialis, and brachioradialis) and the four main elbow extensor muscles (long head triceps, medial head triceps, lateral head triceps, and anconeous).

4.3.5 Statistical Analysis

Minitab (State College, Pennsylvania) was used to generate a general linear model to test the effect of the loading paradigm on both natural elbow muscle co-contraction level and elbow angle positional variability. The general linear model was then used to calculate the least squares means to directly compare each loading paradigm. The general linear model was also used to determine the significance of increased elbow muscle co-contraction level on elbow angle positional variability.

4.3.6 Experimental Setup

The experimental apparatus, subject instructions, and trials were the same as in Chapter 3.

4.4 Results

The scatterplots of elbow muscle co-contraction at, and above, the natural co-contraction level and the corresponding elbow angle positional variability (θ_{SD}) for the destabilizing, constant, and stabilizing loading paradigms are shown in Figure 4.1. A basic regression line was fit to the data for each subject to show the effect of elbow muscle co-contraction on elbow angle positional variability. For the destabilizing loading paradigm, 13 of the 14subjects exhibited a positive relationship between elbow muscle co-contraction and elbow angle positional variability. The constant and stabilizing loading paradigms had similar results with 14 and 11 of the subjects exhibiting a positive relation between co-contraction and positional variability respectively. Positional variability increased with increased co-contraction above the natural level in older adults for each loading paradigm ($p < 0.005$, Table 4.1). This result supports the Variable-Stiffness Signal Dependent Noise (SDN) Theory that states that co-contraction above the natural level will lead to greater positional variability.

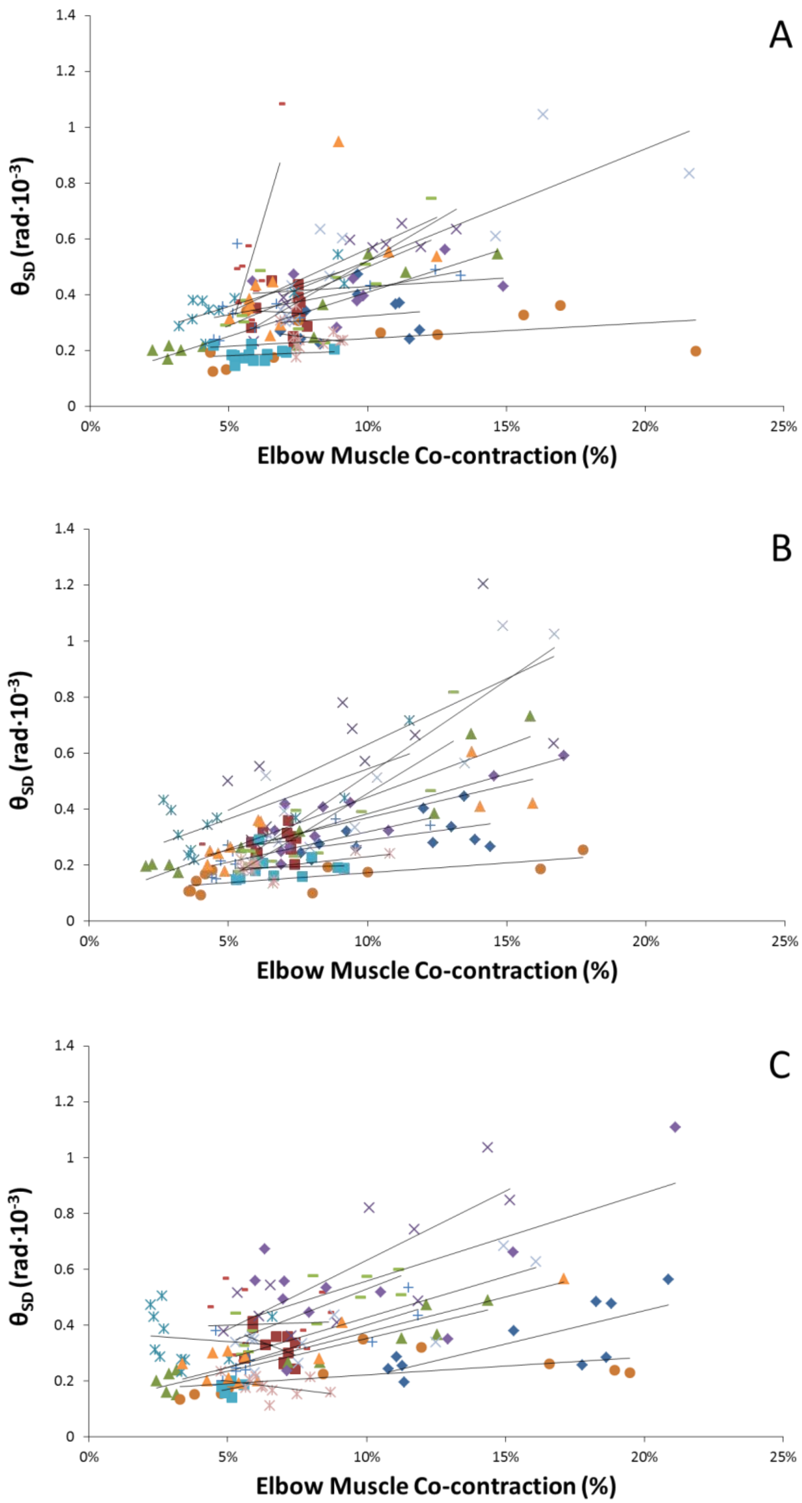


Figure 4.1: Scatter plot showing the relationship between elbow angle positional variability and elbow muscle co-contraction levels at, and above, natural level under the A) destabilizing, B) constant, and C) stabilizing loading paradigms. The regression lines have the same meaning as in Chapter 3, Figure 3.6. At least 11 of the 14 subjects exhibited a positive relationship between elbow muscle co-contraction and elbow angle positional variability.

Table 4.1: Significant factors in the general linear model for elbow angle positional variability for each of the three loading paradigms. Elbow muscle co-contraction significantly increased elbow angle positional variability in older adults during each loading paradigm as shown by the positive coefficient of co-contraction.

Loading Condition	Variable	Coefficient	SE Coefficient	p-value
Destabilizing	Subject	NA	NA	0.00
	Subject*Force	NA	NA	0.00
	Force	-0.81	0.54	0.13
	Co-contraction	2.17	0.33	0.00
Constant	Subject	NA	NA	0.06
	Subject*Force	NA	NA	0.02
	Force	0.54	0.57	0.34
	Co-contraction	2.90	0.34	0.00
Stabilizing	Subject	NA	NA	0.00
	Co-contraction	2.35	0.25	0.00

Older adults were found to have higher natural co-contraction levels than younger adults (Figure 4.2). The average natural co-contraction value during constant loading was 28% higher for older adults than for younger adults. Similar results were found for the destabilizing task (35%) and stabilizing task (34%). Older adults co-contracted at a significantly higher value than younger adults under each loading paradigm ($p < 0.005$).

Older adults were found to have lower positional variability when compared to younger adults performing the same task (Figure 4.3). The positional variability of older adults was 59% lower under the destabilizing loading paradigm, 58% lower under the constant loading paradigm, and 42% lower under the stabilizing loading paradigm. Each of these values was significant at $p < 0.005$.

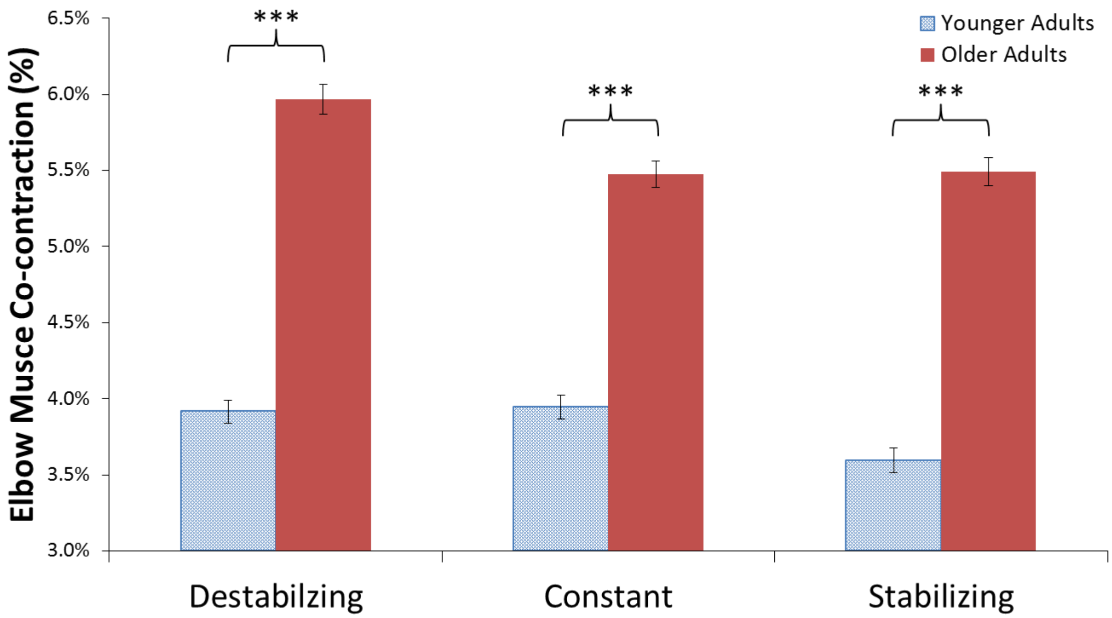


Figure 4.2: Bar graph showing results for elbow muscle co-contraction level (%) for the three different elbow moment loading paradigms. Older adults naturally co-contracted more than younger adults for each loading paradigm (*** indicates $p < 0.005$). Error bars represent the standard error for the least squares means of each loading paradigm.

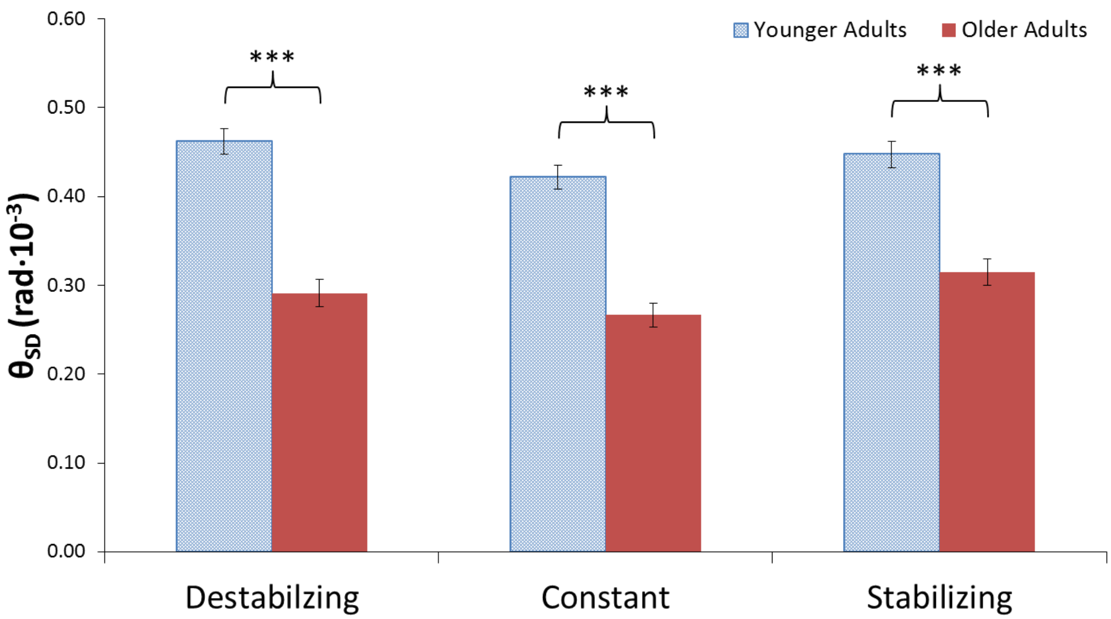


Figure 4.3: Bar graph showing the results for elbow angle positional variability under the three different elbow moment loading paradigms for both younger and older adults. Older adults exhibited lower levels of positional variability than younger adults for each loading paradigm (*** indicates $p < 0.005$). Error bars represent the standard error for the least squares means of each loading condition and age category.

The loading paradigm significantly affects the natural elbow muscle co-contraction level for older adults ($p < 0.005$, Table 4.2). Older adults naturally co-contracted 9% more ($p < 0.005$) during the destabilizing loading paradigm than for the constant loading paradigm (Figure 4.4A). Similarly, subjects co-contracted 9% more ($p < 0.01$) during the destabilizing loading paradigm than during the stabilizing loading paradigm. There was no significant difference between the co-contraction levels during the constant and stabilizing loading paradigms.

The loading paradigm also significantly affected elbow angle variability in older adults (Table 4.3). Older adults exhibited 15% higher positional variability ($p < 0.05$) during the stabilizing loading paradigm than during the constant loading paradigm (Figure 4.4B). Subjects trended towards exhibiting higher elbow angle positional variability during the destabilizing loading paradigm than during the constant loading paradigm. However, this difference was not statistically significant.

Table 4.2: The general linear model for elbow muscle co-contraction suggests that loading paradigm significantly affects elbow muscle co-contraction ($p < 0.005$).

Variable	**p-value**
Subject	0.00
Loading	0.00
Force	0.01
Trial	0.01
Subject*Force	0.00
Subject*Trial	0.02

Table 4.3: Significant factors for the general linear model of elbow angle positional variability. Loading paradigm significantly affected elbow angle positional variability ($p < 0.05$).

