

EFFECTS OF TENDON VIBRATION, LIGHT TOUCH, AND MECHANICAL
NOISE ON POSTURAL CONTROL: IMPLICATIONS FOR SOMATOSENSORY
REWEIGHTING

by
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DEDICATION

This dissertation is dedicated to my wife, Valerie, who is in fact a saint. However, I'm sure even saints get close to reaching their breaking point at times. The sacrifices she has made for more than eight years have made this Ph.D. possible. I don't know if she will endeavor to pursue one of her own someday, but the work she has put in to support me for this one is certainly deserving of a doctorate. Love, we did it! On to the next chapter.

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ABSTRACT

In order to maintain balance equilibrium, the body relies on sensory feedback from the visual, vestibular, and somatosensory systems. It is hypothesized that postural control is maintained by dynamically weighting the contributions of afferents from these systems based upon the relevance and accuracy of their inputs, a concept known as sensory reweighting. The reweighting of sensory afferents for balance is more commonly explained holistically as entire sensory systems being up-weighted or down-weighted based on their appropriateness; however, all three sensory modalities utilized in balance (visual, vestibular, and somatosensory) have various types of sensory receptors whose inputs could be reweighted accordingly within a modality, as opposed to reweighting the entire modality as a whole for postural control. This study investigated contributions from various receptor types specifically within somatosensation to postural control. Tactile and muscle spindle receptors from both the upper- and lower-body were manipulated by utilizing combinations of tendon vibration, fingertip light touch (FLT), and small amounts of mechanical noise intended to induce stochastic resonance (SR), a phenomenon where weak sensory inputs may be enhanced by the addition of noise. Three separate experiments were conducted to assess interaction effects on balance among: 1) mechanical noise delivered to the bottom of the feet and Achilles tendon vibration (Aim 1), 2) FLT conditions and Achilles tendon vibration (Aim 2), and 3) FLT and arm tendon vibration conditions (Aim 3). Results revealed that combinations of somatosensory stimuli produced differing postural effects than the individual

stimuli themselves. It was inferred that these effects during interactions were evidence of reweighting occurring within the somatosensory system itself. This study provides further insight into how the sensory reweighting hypothesis accounts for human postural control and how such forms of somatosensory manipulation might be utilized in the development of countermeasures to combat balance deficits in a multitude of populations at greater fall risk.

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I. INTRODUCTION

Falls are a major area of concern for health care systems in both the U.S. and abroad (Galica et al., 2009; McMillan, Booth, Currie, & Howe, 2014). Fall death rates in the U.S. increased by more than 30% from 2007 to 2016 (Burns & Kakara, 2018), and in 2015 alone, the total medical cost of falls exceeded \$50 billion (Florence et al., 2018). Populations such as the elderly, individuals with sensory deficits, and even astronauts are especially at risk (James J Collins et al., 2003; DeMott, Richardson, Thies, & Ashton-Miller, 2007; Haran & Keshner, 2009; A. P. Mulavara et al., 2010; Nevitt, Cummings, Kidd, & Black, 1989; William H. Paloski, 2000). Those at higher risk of falling often exhibit performance declines in one or more of the sensory systems dedicated to providing feedback for postural control (Lockhart, Smith, & Woldstad, 2005), and it is generally accepted that such sensory deficits can lead to poorer balance and falls.

To mitigate fall risk in individuals, it is important to understand how balance is maintained. Balance is achieved through a complex interaction of sensorimotor processes (Fay B Horak, 2006). It is widely accepted that the three sensory systems most responsible for providing feedback during postural control tasks are the somatosensory, vestibular, and visual systems (Diener, Dichgans, Guschlbauer, & Mau, 1984; F B Horak, Nashner, & Diener, 1990). It is believed that the sensory inputs provided by these three modalities are necessary to create a 'body schema', or representation of the body in space (Head & Holmes, 1911; Holmes & Spence, 2004). The body schema essentially forms a frame of

reference from which to measure changes in posture against, in order to evoke required reactions (Head & Holmes, 1911; Mergner & Rosemeier, 1998).

Although the three sensory modalities contributing to balance may provide redundant information at times, achieving optimal stability in the face of various postural challenges requires information from all their appropriate sensory afferents (F B Horak et al., 1990). For example, individuals with peripheral neuropathy may be capable of maintaining balance by relying more on visual inputs. However, instability may ensue if they close their eyes or find themselves in the dark. This example also suggests how postural stability can be achieved when one sensory system's afferents may be lacking or providing erroneous information.