Variable	**p-value**
Subject	0.11
Loading	0.04
Force	0.60
Trial	0.31
Subject*Force	0.07
Subject*Trial	0.00

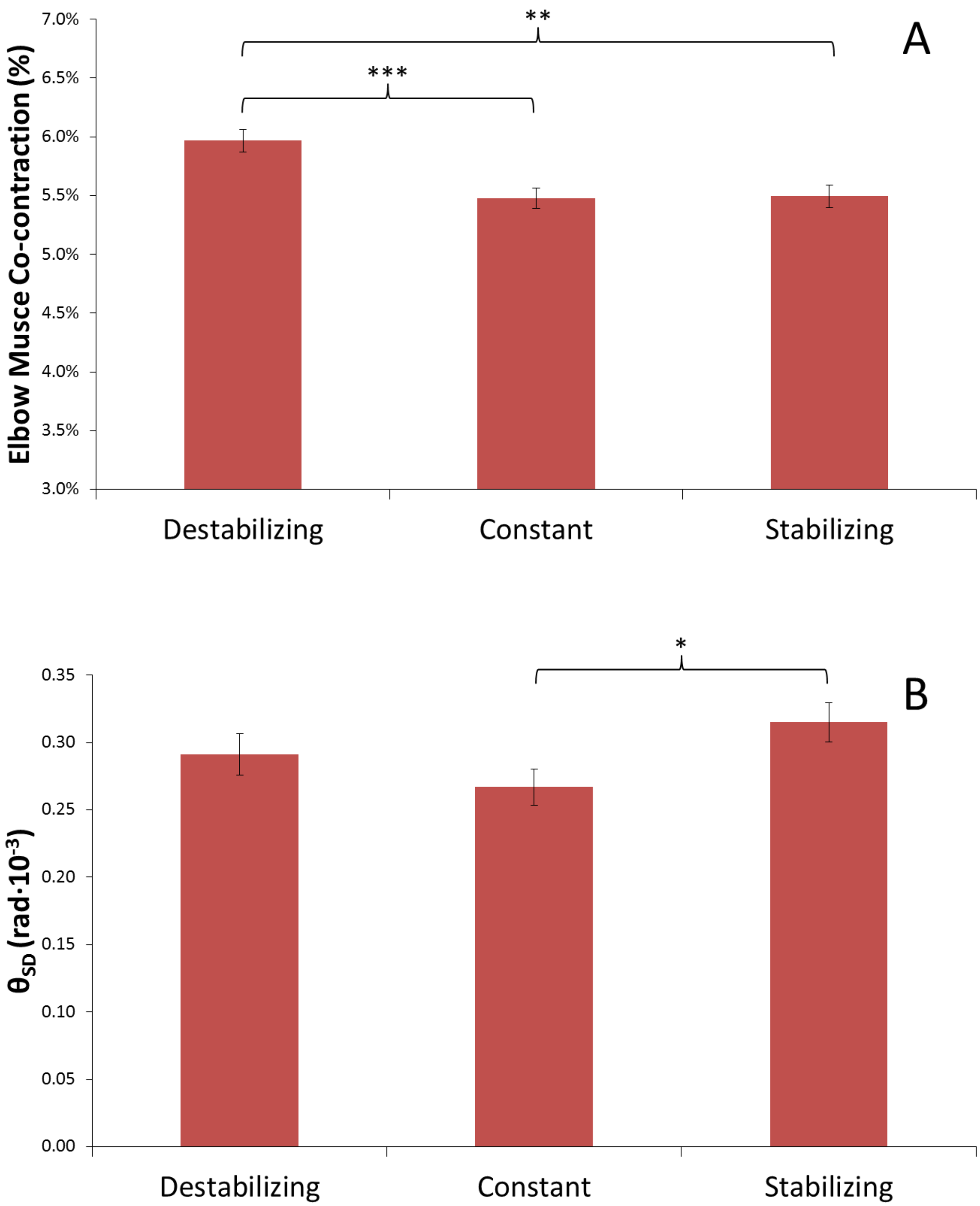


Figure 4.4: Bar graphs showing the results of A) elbow muscle co-contraction level and B) elbow angle positional variability for three different elbow moment loading paradigms. Error bars represent the standard error for the least squares means of each loading paradigm. * indicates $p < 0.05$, ** indicates $p < 0.01$, and *** indicates $p < 0.005$.

4.5 Discussion

The results supporting the primary hypothesis corroborate and extend the Variable-Stiffness SDN Theory (Chapter 2), which predicts that increasing co-contraction above natural levels will lead to higher positional variability. While this was shown for younger adults in Chapters 2-3, it had not been shown for older adults. The increasing positional variability with increasing co-contraction suggests that the current dogma, that higher co-contraction levels can be used to reduce positional variability, is not true for co-contraction above the natural level in older adults.

This study partially supports the secondary hypothesis $H_{4.2}$ by finding higher levels of co-contraction in older adults than younger adults. While this is not a new finding, we believe that this is the first study where the force level was matched to the subject's strength. This is essential because older adults have been shown to have considerably less muscle mass than younger adults (i.e., Tzankoff and Norris, 1977) and therefore higher muscle activation is often necessary for older adults to support the same load. Normalizing the external load to the subject's strength removed the effect of this strength difference from the study. This study also tests elbow loading under three different loading paradigms. This is significantly different from other studies that use one loading paradigm (Seidler-Dorbin et al., 1998; and Klein et al., 2001).

However, older adults had lower positional variability than younger adults for each loading paradigm, partially disproving $H_{4.2}$. It is unclear as to why this result disagrees with the results of previous studies (Ranganathan et al., 2001). Ranganathan et al. found that older adults were less steady when attempting to hold a pin in a hole without touching the sides of the hole. It is also possible that the current study produced different results because the task did not demand positional steadiness. Older adults may not be as capable as younger adults at reducing their level of positional steadiness with increased task demands. More research may yield insight into these seemingly conflicting results.

Subjects were asked to maintain a constant elbow moment once the retaining rod dropped. However, when some of the initial older adults repeated back the instructions they said they were to hold their arm still. When this happened, subjects were again instructed to not think about the position of their hand and arm, but rather concentrate on pulling with the same amount of force. The subject's understanding of the task to be performed was checked and clarified in the later subjects, regardless of age. While this task perception may have affected the results, it is not clear that the problem was limited to older adults.

The results from the different loading paradigms, shown in Figure 4.5, partially supported and partially refuted the predictions of the Variable-Stiffness SDN Theory. The destabilizing loading paradigm produced higher natural co-contraction values than both the constant and the stabilizing loading paradigms. This agrees with the predictions of the Variable-Stiffness SDN Theory. However, the positional variability during the different loading paradigms did not support the theory. Instead, the elbow angular positional variability during the stabilizing loading paradigm was significantly higher than during the constant loading paradigm.

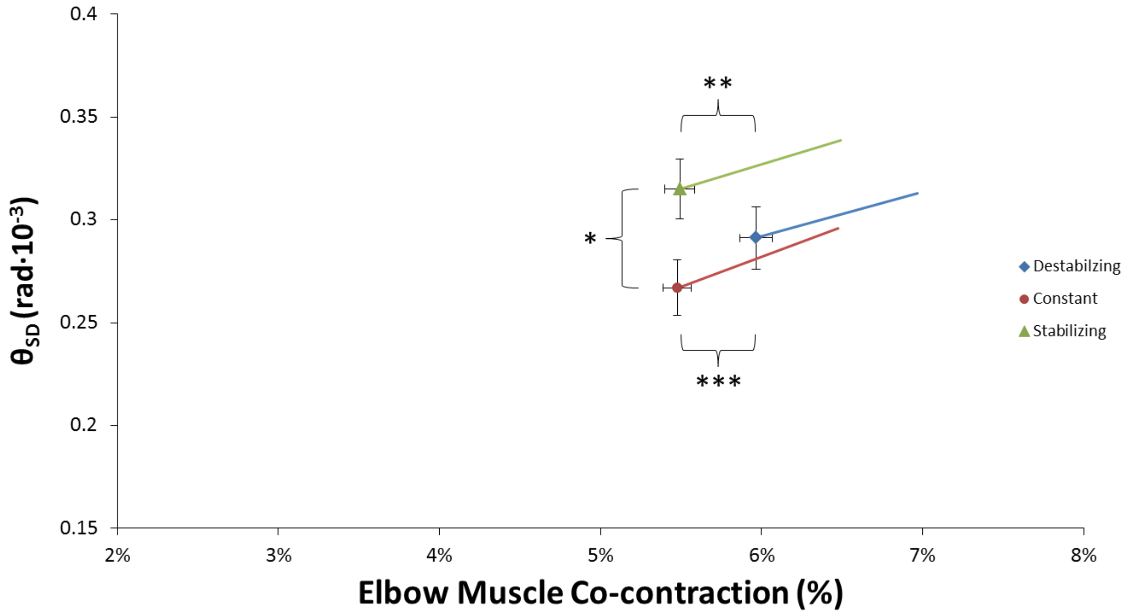


Figure 4.5: Summary of the experimental results showing the least squares means and corresponding standard errors for the natural elbow muscle co-contraction levels and elbow angle positional variability for the three different elbow moment loading paradigms. Co-contraction above the natural level increased positional variability in older adults as shown by the positive sloping lines ($p < 0.005$). Older adults co-contracted at higher levels during the destabilizing loading paradigms than during the stabilizing loading (** indicates $p < 0.01$) and the constant loading (*** indicates $p < 0.005$) paradigms. Older adults exhibited higher levels of positional variability during stabilizing loading than during constant loading (* indicates $p < 0.05$).

One possible explanation is that subjects were less familiar with the stabilizing loading paradigm. The subject's musculoskeletal control may not have taken advantage of the stabilizing behavior of the loading paradigm. This may be caused by a time lag between the subject's elbow moment and the changing external moment. When subjects generated an elbow moment that was greater than the external moment, their elbows flexed. The subjects may have reduced their elbow flexion moment to match the external extension moment. However, simultaneously, the elbow flexion would have caused the external moment to increase. This may have caused the subject's elbow flexion moment to be lower than the newly increased external extension moment causing the subject's elbow to extend. This positive feedback loop may have resulted in elbow angle variability. More research should be done to determine why a seemingly stabilizing loading paradigm resulted in higher levels of positional variability.

4.6 Conclusions

Elbow angular positional variability was found to increase with increased co-contraction above the natural level for healthy older adults. This supports the theory that increased co-contraction seen in older adults may be the source of positional variability rather than a compensatory effect. However, the higher co-contraction levels seen in healthy older adults resulted in lower positional variability in comparison to younger adults.

4.7 References

Burnett, R. A., Laidlaw, D. H., & Enoka, R. M. (2000). Coactivation of the antagonist muscle does not covary with steadiness in old adults. Journal of Applied Physiology, 89(1), 61.

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Galganski, M. E., Fuglevand, A. J., & Enoka, R. M. (1993). Reduced control of motor output in a human hand muscle of elderly subjects during submaximal contractions. Journal of Neurophysiology, 69(6), 2108-2115.

Gribble, P. L., Mullin, L. I., Cothros, N., & Mattar, A. (2003). Role of cocontraction in arm movement accuracy. Journal of Neurophysiology, 89(5), 2396-2405.

Hortobagyi, T., & Devita, P. (2006). Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. Exercise and Sport Sciences Reviews, 34(1), 29-35.

Jones, K. E., Hamilton, A. F. C., & Wolpert, D. M. (2002). Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology, 88(3), 1533.

Klein, C. S., Rice, C. L., & Marsh, G. D. (2001). Normalized force, activation, and coactivation in the arm muscles of young and old men. Journal of Applied Physiology, 91(3), 1341-1349.

Laursen, B., Jensen, B., & Sjøgaard, G. (1998). Effect of speed and precision demands on human shoulder muscle electromyography during a repetitive task. European Journal of Applied Physiology and Occupational Physiology,78(6), 544-548.

Lee, Y., & Ashton-Miller, J. (2011). The effects of gender, level of co-contraction, and initial angle on elbow extensor muscle stiffness and damping under a step increase in elbow flexion moment. Annals of Biomedical Engineering,39(10), 2542-2549.

Moritz, C. T., Barry, B. K., Pascoe, M. A., & Enoka, R. M. (2005). Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle. Journal of Neurophysiology, 93(5), 2449-2459.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. (2001). Effects of aging on hand function. Journal of the American Geriatrics Society, 49(11), 1478-1484.

Seidler-Dobrin, R. D., He, J., & Stelmach, G. E. (1998). Coactivation to reduce variability in the elderly. Motor Control, 2(4), 314-330.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieën, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Smith, C. D., Umberger, G. H., Manning, E. L., Slevin, J. T., Wekstein, D. R., Schmitt, F. A., et al. (1999). Critical decline in fine motor hand movements in human aging. Neurology, 53(7), 1458-1461.

Tzankoff, S. P., & Norris, A. H. (1977). Effect of muscle mass decrease on age-related BMR changes. Journal of Applied Physiology: Respiratory, Environmental and Exercise Physiology, 43(6), 1001-1006.

Chapter 5

Effect of Decreasing Co-contraction Below Normal Values on Elbow Angle Positional Variability

5.1 Abstract

Positional steadiness, the inverse of positional variability, is important for completing activities of daily living. The results from Chapters 3-4 demonstrate that increasing elbow muscle co-contraction above the natural level increases elbow angle positional variability in both younger and older adults. However, the effect of elbow muscle co-contraction on elbow angle positional variability is unknown for co-contraction levels below the natural level. It is difficult to study subject performance at low co-contraction levels because subjects are often unable to consciously reduce their co-contraction below the natural level. This study attempted to achieve lower than natural co-contraction levels by deceiving subjects.

Chapters 3-4 found that younger and older adults exhibit lower levels of co-contraction when resisting a stabilizing loading paradigm (where the external elbow extension moment decreases with increasing elbow flexion angle) than when resisting a destabilizing loading paradigm (where the external elbow extension moment increases with increasing elbow flexion angle). This finding was used to design a test to determine the effect of below-natural elbow muscle co-contraction levels on elbow angle positional variability. Subjects believed the stabilizing loading paradigm was going to be used but

instead the destabilizing loading paradigm was used. These trials were compared to trials where subjects knew a destabilizing loading paradigm was going to be used.

Subjects co-contracted 8.5% less when they believed that the trials were using the stabilizing loading paradigm than when they believed that the trials were going to use the destabilizing loading paradigm ($p < 0.005$). There was no significant difference in the level of elbow angle positional variability between the two sets of trials. The elbow angle variability was not significantly lower when subjects expected the stabilizing load, and therefore co-contracted at lower levels. We conclude that elbow muscle co-contraction can be affected by the subject's perception of the task. However, the effect of decreased co-contraction on positional variability was not determined.

5.2 Introduction

Elbow angle positional variability, the inverse of elbow angle steadiness, is essential in performing activities of daily living such as feeding oneself with a spoon, or inserting a key into a lock. Positioning the hand in the correct position is ineffectual if a person is unable to maintain that position for the time required to complete the task. Decreasing a person's elbow angle positional variability may enable him or her to complete these tasks of daily living.

One possible method of decreasing elbow angle positional variability is to control muscle activation, particularly muscle co-contraction level, during a task. One source of variability is the stochastic noise in muscle force (Jones et al., 2002 and Moritz et al., 2005). Increasing muscle activation, such as that caused by increased co-contraction, increases this stochastic noise in muscle force (Hamilton et al., 2004) which increases positional variability. However, increasing muscle activation also increases muscle stiffness (Sinkjaer et al., 1988 and Blanpied and Smidt, 1993) which attenuates positional variability. Because of these opposing features of increased muscle activation, it is unclear what effect co-contraction has on positional variability.

The effect of co-contraction on positional variability has been studied in subjects who were asked to co-contract at higher levels (Osu et al., 2004), in subjects performing tasks that naturally lead to higher co-contraction (Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006), and in simulations (Selen et al., 2005). Osu et al. (2004) measured lower trial-to-trial variability of the arm position at the end of the task when subjects were instructed to co-contract at higher levels. Subjects were asked to co-contract at three different levels and perform an elbow extension or flexion movement. Osu et al. demonstrated that the trial-to-trial positional variability was lower when subjects consciously co-contracted at higher levels. However, the positional variability measured in Osu et al. was not the variability during the task. It is unclear whether the variability of the hand during the movement was also lower when subjects were asked to increase their co-contraction level.

Osu et al. (2004) also investigated the effect of naturally higher co-contraction by changing the task's accuracy demand. Researchers discovered that subjects co-contracted more when the accuracy demand was increased. Subjects were also found to have lower trial-to-trial positional variability in these trials. This finding suggests that naturally higher co-contraction levels lead to lower endpoint positional variability. Higher co-contraction levels could lead to lower endpoint positional variability due to the increased muscle stiffness associated with increased muscle activation (Sinkjaer et al., 1988; Blanpied and Smidt, 1993; Selen et al., 2006; and Lee and Ashton-Miller, 2011). Gribble et al. (2003) produced similar results and also found that the trial-to-trial variability of the arm trajectory decreased with increased accuracy demand and increased co-contraction. However, it is still unclear whether this finding applies to positional variability during a task.

None of these three studies investigated co-contraction levels that are lower than the subject’s natural co-contraction level for a given task. The original Signal Dependent Noise (SDN) Theory states that natural muscle activation, and therefore the natural co-contraction level, minimizes positional variability. This suggests that increasing co-contraction above natural levels or reducing co-contraction below natural levels will not decrease positional variability.

A SDN Theory based simulation by Selen et al. (2005) suggests that co-contraction reduces positional variability during a task. Selen et al. simulated a single joint, two muscle system which modeled the joint angle over time. As with this current study, the positional variability was defined as the standard deviation of the joint angle over time. The simulation predicts that high amounts of co-contraction decrease positional variability. However, this finding is in opposition to the experimental results from Chapters 2-3 and the simulation results from Chapter 2.

The Variable-Stiffness SDN Theory based-model of the elbow joint described in Chapter 2 suggests that positional variability decreases with increasing co-contraction for low levels of co-contraction. However, for higher ranges of co-contraction, positional variability increases with increasing co-contraction. The latter portion of the theory was supported experimentally by the results from Chapters 2-4 suggesting that positional variability increases in healthy younger and older adults with increased co-contraction above their natural levels. However, these experiments did not investigate the effect of lowering a subject's co-contraction below the natural level, and therefore the model has not been validated for low levels of co-contraction.

The effect of aging on the relationship between co-contraction and positional variability is unknown for below-natural levels of co-contraction. Older adults have naturally higher co-contraction levels than younger adults (Hortobagyi and DeVita, 2006; Seidler-Dorbin et al., 1998; and Klein et al., 2001). Since elbow muscle co-contraction above the natural level increases elbow angle positional variability for both younger and older adults (Chapters 2-4), it is reasonable to expect lower positional variability in older adults as they reduce their co-contraction level to that naturally seen in younger adults. However, younger adults naturally co-contract at the optimal level to minimize positional variability; therefore, decreasing their co-contraction level may increase positional variability.

In this chapter our aim is to experimentally measure the positional variability of subjects co-contracting at values lower than their natural co-contraction levels. The results of Chapters 3-4 demonstrate that both younger and older adults naturally co-contract at a lower level during the stabilizing loading paradigm than during the destabilizing loading paradigm. Therefore, it is hypothesized that subjects will co-contract at lower than natural levels for the destabilizing loading paradigm if they are exposed to multiple trials using the stabilizing loading paradigm and then exposed to the destabilizing loading paradigm without being informed. By changing the loading paradigm without the subject's knowledge, it is hypothesized that subjects will exhibit the natural level of co-contraction for the stabilizing load while actually supporting the destabilizing load. This will allow us to test the main hypothesis presented in Chapter 1 as $H_{5.1}$: *Reduced levels of elbow muscle co-contraction increase elbow angle positional variability in younger and older adults for a destabilizing loading paradigm (an elbow extension moment that increases with decreasing elbow flexion angle as shown in Figure 5.1).*

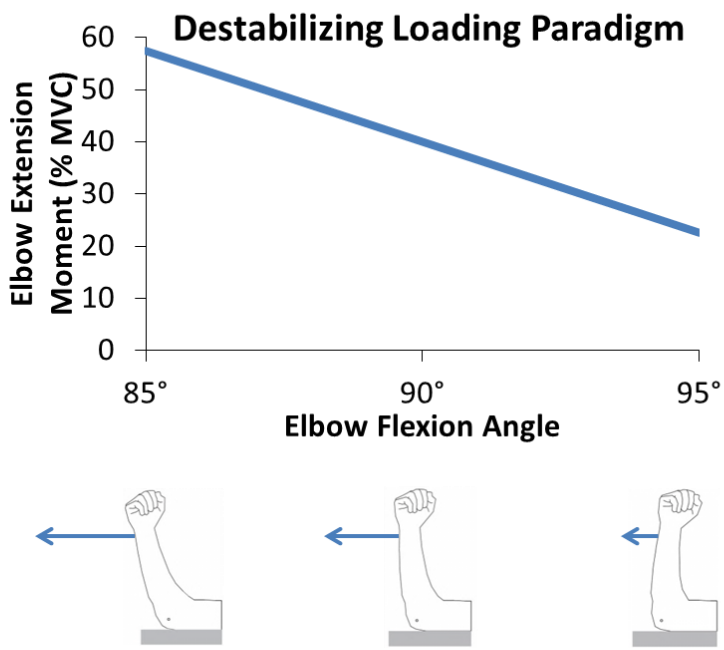


Figure 5.1: The change in external elbow extension moment with elbow flexion angle in the destabilizing loading paradigm.

5.3 Methods

5.3.1 Subjects

Fourteen healthy right handed younger adults (21-30 years old, 7 male and 7 female, mean age = 25 years) and 14 healthy older adults (60-80 years old, 6 male and 8 female, mean age = 68 years) were recruited to take part in this institutional review board-approved study and signed informed consent forms prior to participation. All subjects were screened for previous arm, shoulder, and neck injuries, and for chronic diseases.

5.3.2 EMG Data

Electromyographic (EMG) data were collected from each subject’s four main elbow flexor muscles (long head biceps, short head biceps, brachialis, and brachioradialis) and elbow extensor muscles (long head triceps, medial head triceps, lateral head triceps, and anconeous). The EMG data were collected using a Myosystem 2000 EMG system (Noraxon U.S.A. Inc., Scottsdale, Arizona) with blue dot style electrodes and 2-cm spacing along the muscle fibers. EMG data were collected at 1 kHz on the central computer.

5.3.3 Kinematic Data

The subject’s kinematic data were collected using a Certus Optotrak camera system (Northern Digital Inc., Waterloo, Canada). Infrared markers were placed on the medial side of the subject’s upper arm, elbow, and forearm. These markers were then used to calculate the elbow joint angle for the trials. The kinematic data were collected at 1 kHz on the central computer.

5.3.4 Force Data

Force data were collected using an axial force transducer (ATI Industrial Automation, Apex, North Carolina). The force transducer was attached between the subject’s wrist cuff and the loading cable. The signal from the force transducer was amplified and collected at 1 kHz on the central computer.

5.3.5 Experimental Apparatus

The experimental apparatus was a custom-built apparatus designed to load a subject's wrist with different loading paradigms. The load was generated by a set of seven constant force springs, ranging from 1 lb to 24 lbs, which were mounted to a wall. The springs were joined together using a metal ring which allowed the total spring force to range from 1 lb to 72 lbs. A metal retaining rod was inserted into the ring to hold the load while the subject increased their elbow flexion moment.

A cable connected the metal ring to the loading plate, which generated the destabilizing or stabilizing loading paradigms depending through which loading plate hole the cable was attached. The destabilizing loading paradigm increased the force on the subject's wrist as their elbow extended (Figure 5.1). The stabilizing loading paradigm decreased the external elbow extension moment as the elbow flexion angle increased. For further description on the experimental apparatus, see section 3.3.2.2.

5.3.6 Maximal Force Trials

The subject was seated at the apparatus with the upper arm resting on the table with a wrist cuff attached to the forearm. The chair was adjusted to elevate the shoulder slightly higher than the elbow. A cable attached the wrist cuff to a stationary structure. The subject was instructed to use only his or her elbow muscles to pull against the cable (i.e., not to lean back or twist the torso). The subject was instructed to increase the elbow flexion moment from relaxed to maximal over three seconds, hold their maximal voluntary force for two seconds, and then relax. The verbal cues were given as "One…two...three…hold…hold…relax." The maximal force was defined as the highest

0.25-second average of the trial. The trial was repeated until two maximal force levels were within 10% of each other. No subject repeated the trial more than five times.

5.3.7 Maximal EMG Trials

Maximal EMG values for the muscles were obtained using the EMG from the Maximal Force Trials and from additional trials targeting individual muscles. For the trials targeting individual muscles, the subject's force was resisted by the test administrator. Again, maximal EMG activation values for each muscle were defined as the highest mean activation over a 0.25-second period.

5.3.8 Resting EMG Trials

Subjects were asked to lie down on a mat for the resting EMG trials. Subjects laid down on their backs with their arms relaxed at their sides. Subjects were told to close their eyes and focus on relaxing their muscles. The test administrator then read a script instructing the subjects to relax specific muscle groups. Once the script was completed, two five-second trials were recorded. For each of the eight muscles, the lowest average EMG level over a 0.25-second period on either trial was recorded as the minimum EMG value for the muscle.

5.3.9 Normalizing EMG

The EMG data were normalized to generate the relative activation of each muscle. First, the minimum EMG value of each muscle was subtracted from the corresponding EMG data from the trials. Those values were then divided by the difference between the maximal and minimum EMG values of the muscle to generate the activation percentage.

5.3.10 Experimental Trials

Subjects practiced with both the destabilizing and stabilizing loading paradigms. As described in Chapter 3, the spring cable was preloaded and held in place with a retaining rod. Subjects gradually increased their elbow flexion moment until the retaining rod dropped. Once the retaining rod dropped, subjects were to maintain their current force level (Figure 5.2). Subjects were asked to increase their force at the same speed regardless of the loading paradigm. To ensure the beginning of each trial was identical for the subject regardless of the loading paradigm, the elbow moment level at which the retaining rod dropped was the same (40% of maximal) for each trial.

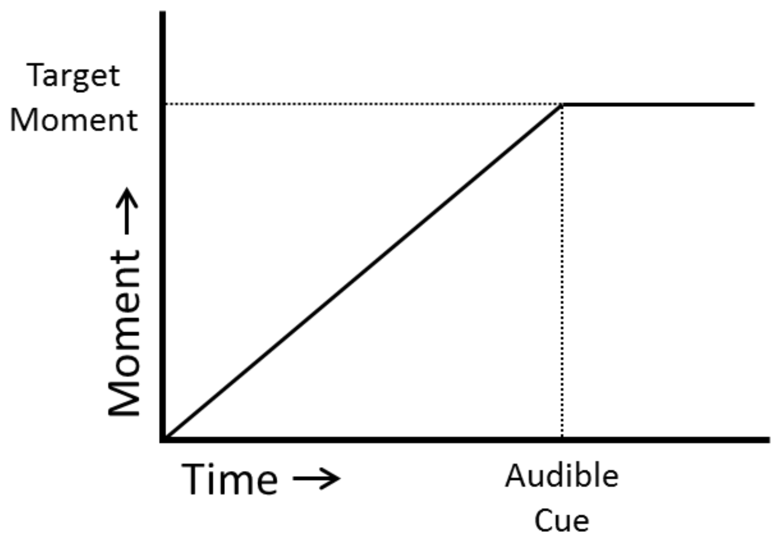


Figure 5.2 : Idealized loading for the beginning of each trial. Subjects increased their elbow flexion moments until they heard the audible cue from the retaining rod. Then, subjects were to hold that elbow flexion moment level for the remainder of the trial.

Before the trials began, a random-number generator was used to determine how many stabilizing trials the subject would be given before each destabilizing loading paradigm trial. The random-number generator was designed to start with two stabilizing loading paradigm trials. The following trial was given a 25% probability of being the destabilizing loading paradigm trial. A maximum of five stabilizing trial were allowed before the destabilizing trial was administered. Subjects were instructed that the first trial would use the stabilizing loading paradigm. Subjects were again allowed to practice the stabilizing loading paradigm before the trials started. Subjects were not told that destabilizing loading paradigm trials would be used. Therefore, subjects were expecting a stabilizing load during the destabilizing loading trial. Because of this, the destabilizing loading paradigm trials performed in this manner will be referred to as “deceived” trials. The set of trials were repeated four times, each time with a new random number.

The tests were as identical as possible regardless of the loading paradigm. Between each trial the test administrator walked behind the curtain and removed the loading cable from the loading plate. The loading cable was then reattached to the loading plate to create the appropriate loading paradigm. The same procedure was followed even if the loading paradigm was not changed.

Subjects also performed trials where they were correctly expecting the destabilizing loading paradigm. The protocol for those trials is described in Chapter 3. These trials will now be referred to as “expected” trials.

5.3.11 Statistical Analysis

Minitab (State College, Pennsylvania) was used to generated a general linear model of the effect of task deception on elbow muscle co-contraction and elbow positional variability. The subjects were treated as random factors to remove the subject-to-subject variability from the data. The least squared means were also calculated for the deceived and expected trials, and the difference in means were tested using Tukey posthoc analysis.

5.4 Results

The results of the secondary hypothesis will be presented first due to its implications on the primary hypothesis. Subjects' co-contraction levels during the destabilizing loading paradigm were significantly lower for the deceived trials than the expected trials (Figure 5.3, $p < 0.005$). This supports the secondary hypothesis $H_{5.2}$: *Elbow muscle co-contraction will be lower than natural levels for a destabilizing loading paradigm when subjects expect a stabilizing loading paradigm.* While older adults had higher co-contraction values, the effect of the deceiving protocol did was not affected by age. The data scatter showing each subject's average elbow muscle co-contraction level and elbow angle positional variability for the two trial types are shown in Figure 5.4 with younger adults represented with the blue markers and older adults represented with the red markers.

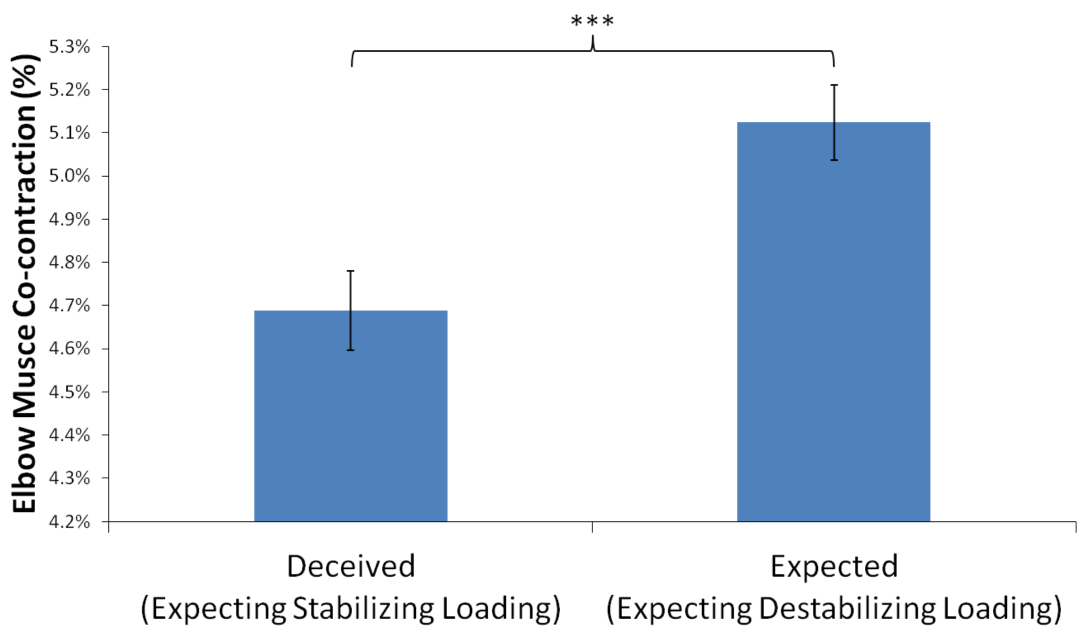


Figure 5.3: Bar graph showing results for elbow muscle co-contraction for the two trial types. Subjects co-contract less when they are expecting a stabilizing load than when they are expecting a destabilizing load (*** indicates $p < 0.005$). Error bars represent the standard error for the least squares means of the trial types.