Sensory reweighting is a proposed process contributing to postural control in which afferents that are found to be inaccurate or detrimental to balance performance are down-weighted in favor of more appropriate sensory information (Dettmer, Pourmoghaddam, O'Connor, & Layne, 2013; Maurer, Mergner, & Peterka, 2006; Mergner, Maurer, & Peterka, 2003; R J Peterka, 2002; Temple, Lee, & Layne, 2014; Volkening et al., 2014). Reweighting of sensory input has been shown to be a dynamic and fast occurring process, especially when the threat to postural equilibrium is great (Carver, Kiemel, & Jeka, 2006; Oie, Carver, Kiemel, Barela, & Jeka, 2005; Temple, Lee, & Layne, 2016).

The role various receptor types play in the process of sensory weighting is an area which would benefit from further investigation. For example, somatosensation is comprised from various receptor types throughout the body.

Mechanoreceptors at the bottom of the feet detect pressure sensations from contact with the support surface, while muscle spindles and Golgi tendon organs detect muscle stretch and joint positions throughout the body. Often the sensory weighting of a specific system, such as the somatosensory, is presumed to occur holistically in a sense without considering individual receptor contributions within that system (Hwang, Agada, Kiemel, & Jeka, 2014; Robert J Peterka, 2003; Temple et al., 2016). Models estimating somatosensory contributions to balance often fail to account for the different types of somatosensory receptor inputs available. Even when some models do (Maurer et al., 2006; Mergner et al., 2003; H van der Kooij, Jacobs, Koopman, & Van Der Helm, 2001; Herman van der Kooij, Jacobs, Koopman, & Grootenboer, 1999), they are usually not tested under conditions where different somatosensory receptors are manipulated in ways that provide competing stimuli affecting balance maintenance, such as where one receptor type receives erroneous input while another type receives enhanced input.

It has been shown that stimulating certain types of somatosensory receptors in a manner detrimental to balance maintenance likely leads to a down-weighting of their afferents (Hatzitaki, Pavlou, & Bronstein, 2004; Hwang et al., 2014; R J Peterka, 2002; Temple et al., 2016), and conversely stimulating receptors in a manner that may benefit balance should lead to an up-weighting of their afferent contributions (Dettmer, Pourmoghaddam, Lee, & Layne, 2015; Galica et al., 2009; Kabbaligere, Lee, & Layne, 2017; A. P. Mulavara et al., 2011; A. A. Priplata et al., 2006; A. a Priplata, Niemi, Harry, Lipsitz, & Collins, 2003; A.

Priplata et al., 2002; Temple, De Dios, Layne, Bloomberg, & Mulavara, 2018).

What is not well understood is how the weighting of somatosensation is affected when multiple sources of somatosensory feedback are manipulated. For example, what if one type of somatosensory receptor receives unreliable stimuli while another type of somatosensory receptor simultaneously receives input commonly known to enhance postural control? Would the weighting of somatosensory afferents consistently reflect utilizing the enhanced input over the unreliable stimuli, or would the unreliable stimuli prove to be too strong of an effect to allow for the enhanced stimuli to be used? Perhaps a middle ground would exist where enhanced somatosensory feedback might be able to reduce effects of erroneous somatosensory feedback, but not completely mitigate the effect. How much might that enhanced somatosensory feedback mitigate the erroneous feedback's effect on postural control? Operationally speaking, it is assumed that if balance is improved under conditions with both enhanced and unreliable stimuli present, compared to conditions with just the unreliable stimuli present, then an up-weighting of the enhanced stimuli is likely occurring with a coinciding down-weighting of the unreliable stimuli.

To answer such questions of possible reweighting within the somatosensory system however, means of perturbing and enhancing balance by delivering stimuli to different receptor types must first be discussed.

Muscle vibration is a common tool used in postural control studies to manipulate proprioceptive afferents of the vibrated muscles. Mechanical vibration at low-amplitude (~1-2 mm) and high-frequency (~70-90 Hz)

preferentially stimulates type Ia primary afferent fibers within the muscle spindles (Michel-Pellegrino, Amoud, Hewson, & Duchêne, 2006; Harm Slijper & Latash, 2004; Temple et al., 2016; Thompson, Bélanger, & Fung, 2007, 2011).