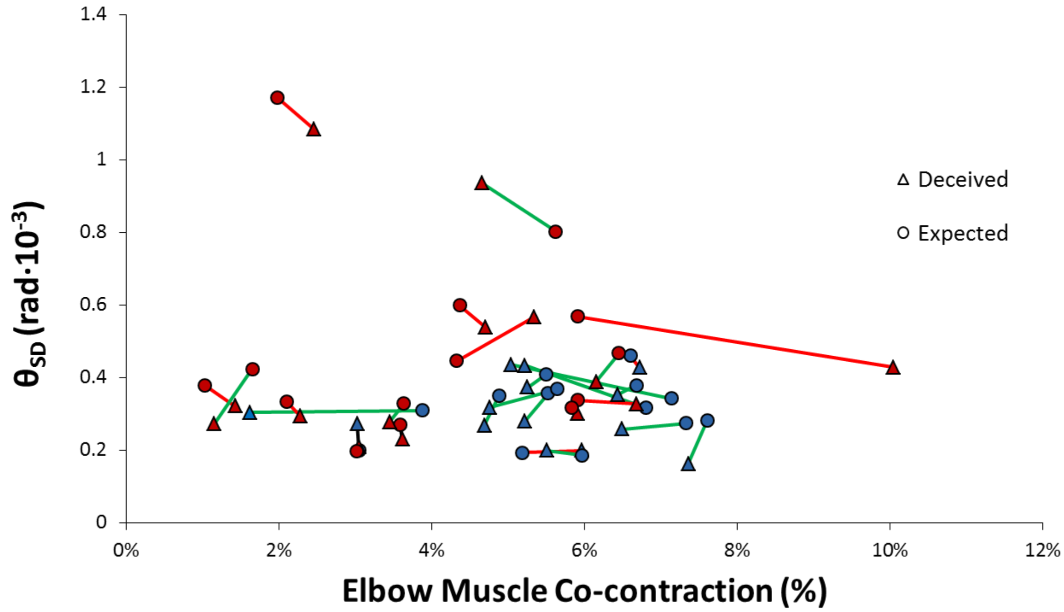


Figure 5.4: The data scatter showing the average elbow muscle co-contraction level and elbow angle positional variability for each subject when they were expecting the destabilizing paradigm (represented by circles) and when they were expecting the stabilizing paradigm (represented by triangles). Green lines indicate subjects whose co-contraction was lower during the deceived trials than during the expected trials while red indicate the opposite. Younger adults are represented by the blue markers and older adults are represented by the red markers.

The elbow angle positional variability (θ_{SD}) during deceived trials, during which the elbow muscle co-contraction level was lower, was not significantly lower than during the expected trials (Figure 5.5). This fails to prove the primary hypothesis: $H_{5.1}$: *Reduced levels of elbow muscle co-contraction will increase elbow angle positional variability in younger and older adults for the destabilizing loading paradigm.*

5.5 Discussion

The secondary hypothesis was supported, suggesting that subjects can co-contract at below-natural levels when performing a task (Figure 5.6). The results also suggest that muscle activation is determined at a higher level in the central nervous system (i.e., not the peripheral nervous system). If the muscle activation level for a task were based solely

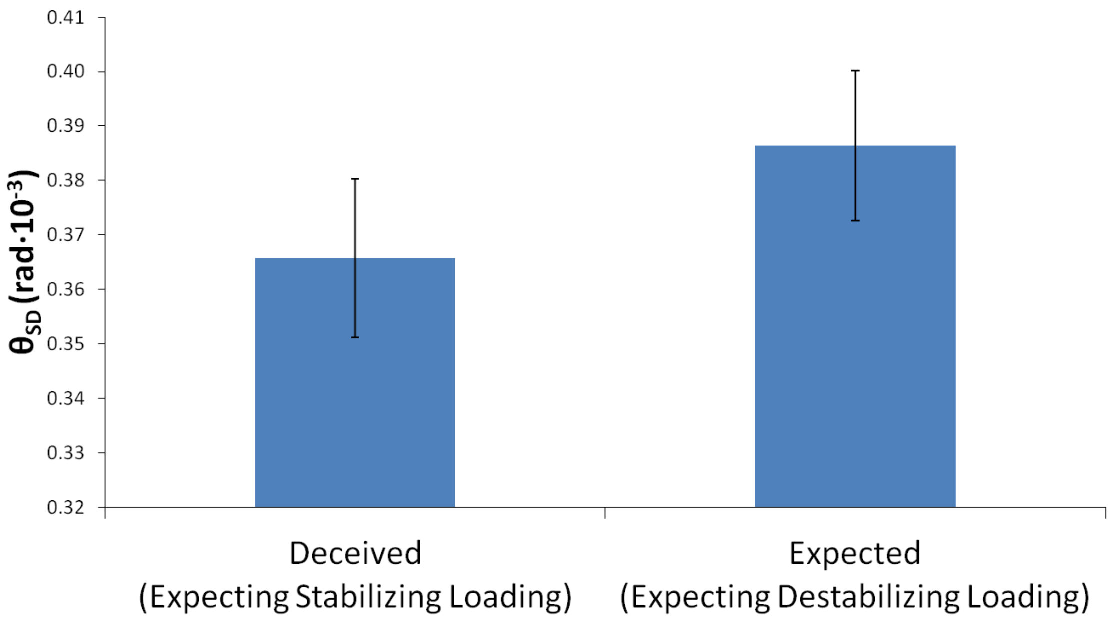


Figure 5.5: Bar graph showing results for elbow angle positional variability for the two trial types. Elbow angle variability was not significantly lower when subjects were not expecting a destabilizing load ($p > 0.05$). Error bars represent the standard error for the least squares means of each trial type.

on what the arm experiences, the co-contraction level would not be affected by the expected loading paradigm. While this study was not designed to be a robust test of where muscle activation planning originates, the results suggest that at least the co-contraction level for the muscles can be affected by the subject's perception of the task.

The primary focus of this study was to investigate the effect of lowering elbow muscle co-contraction levels below the natural level on elbow angle positional variability while performing a quasistatic task. While subjects did exhibit statistically significant lower than natural co-contraction levels, the difference in co-contraction levels was less than 10% of the natural co-contraction level. This small difference could be the reason for the small and statistically insignificant difference in positional variability between the two co-contraction levels. It may not be possible for subjects to co-contract at levels that are appreciably lower than natural co-contraction levels.

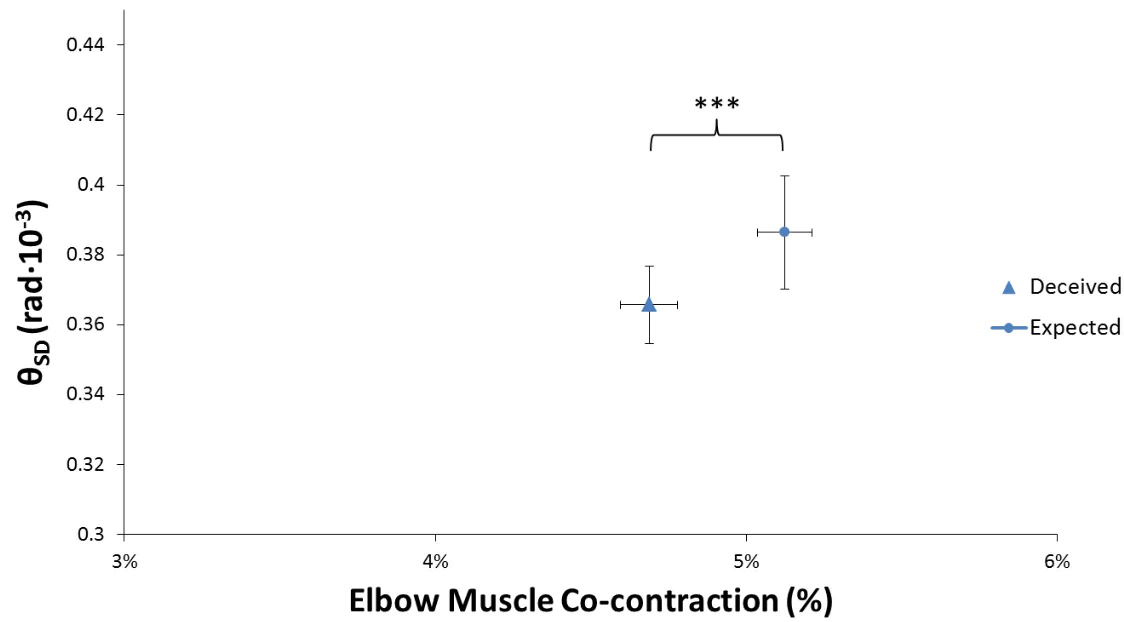


Figure 5.6: Summary of the experimental results showing the least squares means and corresponding standard errors for the elbow muscle co-contraction levels and elbow angle positional variability for the two different trial types. Subjects co-contracted at a lower level when they were deceived into expecting the stabilizing loading paradigm than when they were expecting the destabilizing loading paradigm (*** indicates $p < 0.005$), supporting $H_{5.2}$. However, the decrease in elbow angle positional variability when subjects were deceived into expecting the stabilizing loading paradigm was not significant.

The Variable-Stiffness SDN Theory model was not bolstered by this study. The theory suggests that any co-contraction level that is lower than natural will produce positional variability that is greater than the variability during natural muscle activation. Since the decrease in positional variability in this study was not statistically significant, this theory is not refuted by this experiment. The finding of no significant change in positional variability for a small decrease in co-contraction may be explained by the shape of the postural variability versus co-contraction level curve predicted by the Variable-Stiffness SDN Theory simulation described in Chapter 3 (Figure 3.5). The curve is very shallow near the optimal co-contraction level. Therefore, the change in postural variability for a small decrease in co-contraction level is also very small. This may explain the lack of evidence to support the Variable-Stiffness SDN Theory.

One limitation of this study is that subjects may have been anticipating which trials were going to be destabilizing loading trials. If a subject was able to determine when the destabilizing loading paradigm would occur, then the subject was not deceived. The expected result of these anticipated deceived trials would be the same as the expected trials and therefore they would decrease the difference in co-contraction levels. Because of this, care was taken to not allow the subjects to know when the loading paradigm would change. However, if a subject performed five stabilizing loading paradigm trials, the sixth trial was always a destabilizing loading paradigm trial. This was done to limit the number of trials a subject would be asked to perform. The probability that a subject would perform a sixth trial was less than 2% and therefore this is not expected to be a large source of error.

5.6 Conclusions

Subjects co-contracted below their natural level for the destabilizing loading paradigm when they were anticipating a stabilizing loading paradigm. While the difference in co-contraction levels was significant, the difference was small, which may have contributed to the lack of difference in positional variability. This neither supports nor refutes the claims of Variable-Stiffness SDN Theory. The findings also yield no insights as to the relation between below-natural co-contraction levels and positional variability.

5.7 References

Blanpied, P., & Smidt, G. L. (1993). The difference in stiffness of the active plantarflexors between young and elderly human females. Journal of Gerontology, 48(2), M58.

Gribble, P. L., Mullin, L. I., Cothros, N., & Mattar, A. (2003). Role of cocontraction in arm movement accuracy. Journal of Neurophysiology, 89(5), 2396-2405.

Hamilton, A. F., Jones, K. E., & Wolpert, D. M. (2004). The scaling of motor noise with muscle strength and motor unit number in humans. Experimental Brain Research., 157(4), 417-430.

Hortobagyi, T., & Devita, P. (2006). Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. Exercise and Sport Sciences Reviews, 34(1), 29-35.

Jones, K. E., Hamilton, A. F. C., & Wolpert, D. M. (2002). Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology, 88(3), 1533.

Klein, C. S., Rice, C. L., & Marsh, G. D. (2001). Normalized force, activation, and coactivation in the arm muscles of young and old men. Journal of Applied Physiology, 91(3), 1341-1349.

Lee, Y., & Ashton-Miller, J. (2011). The effects of gender, level of co-contraction, and initial angle on elbow extensor muscle stiffness and damping under a step increase in elbow flexion moment. Annals of Biomedical Engineering,39(10), 2542-2549.

Moritz, C. T., Barry, B. K., Pascoe, M. A., & Enoka, R. M. (2005). Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle. Journal of Neurophysiology, 93(5), 2449-2459.

Osu, R., Kamimura, N., Iwasaki, H., Nakano, E., Harris, C. M., Wada, Y., & Kawato, M. (2004). Optimal impedance control for task achievement in the presence of signal-dependent noise. Journal of Neurophysiology, 92(2), 1199-1215.

Seidler-Dobrin, R. D., He, J., & Stelmach, G. E. (1998). Coactivation to reduce variability in the elderly. Motor Control, 2(4), 314-330.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2005). Can co-activation reduce kinematic variability? A simulation study. Biological Cybernetics, 93(5), 373-381.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieën, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Sinkjaer, T., Toft, E., Andreassen, S., & Hornemann, B. C. (1988). Muscle stiffness in human ankle dorsiflexors: Intrinsic and reflex components. Journal of Neurophysiology, 60(3), 1110-1121.

Chapter 6

Effect of Aging and Target Size on Elbow Muscle Co-contraction Level and Elbow Angle Positional Variability during a Pouring Task

6.1 Abstract

The effect of elbow muscle co-contraction on elbow angle positional variability and pouring accuracy for a water pouring task is unknown. Some studies suggest that increasing co-contraction improves performance in tasks requiring low positional variability while others suggest that it hinders the task. This study investigated the effect of pouring target size on elbow muscle co-contraction level and elbow angle positional variability. The effect of elbow muscle co-contraction on elbow angle positional variability and pouring accuracy was also studied.

Twelve healthy younger adults (mean age = 25 years) and 14 older adults (mean age = 68 years) were asked to pour into four different sized circular openings (ø = 8 mm, 16 mm, 30 mm, and 102 mm) at a consistent rate and from a consistent height. The elbow muscle co-contraction level, elbow angle positional variability, and pouring accuracy were recorded. Subjects were found to co-contract at higher levels as the opening size decreased ($p < 0.05$). Subjects also were found to have lower elbow angle positional variability for the smallest opening size (ø = 8 mm) than for the two medium sized openings (ø = 16 mm and 30 mm, $p < 0.05$). Increasing elbow muscle co-contraction was found to increase elbow angle positional variability ($p < 0.05$) and decrease pouring

accuracy ($p < 0.001$) within a target opening size. We conclude that the increased co-contraction, naturally seen as task demands increase, inhibits rather than enhances performance.

6.2 Introduction

Many older adults struggle with activities of daily living that require low hand positional variability, the inverse of positional stability. This difficulty may result from the increased positional variability that older adults exhibit when trying to maintain a precise hand posture (Ranganathan et al., 2001). Subjects were asked to hold a 1 mm pin in a 2 mm hole and the number of times the pin touched the edge was recorded. Older adults touched the edge six times more often than younger adults, suggesting that older adults have greater difficulty controlling their hand position when presented with a positionally demanding task. Knowing the source of the increased positional variability in older adults may help occupational therapists assist older adults in completing these activities of daily living.

One source of hand positional variability is the variation in muscle contractile force due to the stochastic noise in the muscle activation (Galganski et al., 1993; Jones et al., 2002; and Moritz et al., 2005). This variation in contractile force scales with the activation level (Graves et al., 2000). Elbow muscle co-contraction increases the activation level of the muscle and therefore leads to increased elbow angle positional variability as shown in Chapters 2-4. It is possible that older adults are co-contracting at a higher than optimal level, resulting in increased positional variability.

Older adults naturally co-contract at higher levels than younger adults (Hortobagyi and DeVita, 2006; Seidler-Dorbin et al., 1998; and Klein et al., 2001).

Chapter 4 also demonstrated that older adults have higher natural elbow muscle co-contraction levels than younger adults when resisting an elbow extension load normalized to subject strength. This increased co-contraction may be the source of the increased positional variability seen in older adults.

However, the results from Chapter 4 demonstrated that older adults do not have higher positional variability when resisting an elbow extension moment. One possible explanation is that subjects were not asked to maintain a defined position during the trials. This created a task that allowed for higher arm positional variability. Both younger and older adults may perform differently during a task that requires low positional variability.

In Chapter 4, subjects were asked to increase their co-contraction level during the trials to determine the effect of elbow muscle co-contraction on elbow angle positional variability. Directing the subjects' attention to co-contraction rather than the task at hand may have affected the results of the study. Testing the effect of naturally increased co-contraction on positional variability during a task requiring positional accuracy may yield insight into the effect of co-contraction on positional variability during the desired tasks. This would help determine whether the increased co-contraction seen in older adults reduces positional variability or if it is a source of increased positional variability.

Gribble et al. (2003) found that subjects co-contract more when attempting to reach smaller targets. This has been used to suggest that subjects can increase co-contraction in order to perform better at tasks requiring low positional variability (Faisal et al., 2008). However, this supposition is not supported by the results of Chapters 2-4 where increased subject co-contraction above the natural level led to increased positional variability in both younger and older adults. Another possible explanation for the increased co-contraction seen in Gribble et al. (2003) is that the increased task difficulty causes subjects to increase their co-contraction. This increased co-contraction could be used to reduce the effect of any external forces that might affect the motion, but it may also increase positional variability caused by increased muscle activation. Measuring the co-contraction levels and positional variability while subjects perform positionally demanding tasks of different degrees of difficulty will allow us to better understand the effect of natural co-contraction level on positional variability.

The primary hypothesis for this study is stated in Chapter 1 as $H_{6.1}$: *Increased elbow muscle co-contraction will result in increased elbow angle positional variability during a task demanding positional accuracy (pouring water)*. These findings will demonstrate whether the increase in a subject's co-contraction level increases the positional variability or whether it does in fact decrease the positional variability. The secondary hypothesis was presented in Chapter 1 as $H_{6.2}$: *Elbow angle positional variability and elbow muscle co-contraction will increase as the accuracy demand increases (size of the container opening decreases)*. This will test the subject's natural muscle activation strategy and determine if it correctly decreases positional variability when the subject performs a more positionally demanding task or whether the activation strategy is a hindrance to accomplishing the task. Lastly, we will test hypothesis presented in Chapter 1 as $H_{6.3}$: *Elbow muscle co-contraction and elbow angle positional variability will be higher for older adults than younger adults for an accuracy demanding task regardless of the accuracy level demanded*. These results will give us a better understanding of the difficulties that older adults have when performing these positionally demanding tasks.

6.3 Methods

6.3.1 Subjects

Twelve healthy younger adults (21-30 years of age, 5 male and 7 female, mean age = 25 years) and 14 healthy older adults (60-80 years of age, seven male and seven female, mean age = 68 years) were recruited for this institutional review board-approved study. Subjects were screened for previous injury to their dominant hand, arm, or shoulder; joint problems such as arthritis or sprains; nerve injury; and chronic conditions. Subjects signed an informed consent form before participating in the study.

6.3.2 Equipment

Subjects poured two-thirds of a cup of water from a generic tea pot. The water was poured into a generic container through various sized circular openings. Those openings were created by placing an inverted funnel over the opening of the generic container to restrict the opening size. The funnels created three different sized openings of 8 mm, 16 mm, and 30 mm in diameter. For the fourth target size, no funnel was placed on top of the generic container which left the original opening size of 102 mm in diameter. For each trial, the generic container was placed in a large Pyrex® glass bowl to collect the water that missed the target opening.

6.3.3 Subject Instructions

Subjects poured from a standing position using their natural grip and arm position. Subjects were also asked to pour from an approximate pour height of five inches (shown as h in Figure 6.1). A height gauge was used at the beginning of each trial to measure the proper pour height. This gauge was removed before the trial to not interfere with the task. Subjects were instructed to pour the water into the target over a five second period. The test administrator verbally counted the seconds during the trial to help the subject gauge the correct pouring speed. The consistent pouring height and speed specifications ensured that the trials were similar independent of the subject and target size.

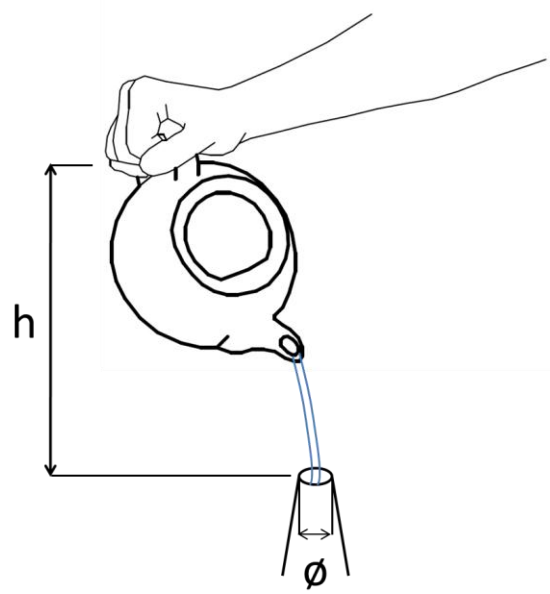


Figure 6.1: Subjects poured water from a teapot into a circular target opening. Subjects poured from a prescribed height (h = 5 in) into four targets of varied diameters (ø = 8 mm, 16mm, 30mm, and 102 mm).

6.3.4 Pouring Accuracy

The pouring accuracy was defined as the amount of water poured into the target as a percentage of total water poured. The amount of water in the target container and in the glass bowl was weighed using an OXO kitchen scale. The dry weights of the target container and glass bowl were subtracted from the measurements to determine the amount of water in each container.

6.3.5 Kinematic Data

The kinematics of the subject's arm and the teapot were recorded during the trial using an Optotrak Certus camera (Northern Digital Inc., Waterloo, Canada). Three infrared emitting markers were affixed to the lateral side of the subject's upper arm, elbow, and forearm using double-sided stickers. These markers were used to calculate the joint angle during the pouring task.

A triad of markers was also affixed to the teapot. The triad was used to calculate the position and orientation of the teapot during the trials. The center of the opening was digitized to permit the calculation of the opening position during the trial without obstructing the water flow. All kinematic data were collected at 500Hz and recorded on a central computer.

The elbow angle and spout distance from the center were filtered using a third-ordered Butterworth high pass filter at 4 Hz. This was done to minimize the effect of the positional drift that occurred during the pouring process as the subject moved or tilted the pot forward due to the decrease in water flow from the pot.

6.3.6 EMG Data

Muscle activation levels were collected using a Myosystem 2000 EMG system (Noraxon U.S.A. Inc., Scottsdale, Arizona). Blue dot sensors were attached to the subject's eight major elbow flexor muscles (long head biceps, short head biceps, brachioradialis, and brachialis) and extensor muscles (long head triceps, short head triceps, lateral head triceps, and anconeous) with 2-cm spacing along the muscle fiber direction.

6.3.7 Maximal EMG Trials

Subjects completed maximal effort trials and resting trials to establish minimum and maximal EMG values. For the maximal trials, subjects were seated with their elbows and upper arms resting on a table. Their elbows were flexed to 90° with their forearms held vertically. A wrist cuff was placed around each subject's wrist with a cable that was attached to a static structure. The subjects were instructed to first increase their force level over three seconds from resting to their maximal elbow flexion force, then hold their maximal force for two seconds, and finally relax. The subjects were given a verbal count during the trial of "1...2…3…hold…hold…relax." To isolate the extensor muscles, subjects were asked to push while the test administrator resisted their force. The maximal muscle activation level was defined for each muscle as the highest mean muscle EMG activation over a 0.25-second period.

6.3.8 Minimal EMG Trials

To obtain the minimum EMG levels for each subject, subjects were asked to lie down on their backs with their arms relaxed at their sides. The subject's EMG was recorded after a script was read instructing the subject to relax various muscle groups. The minimum muscle activation level was defined for each muscle as the lowest average EMG activation over 0.25-second period.

6.3.9 EMG Normalization

The EMG data from the trials was scaled to reflect the muscle activation percentage. The activation percentage was calculated by subtracting the minimal activation values from the muscle activation and then dividing by the difference between the maximal and minimal activation values for each muscle. This accounted for the differences in EMG signal levels for the subjects.

6.3.10 Co-contraction

Co-contraction was calculated as a percentage of the maximal elbow extensor moment. Physiological data were used to determine the relative amount of elbow extension moment that each extensor muscle could produce (Murray et al., 2000). The activation percentage for each muscle was then multiplied by the maximal elbow extensor moment for the muscle. The extensor moments were then added together and divided by the maximum possible extensor moment:

$$Co - contraction\ \% = \frac{\sum_{n=1}^{4} MuscleActivation\%_n \times MaxExtensorMoment_n}{\sum_{n=1}^{4} MaxExtensorMoment_n} \cdot 100$$

This generated a co-contraction percentage that was between 0% and 100%. While other methods of calculating co-contraction have been used (e.g. Wolstad, 1989; Osu et al., 2004; and Selen et al., 2005), this current study investigates the relative amounts of co-contraction and therefore the method of calculating co-contraction does not significantly impact the results.

6.3.11 Statistical Analysis

Data analysis was performed using Minitab (State College, Pennsylvania) statistical software. A general linear model was constructed to determine the effect of target size on elbow muscle co-contraction level while eliminating the subject to subject variability by using the subject as a random effect. This was used to create a model: $Co - contracion = \beta_0 + \beta_1 x_1$ where β_0 is a constant, $\beta_1$is the coefficient for target size, and x_1 is the target size.

However, the positional variability data did not vary linearly with respect to the target size. Because of this, the elbow angle variability data were analyzed using the target size as a categorical variable. This analysis allowed the least squares means of the positional variability to be compared between target sizes. A Tukey post-hoc analysis was done to determine whether the differences between target sizes were significant. This analysis was also applied to the elbow muscle co-contraction level to compare the differences between target sizes on co-contraction.

6.4 Results

The data support the primary hypothesis $H_{6.1}$. Co-contraction was found to be a significant factor in predicting elbow angle variability within a target size (

Table 6.1). Increased co-contraction increased positional variability (θ_{SD}) when blocking on subjects and target level to remove the difference between subjects and targets ($p < 0.05$).

Table 6.1: General linear model for positional variability suggesting that target size and co-contraction have a significant effect.

Variable	p-value
Subject	0.000
CC	0.011
Target	0.038
Subject*Target	0.000

Elbow muscle co-contraction was found to have a significant effect on success rate independent of subject and target size (Table 6.2, $p < 0.001$). This indicates that as the subject's elbow muscle co-contraction increases, the subject's ability to pour into the target decreases. The data scatter is shown in Figure 6.2 for all of the subjects pouring into the smallest target. Of the 27 subjects, 21 exhibited lowever levels of success with increased elbow muscle co-contraction levels. Similar results were found for the other three target sizes.

Table 6.2: General linear model for pouring accuracy. When including subject as a random effect and target as a covariate, elbow muscle co-contraction has a significant effect on pour success rate ($p = 0.001$).

Variable	**p-value**	**Coefficient**	**SE Coefficient**
Co-contraction	0.001	95.7	1.6
Subject	0.000		
Target	0.000		
Subject*Target	0.000		

Target size was also found to be a significant factor in predicting the elbow muscle co-contraction level (Table 6.3, $p < 0.001$). Subjects co-contracted 17% more when pouring into the smallest target than when they poured into the largest target (Figure 6.3).

Table 6.3: General linear model for elbow muscle co-contraction suggesting pouring target size has a significant effect.

Variable	**p-value**
Subject	0.000
Target	0.000

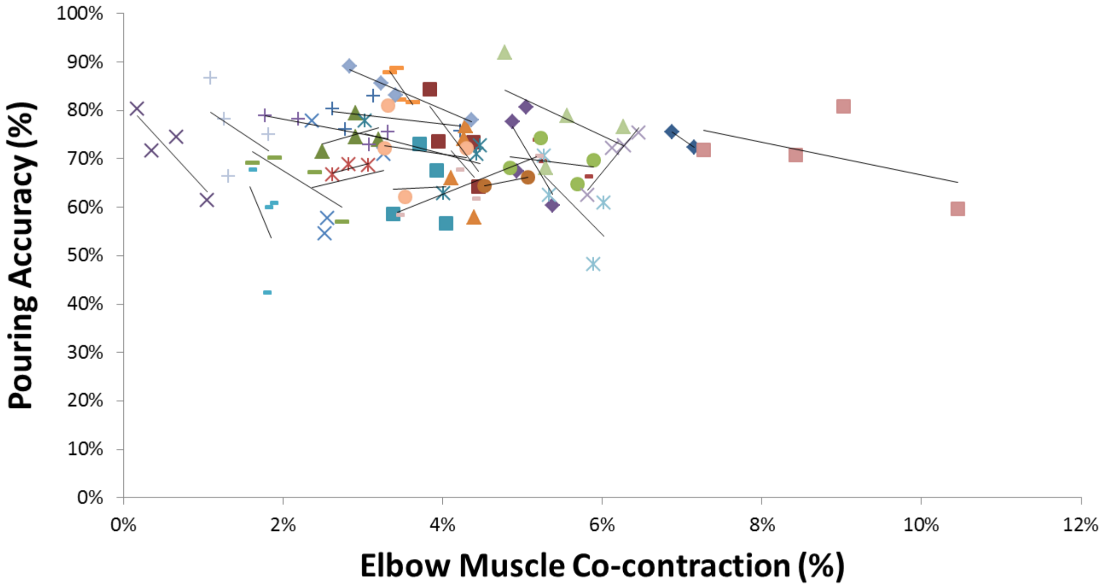


Figure 6.2: Data scatter for subjects showing the effect of co-contraction on pouring success. Of the 27 subjects, 21 had negatively sloped trend lines, indicating that increasing co-contraction decreased their ability to pour water through the target opening.

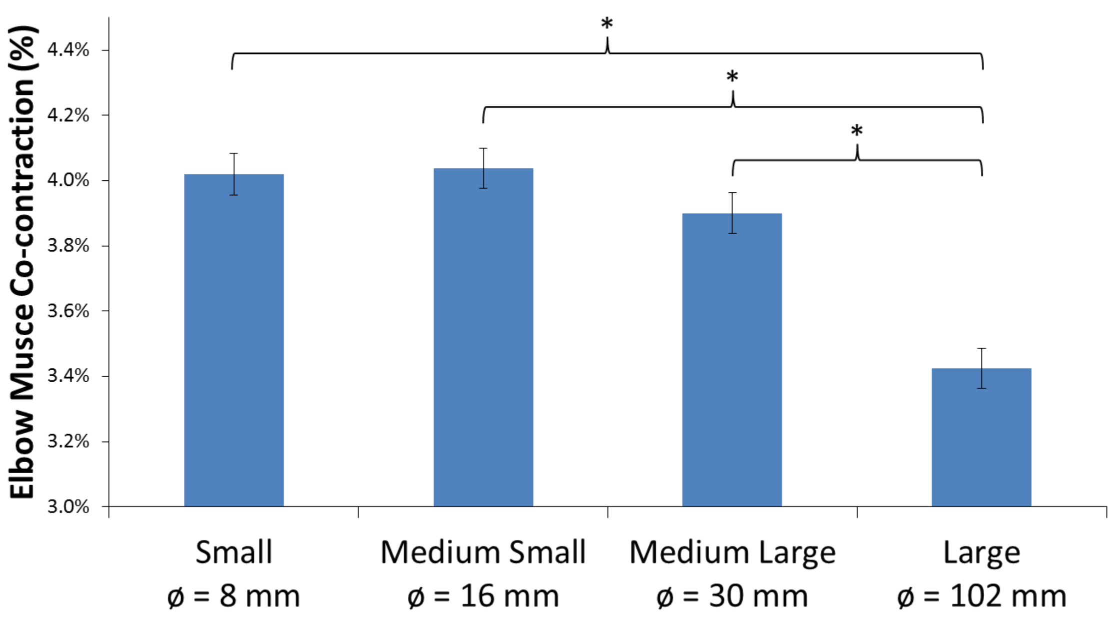


Figure 6.3: Bar graph showing results for elbow muscle co-contraction level for the four different pouring target diameters. The elbow muscle co-contraction level when pouring into the largest container was significantly lower than when pouring into the other three containers (* indicates $p < 0.05$). Error bars represent the standard error for the least squares means of each loading paradigm.