Stimulating these fibers within the muscle spindles often generates proprioceptive misinformation about muscle length, creating the perception that the muscle is being stretched and often producing a tonic vibration reflex (G. Eklund & Hagbarth, 1966; Goran Eklund, 1972; Harm Slijper & Latash, 2004).

When vibration is applied to musculature controlling the ankle joint in the anterior-posterior (A/P) direction, postural sway towards the side of the vibration is commonly observed (Caudron, Langlois, Nougier, & Guerraz, 2010). For example, when the tibialis anterior muscles are vibrated during quiet stance, the lengthening sensation from the vibration usually causes them to contract, causing ankle dorsiflexion and consequent forward lean (Teasdale, Furmanek, Germain Robitaille, de Oliveira, & Simoneau, 2017; Temple et al., 2014, 2016). Likewise, when the gastrocnemius or Achilles tendon is vibrated, the calf contracts, causing ankle plantar flexion and subsequent backward lean (Eklund 1972; Houser 2007; Caudron et al. 2010a; Caudron et al. 2010b; Kabbaligere et al. 2017; Teasdale et al. 2017). These types of vibration perturbations sufficiently disrupt proprioceptive muscle spindle afferents, resulting in observable increases in postural sway and center of pressure (CoP) motion, as previously observed in studies stimulating the lower-body (Caudron, Langlois, et al., 2010; Houser, 2007; Kabbaligere et al., 2017; Temple et al., 2014, 2016).

Similar to vibration of musculature about the ankle, vibration has also

been shown to excite muscle spindle primary endings in the arm (Goodwin, McCloskey, & Matthews, 1972; Rogers, Bendrups, & Lewis, 1985; Roll & Vedel, 1982). When the tendons and muscles controlling the elbow joint are vibrated, tonic vibration reflexes and lengthening sensations also occur. Vibration of the biceps tends to produce a sensation that muscle is extending and often results in elbow flexion from contraction of the biceps, while vibration of the triceps produces sensations it is extending and results in contraction of the triceps, producing elbow extension (Goodwin et al., 1972; Rogers et al., 1985; Roll & Vedel, 1982).

Utilizing fingertip light touch (FLT) is another means by which somatosensory information may be manipulated during postural control. When an individual lightly touches a stable surface during balance maintenance, cutaneous mechanoreceptors in the fingertip are then able to provide information about body sway. Several studies have noted decreases in CoP displacement movements to occur when utilizing FLT, which typically is interpreted as better postural control (Bove, Bonzano, Trompetto, Abbruzzese, & Schieppati, 2006; Houser, 2007; J J Jeka & Lackner, 1994; John J. Jeka, 2016; John J. Jeka & Lackner, 1995; J R Lackner, Rabin, & DiZio, 2000; James R. Lackner et al., 1999). The additional afferent input from contact with a stable surface alone is capable of providing the improved postural control, as many studies ensure the fingertip forces in contact remain less than one Newton (N) to ensure the benefits are not the result of mechanical support (Baccini et al., 2007; Baldan, Alouche, Araujo, & Freitas, 2014; Cunha, Alouche, Araujo, & Freitas, 2012; Dickstein,

Shupert, & Horak, 2001; James R. Lackner et al., 1999; Rabin, DiZio, Ventura, & Lackner, 2008). Decreased postural sway has even been observed when utilizing light touch in a seated position (Maaswinkel, Veeger, & Dieen, 2014). In general, it seems that the additional somatosensory input provided through FLT serves to improve postural control; however, the interaction effects that might occur when it is combined with other forms of manipulated somatosensory feedback, such as muscle vibration need to be further examined.

It would be assumed that upper-body, arm vibration stimulations alone would not greatly impact maintenance of normal upright stance, as this balancing task is primarily presumed to occur through control of musculature applying torque about lower-body joints, such as the ankles and hips (F B Horak & Nashner, 1986; Temple et al., 2014). Furthermore, biceps tendon vibration alone has not been shown to significantly impact postural control (Rabin et al., 2008). However, it is not clear how a stimulus such as arm muscle vibration might impact postural control when combined with another stimulus such as light touch. A study combining biceps vibration with FLT while in a Romberg stance did note greater postural sway with biceps vibration and light touch present, than when just light touch was present (Rabin et al., 2008). It is unknown if the same results would occur in a more normal, shoulder-width stance. Furthermore, the interaction effect of triceps muscle vibration and light touch has not been studied. Would triceps vibration disrupt the typically enhanced postural control effect seen with light fingertip touch, as was noted with biceps vibration and light touch (Rabin et al., 2008)?