The elbow angle positional variability did not vary monotonically with target size. The variability increased with decreasing target size for the largest three target sizes (Figure 6.4). However, the elbow angle positional variability decreased when subjects poured into the smallest target. Consequently, the elbow angle variability analysis was performed with the target size as a categorical variable. Subjects exhibited significantly lower elbow angle positional variability when pouring into the smallest target in comparison to the two medium sized targets ($p < 0.05$).

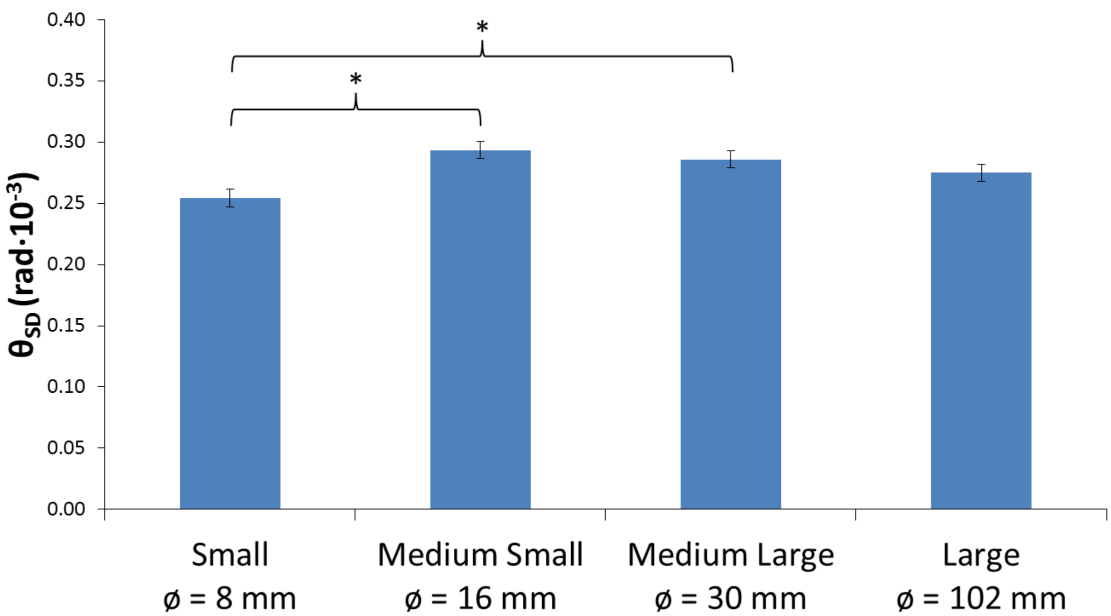


Figure 6.4: Bar graph showing results for elbow angle positional variability for the four different pouring target sizes. The variability was lower when subjects poured through the smallest target trials than in trials for the next two largest targets (* indicates $p < 0.05$). Error bars represent the standard error for the least squares means of each pouring target size.

Older adults co-contracted significantly more while pouring than younger adults (Figure 6.5, $p < 0.005$). Older adults co-contracted 20% more on average than younger adults. Older adults were also found to have lower elbow angle positional variability for the two largest pouring target sizes (Figure 6.6, $p < 0.05$) and overall (Figure 6.7, p < 0.005). Similar, yet less significant results were found for the spout variability. The success rate was not significantly different between younger adults and older adults.

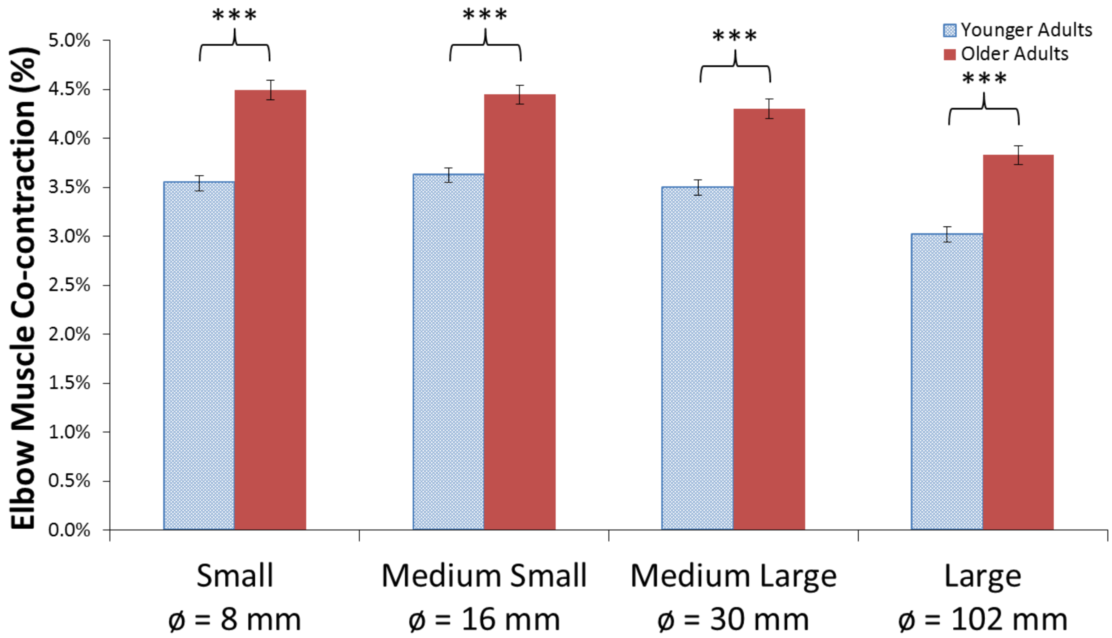


Figure 6.5: Bar graph showing elbow muscle co-contraction level for the four different pouring target sizes. Older adults co-contracted higher than younger adults when pouring water into different sized targets (*** indicates p < 0.005). Error bars represent the standard error for the least squares means of each loading paradigm.

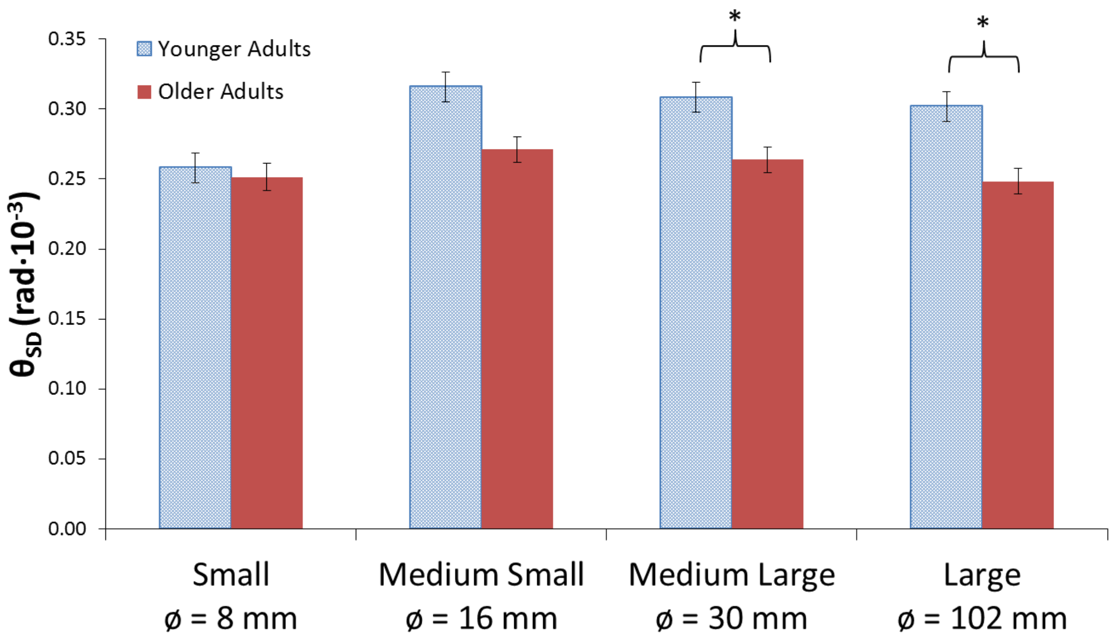


Figure 6.6: Bar graph showing elbow muscle positional variability for the four different pouring target sizes. Older adults co-contracted higher than younger adults when pouring water into the two largest targets (* indicates $p < 0.05$). Error bars represent the standard error for the least squares means of each loading paradigm.

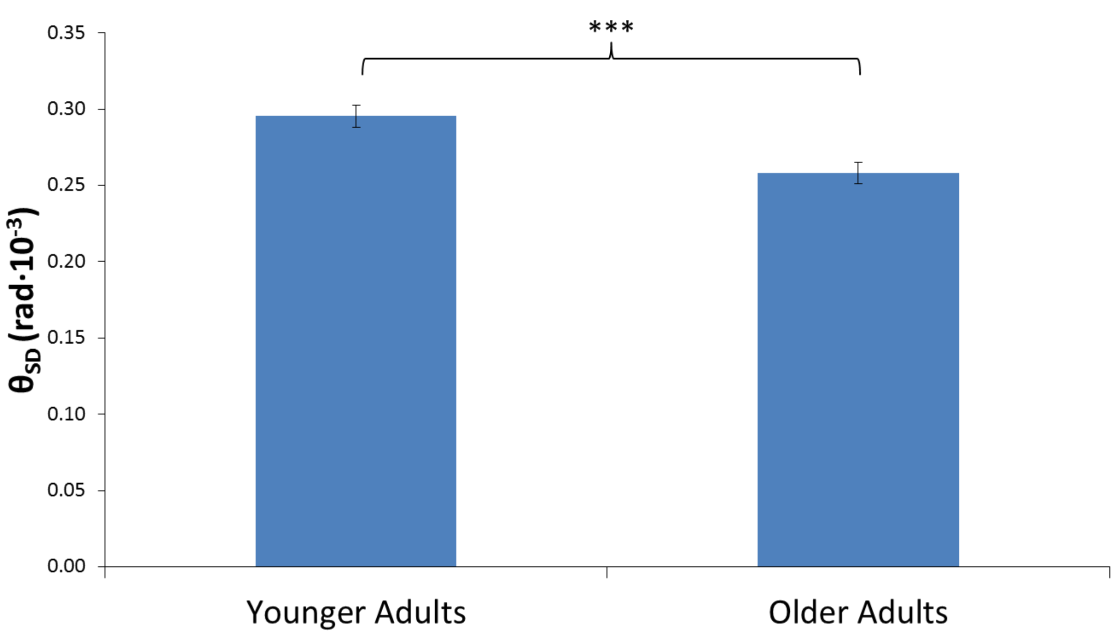


Figure 6.7: Bar graph showing the results for elbow angle positional variability for older and younger adults. Older adults had lower elbow angle positional variability than younger adults while pouring (*** indicates $p < 0.05$). Error bars represent the standard error for the least squares means of each loading paradigm.

6.5 Discussion

The primary hypothesis was supported by this study which found an increase in elbow angle positional variability with increasing co-contraction when performing a positionally demanding task. The current dogma suggests that subjects increase co-contraction when presented with a more demanding task in order to decrease their positional variability. The results from this study suggest that the increased co-contraction level actually decreases the subject’s ability to perform the task.

Additionally, the study found that increased co-contraction above natural levels decreases performance in a pouring task. This suggests that decreasing the subject’s co-contraction level may increase task performance. Future research could bolster this claim by testing subjects’ performance while the subjects co-contract below the subjects’ natural levels.

The secondary hypothesis, $H_{6.2}$: Positional variability and co-contraction will increase as the target size decreases, was partially supported by this study. Co-contraction increased monotonically as target size decreased; however, elbow angle positional variability increased with decreasing target size only for the three largest target sizes. The elbow angle positional variability decreased for the smallest target size. The increase in co-contraction with a decrease in target size had been shown in other, less directly applicable, experiments (Laursen et al., 1998; Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006). The results of this study suggest that subjects are either changing their muscle activation to complete the task successfully or that they are more nervous about the increasing difficulty of the task and therefore activating their muscles in a manner that is non-optimal.

Another possible explanation for this decrease in elbow angle variability for the smallest target trials may be the subject's attention. It has been shown that subjects perform worse at tasks when they are distracted (e.g., Huxhold et al., 2006 and Donker et al., 2007). The pouring task is a natural task that may not require much concentration to perform. Therefore, the subjects may have had decreased attention for the three largest target sizes. However, the smallest target presented an extremely demanding task. The size of the water stream was as large as the target size. This meant that even very capable subjects were unable to complete the task with a 100% success rate and therefore required subject's heightened attention when performing those trials. This heightened attention could explain the decrease in positional variability during those trials. However, increased co-contraction for the smallest target size still led to increased elbow angle variability and decreased performance.

The increased co-contraction between target levels did not always result in higher positional variability. In fact, the smallest target size had both high co-contraction and lower positional variability than the medium two target sizes. More research needs to be performed to determine why the co-contraction differences between targets did not result in increased variability when the co-contraction within a target level did increase variability.

The difference in significance between elbow angle variability and spout variability may be partially explained by postural sway. While elbow angle variability measured only the variability of the elbow joint, the spout variability included postural sway combined with variability from the shoulder, elbow, and hand. This study primarily investigated elbow variability and its relation to elbow muscle co-contraction to avoid these confounding factors.

Older adults co-contracted more than younger adults during the pouring task. This result was reported in previous studies (Hortobagyi and DeVita, 2006; Seidler-Dorbin et al., 1998; and Klein et al., 2001) but rejected by others (Valour and Pousson, 2003). While the older adults had higher co-contraction levels, older adults exhibited lower amounts of positional variability than younger adults which agrees with the findings of Chapter 4, but is in contrast to the findings of Ranganathan et al. (2001). It is possible that younger adults were more focused on pouring the water at the correct speed and height and were not as concerned about the success rate. This is slightly supported by the trend of younger adults' success rate being lower than that of older adults. However, this difference was not significant and therefore warrants further investigation. Future research could include a distraction task to determine whether the attention levels vary between younger and older adults.

One limitation of the study was the lack of uniform practice time for the subjects. Subjects were allowed to practice in order to get the correct pouring pace and height. However, some subjects needed more practice than others to pour at the correct pace. It was observed that older adults struggled more with the speed of pouring, which may have resulted in more practice trials; however, no data were collected to document this effect. However, the effect of trial number was tested for each analysis and was found to be insignificant suggesting that all subjects had fully learned the task before beginning the trials. While the practice trials were necessary for consistency between the subjects, it may have allowed some subjects to become more familiar with the task and therefore to perform better during the actual trials.

6.6 Conclusions

Smaller targets led to higher co-contraction levels, suggesting that subjects increase co-contraction in order to improve performance. However, for a given target size, increased co-contraction leads to greater positional variability and decreased performance. It may be that the increased difficulty of the task causes the subject to co-contract at higher levels, but that co-contraction causes higher variability that confounds the task. This premise is supported by experiments such as Gribble et al. (2003), which show a decrease in co-contraction and increase in performance as subjects became more familiar with the task. The research presented in this chapter suggests that occupational therapists may be able to increase their patient's performance in precision tasks by helping them lower their level of co-contraction.

6.7 References

Donker, S. F., Roerdink, M., Greven, A. J., & Beek, P. J. (2007). Regularity of center-of-pressure trajectories depends on the amount of attention invested in postural control. Experimental Brain Research, 181(1), 1-11.

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Galganski, M. E., Fuglevand, A. J., & Enoka, R. M. (1993). Reduced control of motor output in a human hand muscle of elderly subjects during submaximal contractions. Journal of Neurophysiology, 69(6), 2108-2115.

Graves, A. E., Kornatz, K. W., & Enoka, R. M. (2000). Older adults use a unique strategy to lift inertial loads with the elbow flexor muscles. Journal of Neurophysiology, 83(4), 2030.

Gribble, P. L., Mullin, L. I., Cothros, N., & Mattar, A. (2003). Role of cocontraction in arm movement accuracy. Journal of Neurophysiology, 89(5), 2396-2405.

Hortobagyi, T., & Devita, P. (2006). Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. Exercise and Sport Sciences Reviews, 34(1), 29-35.

Huxhold, O., Li, S. C., Schmiedek, F., & Lindenberger, U. (2006). Dual-tasking postural control: Aging and the effects of cognitive demand in conjunction with focus of attention. Brain Research Bulletin, 69(3), 294-305.

Jones, K. E., Hamilton, A. F. C., & Wolpert, D. M. (2002). Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology, 88(3), 1533.

Klein, C. S., Rice, C. L., & Marsh, G. D. (2001). Normalized force, activation, and coactivation in the arm muscles of young and old men. Journal of Applied Physiology, 91(3), 1341-1349.

Laursen, B., Jensen, B., & Sjøgaard, G. (1998). Effect of speed and precision demands on human shoulder muscle electromyography during a repetitive task. European Journal of Applied Physiology and Occupational Physiology,78(6), 544-548.

Moritz, C. T., Barry, B. K., Pascoe, M. A., & Enoka, R. M. (2005). Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle. Journal of Neurophysiology, 93(5), 2449-2459.

Murray, W. M., Buchanan, T. S., & Delp, S. L. (2000). The isometric functional capacity of muscles that cross the elbow. Journal of Biomechanics, 33(8), 943-952.

Osu, R., Kamimura, N., Iwasaki, H., Nakano, E., Harris, C. M., Wada, Y., & Kawato, M. (2004). Optimal impedance control for task achievement in the presence of signal-dependent noise. Journal of Neurophysiology, 92(2), 1199-1215.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. (2001). Effects of aging on hand function. Journal of the American Geriatrics Society, 49(11), 1478-1484.

Seidler-Dobrin, R. D., He, J., & Stelmach, G. E. (1998). Coactivation to reduce variability in the elderly. Motor Control, 2(4), 314-330.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2005). Can co-activation reduce kinematic variability? A simulation study. Biological Cybernetics, 93(5), 373-381.

Selen, L. P., Beek, P. J., & van Dieën, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieën, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Valour, D., & Pousson, M. (2003). Compliance changes of the series elastic component of elbow flexor muscles with age in humans. Pflugers Archiv : European Journal of Physiology, 445(6), 721-727.

Woldstad, J. C. (1989). *Electromyographic analysis of rapid, accurate elbow movements* (Doctoral dissertation). University of Michigan, Ann Arbor, MI.

Chapter 7

General Discussion

This thesis examined the effect of co-contraction on the positional variability *during* a task. This is significantly different from previous studies which investigated the trial-to-trial endpoint positional variability (position once the task was completed) (Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006). While the trial-to-trial endpoint positional variability is important for reaching tasks, it is not applicable to quasistatic tasks.

Adding variable muscle stiffness to the SDN Theory can predict co-contraction during quasistatic tasks. Previous models based on the SDN Theory were used to predict muscle activation based on the trial-to-trial endpoint variability and constant muscle stiffness (Harris and Wolpert, 1998; and Haruno and Wolpert, 2005). These models are not meaningful for quasistatic tasks that do not require a movement and therefore do not have an end position of a movement. These models also do not include the increasing joint stiffness caused by co-contraction, and therefore do not incorporate the beneficial aspect of to increased co-contraction. Lastly, these models do not optimize the positional variability essential for performing tasks that require hand steadiness. Our model using Variable-Stiffness SDN Theory predicts natural co-contraction in the range of that seen in literature (Chapter 2). The Variable-Stiffness SDN Theory model also correctly predicts the effect of increasing co-contraction above the natural level on positional variability (Chapters 3-4). Lastly, the Variable-Stiffness SDN Theory correctly predicts higher natural co-contraction levels for the destabilizing loading paradigm when compared to a stabilizing loading paradigm (Chapter 3).

Increasing elbow muscle co-contraction above the natural level increases the elbow angle positional variability (Chapters 2-4). Research had suggested that increased co-contraction improves positional steadiness (Gribble et al., 2003; Osu et al., 2004; Selen et al., 2006; and Faisal and Wolpert, 2008). However, previous research investigated the effect of co-contraction on trial-to-trial positional variability. Experiments in this thesis demonstrate that increasing co-contraction above the natural level increases positional variability *during* a task in both younger (Chapters 2-3) and older adults (Chapter 4) performing a quasistatic task.

Increased elbow muscle co-contraction decreases performance of a positionally demanding task (pouring water through small targets, Chapter 6). While increased co-contraction is associated with decreased trial-to-trial endpoint positional variability (Gribble et al., 2003; Osu et al., 2004; and Selen et al., 2006) and increased positional variability during a task (Chapters 2-4), the effect of co-contraction on performance of a precision demanding task was unknown. Increased co-contraction was found to decrease performance during a pouring task (Chapter 6). This suggests that subjects should decrease their elbow muscle co-contraction level when performing positionally demanding tasks.

One limitation is the single joint and planar nature of the studies. The simulation used a basic, planar elbow model to test the effect of elbow muscle co-contraction on elbow angle positional variability. This simplified model was used to permit a clear definition of agonist and antagonist muscles and therefore co-contraction. For three dimensional models, the distinction between agonist and antagonist muscles is less clear as antagonist muscle activation is often necessary to balance out-of-plane moments.

The experimental studies of Chapters 2-5 were studied as single joint with planar motions. Because of this, the pronation and supination moments were assumed to be negligible. Again, this was done to simplify the system. Many of the elbow muscles also span the shoulder joint. An external shoulder moment would require additional activation of these biarticular muscles. While the nature of the experimental setup did not eliminate all out-of-plane and non-elbow joint moments, it did reduce these moments.

Another limitation was the inability to experimentally reduce the subjects' co-contraction levels substantially below their natural level. The experiment in Chapter 5 measured the effect of reducing elbow muscle co-contraction below the natural level; however, despite the efforts to reduce co-contraction, the co-contraction level was reduced by less than 10%. No significant difference in elbow angle positional variability was found during the trials with reduced co-contraction. It is possible that a larger reduction of elbow muscle co-contraction would have had a significant effect on elbow angle positional variability.

Subjects were not surveyed after completing the trials to determine their interpretation of the instructions nor to determine the strategies used by the subjects. Since this data was not collected, it is not possible to determine if the subjects correctly understood the tasks presented to them and if the same strategy was used during each trial. It is possible the subjects forgot the instructions or changed strategies during the trials which could have affected the results.

This study did not find increased positional variability in older adults in comparison to younger adults. This was surprising since older adults were found to have lower performance levels when maintaining a precision pinch posture (Ranganathan et al, 2001). It is possible that the healthy older adults recruited for these studies were not a representative sample of the general older population. A broader sample of older adults may yield different results when compared to younger adults.

We conclude from these studies that increasing elbow muscle co-contraction increases elbow angle variability and decreases performance during a pouring task. This suggests that performance in positionally demanding tasks can be improved by decreasing co-contraction. This finding may be used by occupational therapists to assist older adults in performing activities of daily living that require precision.

7.1 References

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Gribble, P. L., Mullin, L. I., Cothros, N., & Mattar, A. (2003). Role of cocontraction in arm movement accuracy. Journal of Neurophysiology, 89(5), 2396-2405.

Harris, C. M., & Wolpert, D. M. (1998). Signal-dependent noise determines motor planning. Nature, 394(6695), 780-784.

Haruno, M., & Wolpert, D. M. (2005). Optimal control of redundant muscles in step-tracking wrist movements. Journal of Neurophysiology, 94(6), 4244-4255.

Osu, R., Kamimura, N., Iwasaki, H., Nakano, E., Harris, C. M., Wada, Y., & Kawato, M. (2004). Optimal impedance control for task achievement in the presence of signal-dependent noise. Journal of Neurophysiology, 92(2), 1199-1215.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. (2001). Effects of aging on hand function. Journal of the American Geriatrics Society, 49(11), 1478-1484.

Selen, L. P., Beek, P. J., & van Dieen, J. H. (2006). Impedance is modulated to meet accuracy demands during goal-directed arm movements. Experimental Brain Research, 172(1), 129-138.

Selen, L. P., van Dieen, J. H., & Beek, P. J. (2006). Impedance modulation and feedback corrections in tracking targets of variable size and frequency. Journal of Neurophysiology, 96(5), 2750-2759.

Chapter 8

Conclusions

- The Variable-Stiffness Signal Dependent Noise (SDN) Theory model predicts an optimal elbow muscle co-contraction level that minimizes elbow angle positional variability (Chapter 2).
- Above-natural elbow muscle co-contraction increases elbow angle positional variability in younger and older adults (Chapters 2-4).
- Optimal/natural elbow muscle co-contraction is higher for a destabilizing loading moment paradigm than a stabilizing loading moment paradigm in younger and older adults (Chapters 3-4).
- Natural elbow muscle co-contraction level is higher and elbow angle positional variability is lower for older adults than younger adults (Chapters 3-4).
- Elbow muscle co-contraction increases as pouring target size decreases (Chapter 6).
- Increasing elbow muscle co-contraction increases elbow positional variability and decreases pouring accuracy (Chapter 6).

Chapter 9

Suggestions for Future Research

While this thesis was able to prove that increased co-contraction increases positional variability, the relationship between lower co-contraction and positional variability is not known. This knowledge gap could be addressed by providing biofeedback to the subjects that may enable the subject to reduce their co-contraction level. Feedback could be provided by converting the electromyography (EMG) signal into an audible cue for the subject with the pitch indicating the co-contraction level. Subjects could perform a task to establish their natural co-contraction level and positional variability. Next they could perform the task with the audible biofeedback to help them decrease their co-contraction level. By randomizing these trials with natural trials we could determine the effect of the decreased co-contraction on task performance.

Another area of future research is the theory that humans optimize their musculoskeletal control to maximize their performance for a particular task. Standing subjects have more postural sway when they are asked to stand still than when they are asked to stand still and perform a nominal distraction task (Huxhold et al., 2006). Psychologists have explained this as the greater brain activity level enabling subjects to decrease postural sway. However, other studies have shown decreased performance with dual tasks (Beilock et al., 2002). The hypothesis that subjects perform better because they are less able to focus on the desired task of minimizing postural sway should be tested. To do this, subjects could be tested while they are focusing on standing still and when they are only asked to stand. This would test the hypothesis that subjects' awareness of a task decreases their performance and distraction from the task can improve performance.

This research would have major implications for people who struggle to perform activities of daily living that require steadiness. While it is difficult for people to ignore the demand of the task at hand, they may be able to perform a distraction task to decrease their focus on the task demand and therefore improve performance. If this is true, occupational therapists could teach this technique to aid their patients in these essential life tasks.

Lastly, this research did not directly investigate the effect of elbow muscle co-contraction level on hand positional variability. While it is reasonable to assume that higher elbow angle positional variability leads to increased hand positional variability, this must be verified to make this research more applicable to tasks requiring low hand variability.

9.1 References

Beilock, S. L., Carr, T. H., MacMahon, C., & Starkes, J. L. (2002). When paying attention becomes counterproductive: Impact of divided versus skill-focused attention on novice and experienced performance of sensorimotor skills. Journal of Experimental Psychology: Applied, 8(1), 6.

Huxhold, O., Li, S. C., Schmiedek, F., & Lindenberger, U. (2006). Dual-tasking postural control: Aging and the effects of cognitive demand in conjunction with focus of attention. Brain Research Bulletin, 69(3), 294-305.

Appendix 1

Moment Variability for One Agonist and One Antagonist Muscle

In the case of a two muscle model of the elbow (one agonist and one antagonist), the total moment on the elbow is:

$$Elbow\ Moment = M_{agonist} + M_{antagonist} + M_{ext}$$

The moments generated by the muscles can be broken down into:

$$M_{muscle} = SF \cdot PCSA \cdot MA \cdot Act\%$$

where SF is the specific force of muscle, PCSA is the physiological cross sectional area of the muscle, MA is the moment arm of the muscle, and Act% is the activation level of the muscle. Let

$$Act\% = Target\% + Noise\%$$

where $Target\%$ is the target activation level and $Noise\%$ is a random activation level caused by noise in the system. The noise is assumed to have a mean of 0 and variance of σ. While the magnitude of σ is a function of target activation ($\sigma = CV \cdot SF \cdot PCSA \cdot Target\%$), the noise is introduced after the target signal is sent and therefore the noise on the agonist and antagonist signal are independent. Assign the following values to the constants:

$$C_{agonist} = SF \cdot PCSA_{agonist} \cdot MA_{agonist}$$

$$C_{antagonist} = SF \cdot PCSA_{antagonist} \cdot MA_{antagonist}$$

Now an equation for the elbow moment can be written as:

$$Elbow\ Moment = C_{agonist} \cdot Act\%_{agonist} + C_{antagonist} \cdot Act\%_{antagonist} + M_{ext}$$

$$= C_{agonist} \cdot \left(Target\%_{agonist} + Noise\%_{agonist}\right) + C_{antagonist} \cdot (Target\%_{agonist} + Noise\%_{agonist}) + M_{ext}$$

Now, to calculate the variance of the elbow moment:

$$Variance = \boldsymbol{E}[ElbowMoment^2] - (\boldsymbol{E}[ElbowMoment])^2$$

First we'll expand the second term using the linearity of expectation:

$$\boldsymbol{E}[ElbowMoment]$$

$$= \boldsymbol{E}\left[\, C_{agonist} \cdot \left(Target\%_{agonist} + Noise\%_{agonist}\right) + C_{antagonist} \cdot \left(Target\%_{agonist} + Noise\%_{agonist}\right) + M_{ext}\right]$$

$$= \boldsymbol{E}[C_{agonist} \cdot Target\%_{agonist} + C_{agonist} \cdot Noise\%_{agonist} + C_{antagonist} \cdot Target\%_{agonist} + C_{antagonist} \cdot Noise\%_{agonist} + M_{ext}]$$

$$= \boldsymbol{E}[C_{agonist} \cdot Target\%_{agonist}] + \boldsymbol{E}[C_{agonist} \cdot Noise\%_{agonist}] + \boldsymbol{E}[C_{antagonist} \cdot Target\%_{agonist}] + \boldsymbol{E}[C_{antagonist} \cdot Noise\%_{agonist}] + \boldsymbol{E}[M_{ext}]$$

However, $\boldsymbol{E}[constant] = constant$ and the only non-constants are the Noise%.

$$= C_{agonist} \cdot Target\%_{agonist} + C_{agonist} \cdot \boldsymbol{E}[Noise\%_{agonist}] + C_{antagonist} \cdot Target\%_{agonist} + C_{antagonist} \cdot \boldsymbol{E}[Noise\%_{agonist}] + M_{ext}$$

Since the mean of the noise=0,

$$\boldsymbol{E}[ElbowMoment]$$

$$= C_{agonist} \cdot Target\%_{agonist} + C_{antagonist} \cdot Target\%_{agonist} + M_{ext}$$

Since we wish to investigate a quasistatic situation,

$$\boldsymbol{E}[ElbowMoment] = 0 = (\boldsymbol{E}[ElbowMoment])^2$$

Next expand the first term: $\boldsymbol{E}[ElbowMoment^2]$.

$$
\begin{aligned}
&ElbowMoment^2 \\
&\quad = \left(C_{agonist} \cdot Target\%_{agonist} + C_{antagonist} \cdot Target\%_{agonist} + M_{ext}\right)^2 \\
&\quad + 2 \cdot M_{ext} \cdot C_{agonist} \cdot Noise\%_{agonist} + C_{agonist}^2 \cdot Noise\%_{agonist}^2 + 2 \\
&\quad \cdot M_{ext} C_{antagonist} \cdot Noise\%_{antagonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \\
&\quad \cdot Noise\%_{agonist} Noise\%_{antagonist} + C_{antagonist}^2 \cdot Noise\%_{antagonist}^2 + 2 \\
&\quad \cdot C_{agonist}^2 \cdot Noise\%_{agonist} \cdot Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \\
&\quad \cdot Noise\%_{antagonist} Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \\
&\quad \cdot Noise\%_{agonoist} \cdot Target\%_{antagonist} + 2 \cdot C_{antagonist}^2 \\
&\quad \cdot Noise\%_{antagonist} \cdot Target\%_{antagonist}
\end{aligned}
$$

But $\left(C_{agonist} \cdot Target\%_{agonist} + C_{antagonist} \cdot Target\%_{agonist} + M_{ext}\right) = 0$ so

$$
\begin{aligned}
&ElbowMoment^2 \\
&\quad = 2 \cdot M_{ext} \cdot C_{agonist} \cdot Noise\%_{agonist} + C_{agonist}^2 \cdot Noise\%_{agonist}^2 + 2 \\
&\quad \cdot M_{ext} C_{antagonist} \cdot Noise\%_{antagonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \\
&\quad \cdot Noise\%_{agonist} Noise\%_{antagonist} + C_{antagonist}^2 \cdot Noise\%_{antagonist}^2 + 2 \\
&\quad \cdot C_{agonist}^2 \cdot Noise\%_{agonist} \cdot Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \\
&\quad \cdot Noise\%_{antagonist} Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \\
&\quad \cdot Noise\%_{agonoist} \cdot Target\%_{antagonist} + 2 \cdot C_{antagonist}^2 \\
&\quad \cdot Noise\%_{antagonist} \cdot Target\%_{antagonist}
\end{aligned}
$$

$$\boldsymbol{E}[ElbowMoment^2]$$

$$= 2 \cdot M_{ext} \cdot C_{agonist} \cdot \boldsymbol{E}[Noise\%_{agonist}] + C_{agonist}^2 \cdot \boldsymbol{E}[Noise\%_{agonist}^2]$$

$$+ 2 \cdot M_{ext} C_{antagonist} \cdot \boldsymbol{E}[Noise\%_{antagonist]} + 2 \cdot C_{agonist} \cdot C_{antagonist}$$

$$\cdot \boldsymbol{E}\left[Noise\%_{agonist}\right] \cdot \boldsymbol{E}\left[Noise\%_{antagonist}\right] + C_{antagonist}^2$$

$$\cdot \boldsymbol{E}\left[Noise\%_{antagonist}^2\right] + 2 \cdot C_{agonist}^2 \cdot \boldsymbol{E}\left[Noise\%_{agonist}\right]$$

$$\cdot Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist}$$

$$\cdot \boldsymbol{E}\left[Noise\%_{antagonist}\right] Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist}$$

$$\cdot \boldsymbol{E}\left[Noise\%_{agonoist}\right] \cdot Target\%_{antagonist} + 2 \cdot C_{antagonist}^2$$

$$\cdot \boldsymbol{E}\left[Noise\%_{antagonist}\right] \cdot Target\%_{antagonist}$$

$$= 2 \cdot M_{ext} \cdot C_{agonist} \cdot \mathbf{0} + C_{agonist}^2 \cdot \boldsymbol{E}\left[Noise\%_{agonist}^2\right] + 2 \cdot M_{ext} C_{antagonist} \cdot \mathbf{0} + 2$$

$$\cdot C_{agonist} \cdot C_{antagonist} \cdot \mathbf{0} \cdot \mathbf{0} + C_{antagonist}^2 \cdot \boldsymbol{E}\left[Noise\%_{antagonist}^2\right] + 2$$

$$\cdot C_{agonist}^2 \cdot \mathbf{0} \cdot Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \cdot \mathbf{0}$$

$$\cdot Target\%_{agonist} + 2 \cdot C_{agonist} \cdot C_{antagonist} \cdot \mathbf{0} \cdot Target\%_{antagonist}$$

$$+ 2 \cdot C_{antagonist}^2 \cdot \mathbf{0} \cdot Target\%_{antagonist}$$

$$= C_{agonist}^2 \cdot \boldsymbol{E}\left[Noise\%_{agonist}^2\right] + C_{antagonist}^2 \cdot \boldsymbol{E}[Noise\%_{antagonist}^2]$$

Since $\sigma_X^2 = \boldsymbol{E}[(X - \boldsymbol{E}[X])^2]$ and $\boldsymbol{E}[Noise\% = 0]$, $\sigma_{Noise\%}^2 = \boldsymbol{E}[(Noise\% - 0)^2] = \boldsymbol{E}[Noise\%^2]$. Therefore,

$$\boldsymbol{E}[ElbowMoment^2]$$

$$= C_{agonist}^2 \cdot \boldsymbol{E}\left[Noise\%_{agonist}^2\right] + C_{antagonist}^2 \cdot \boldsymbol{E}\left[Noise\%_{antagonist}^2\right]$$

$$= C_{agonist}^2 \cdot \sigma_{Noise\%,agonist}^2 + C_{antagonist}^2 \cdot \sigma_{Noise\%,antagonist}^2$$

Appendix 2

A Theoretical Study of the Effect of Elbow Muscle Co-Contraction Level on Forearm Steadiness

(Abstract presented at the American Society of Biomechanics, Pennsylvania State University, August 2009)

A2.1 Introduction

Some older adults struggle to perform everyday tasks requiring fine motor skills. These include precise movements such as buttoning shirts, tying shoes, feeding themselves, using phones, or inserting a key in a lock. The performance of many of these tasks is frustrated by a lack of hand steadiness. The literature suggests that older adults have less hand steadiness than younger adults (Ranganathan et al, 2001) as well as higher amounts of co-contraction (Laidlaw et al., 2002).

Upper extremity steadiness has been quantified as the variation in hand position both for static tasks (Laidlaw et al., 2002) and for reaching tasks (Gribble et al., 2003). Increased co-contraction is associated with lower inter-movement trajectory variability within same subjects (Faisal et al., 2008). No association was found between co-contraction and acceleration variability between subjects (Laidlaw et al., 2002). However, the effect of co-contraction on within-subject positional variability within a given single movement or task, which is closely related to steadiness, has not been studied.

The goal of this research, therefore, was to determine if the additional co-contraction exhibited by older adults is a compensatory action to steady the hand or if it is a source of additional positional variability. Our working hypothesis is that, in the healthy individual, elbow muscle co-contraction level largely determines the steadiness of the forearm. Our primary hypothesis is that there exists an optimum co-contraction level that maximizes forearm steadiness. The secondary hypothesis is that an age-related decrease in muscle contractile strength leads to decreased optimal steadiness.

A2.2 Methods

We developed a planar model of the upper and lower arm complete with four major agonist and four major antagonist muscles of the elbow (Figure A2.1). The moment and stiffness developed by each muscle was considered a function of muscle cross-sectional area, moment arm, and activation level. The muscles develop a net moment about the joint, along with a rotational resistance to internal or external perturbing moments that is based upon the short range muscle resistance to stretch (but not compression, of course). We assumed the normal variability (SD) in the magnitude of each muscle force to be a function of muscle cross-sectional area and activation level (Hamilton et al, 2004). The variation in each muscle force contributes to variability in the net joint moment as well as net joint stiffness. The stiffness was then used to calculate the positional variability of the forearm, assuming the upper arm was grounded.

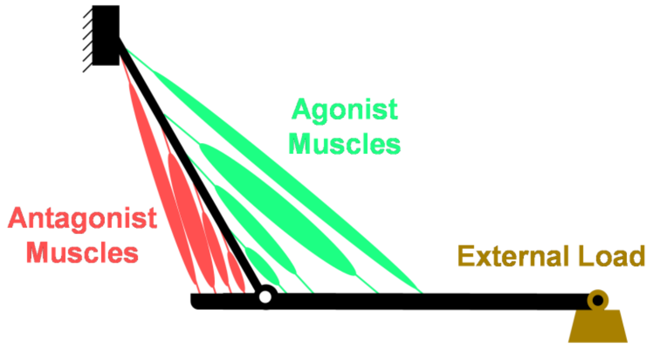


Figure A2.1: Simple Elbow Joint Model

Given the external moment applied to the joint and the co-contraction level of the muscles, a MATLAB algorithm solved for the muscle activation pattern that minimizes the positional variability of the elbow. The co-contraction level was then varied to determine the effect of co-contraction on forearm steadiness and test the primary hypothesis.

We next varied the muscle properties of the model muscles to simulate how these might affect the steadiness of the forearm in younger and older adults. The simulation was then used to find the optimum co-contraction level that maximized the steadiness of the hand for both the younger adult model and the older adult model. The secondary hypothesis was then tested.

A2.3 Results and Discussion

The simulation results demonstrated an optimum co-contraction level that maximizes forearm steadiness supporting the primary hypothesis (Figure A2.2). Higher levels of co-contraction were associated with more variability. This finding is contrary to studies which examined the variability between movements and then applied the same reasoning to a single movement (Gribble et al., 2003,Faisal et al., 2008).

The simulation results suggest that people with weaker muscles, such as many older adults, will show a decrease in steadiness supporting the secondary hypothesis (Figure A2.3). This corroborates findings of manual steadiness in older subjects (Ranganathan et al, 2001). While muscle strength is not the only change that occurs with age, the model was sufficiently sensitive to this parameter to replicate the aging effect on steadiness.

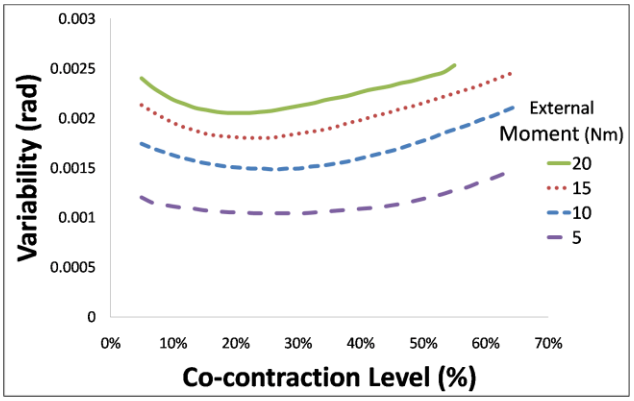


Figure A2.2: Simulation Data Showing Optimal Co-Contraction Levels for Varying External Moments

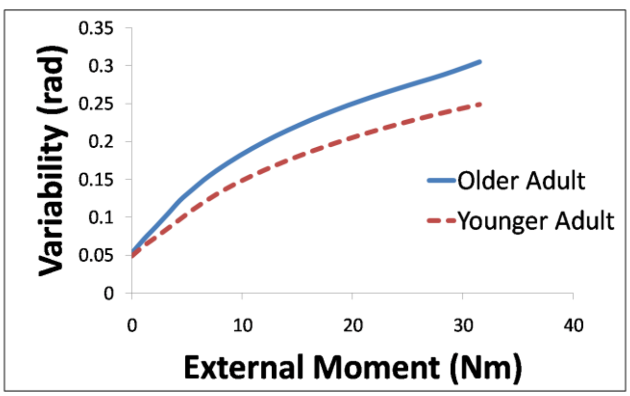


Figure A2.3: Simulation Data Suggesting Older Adults have Higher Variability

The simulation also suggests that older adults have lower optimal co-contraction levels when exerting the same external moment as younger adults. This result suggests that the higher level of co-contraction seen in older adults is not to increase steadiness. Since co-contraction decreases the effect of internal or external disturbances, the elderly may be favoring joint stiffness over steadiness. If these modeling predictions were to be validated by experiment, then future interventions might target co-contraction levels in older adults.

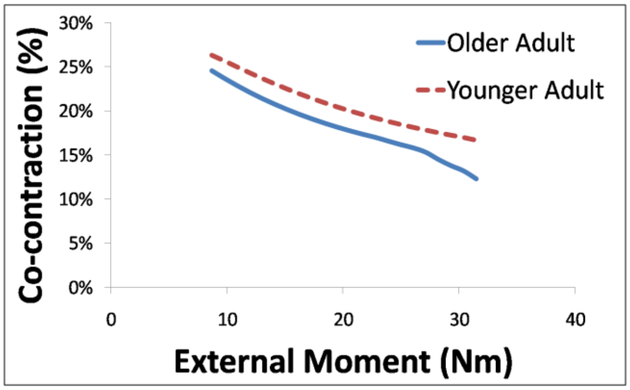


Figure A2.4: Simulation Data Suggesting Older Adults have Lower Optimal Co-Contraction Levels than Younger Adults

A2.4 Conclusions

The results suggest that elbow muscle co-contraction level plays a significant role in forearm steadiness. Positional variability is predicted to be increased at the greatest co-contraction levels which is contrary to current dogma. The simulation also predicts reduced steadiness in older adults based solely on the decrease in muscle cross-sectional area associated with aging.

A2.5 References

Faisal, A. A., Selen, L. P., & Wolpert, D. M. (2008). Noise in the nervous system. Nature Reviews.Neuroscience, 9(4), 292-303.

Gribble, P. L., Mullin, L. I., Cothros, N., & Mattar, A. (2003). Role of cocontraction in arm movement accuracy. Journal of Neurophysiology, 89(5), 2396-2405.

Hamilton, A. F., Jones, K. E., & Wolpert, D. M. (2004). The scaling of motor noise with muscle strength and motor unit number in humans. Experimental Brain Research., 157(4), 417-430.

Laidlaw, D. H., Hunter, S. K., & Enoka, R. M. (2002). Nonuniform activation of the agonist muscle does not covary with index finger acceleration in old adults. Journal of Applied Physiology, 93(4), 1400-1410.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. (2001). Effects of aging on hand function. Journal of the American Geriatrics Society, 49(11), 1478-1484.

A2.6 Acknowledgements

NIA Training Grant AG000114