Another suggested way to manipulate somatosensory afferents would be to improve stimuli detecting capabilities through a process known as stochastic resonance (SR). SR is a phenomenon where the response of nonlinear systems to weak input signals, such as in sensory neurons, may be enhanced by a particular level of noise being added to the system (J J Collins, Chow, & Imhoff, 1995). An inverted-U shape is commonly observed in SR phenomena, where sensory performance capabilities of the system receiving added noise increases in performance characteristics to an optimal level as the amplitude of noise also increases. The addition of more noise beyond this optimal level however, tends to deteriorate signal detection and thus result in performance decrements (McDonnell & Abbott, 2009). In several human sensory system SR studies, the amount of noise needed to achieve optimal sensory detecting capabilities is often at levels imperceptible to the subjects (Dettmer et al., 2015; A. P. Mulavara et al., 2011; A. Priplata et al., 2002). Furthermore, there is evidence that the SR phenomenon is capable of occurring in multiple biological systems, thus the multiple sensory systems involved in maintaining postural control may benefit from noise induced sensory improvement, including the visual (Keiichi Kitajo, Nozaki, Ward, & Yamamoto, 2003; Loader et al., 2007; Sasaki et al., 2008, 2006; Simonotto et al., 1997), vestibular (Goel et al., 2015; A. Mulavara et al., 2015; A. P. Mulavara et al., 2011; Temple et al., 2018), and somatosensory (Collins et al. 1996b; Collins et al. 1997; Collins et al. 2003; Priplata et al. 2003; Priplata et al. 2006) systems. For the purposes of this research, the intent is to focus on utilizing the potential for balance enhancement through SR increasing

somatosensory afferents.

Several studies have noted improvements in postural control when delivering small amounts of mechanical noise to the bottom of the feet, and have attributed the improvements to SR benefitting foot mechanoreceptor input (Dettmer et al., 2015; A. A. Priplata et al., 2006; A. a Priplata et al., 2003; A. Priplata et al., 2002). Effects of SR improving mechanoreceptor detection of weak signals in the fingertips have also been observed when utilizing mechanical noise (Collins et al. 1996b; Collins et al. 1997). A few studies have even noticed improved postural control when applying mechanical noise to the fingertip over light touch itself, which was attributed to SR (Kimura, Kouzaki, Masani, & Moritani, 2012; Magalhães & Kohn, 2011b). Often the postural improvements seen with SR have a more pronounced effect in individuals with decreased somatosensory capabilities, such as the elderly (Dettmer et al., 2015; A. a Priplata et al., 2003), patients with stroke, or those with diabetes (A. A. Priplata et al., 2006). However, it is unknown if an SR effect can be demonstrated in healthy individuals who are simultaneously being perturbed by a strong proprioceptive stimulus such as tendon vibration. For this reason, the current project explored the interaction of multiple somatosensory inputs and their effect on postural control. Specifically, interactions between somatosensory stimuli that are commonly beneficial (subthreshold mechanical noise and light touch) and somatosensory stimuli that is commonly detrimental (tendon vibration) to postural control were observed. Our questions concerned what happens to postural control when the delivery of these competing afferent signals occur from different

somatosensory receptors (mechanoreceptors and muscle spindles), generated from both the upper- (fingertip mechanoreceptors and arm muscle spindles) and lower-body (foot mechanoreceptors and calf muscle spindles). This research assists in a better understanding of contributions from the somatosensory system to postural control.

To assess postural control, most studies collect forms of kinetic metrics. Center of pressure (CoP) data obtained through force plate platforms is most commonly used to assess postural control (Benda, Riley, & Krebs, 1994; Duarte & Freitas, 2010; Winter, Prince, Stergiou, & Powell, 1993). Greater CoP motion and speeds are generated with increased sway and are associated with poorer control during static stance (Benda et al., 1994; Capiciková, Rocchi, Hlavacka, Chiari, & Cappello, 2006; Diderik J A Eikema, Hatzitaki, Tzovaras, & Papaxanthis, 2014; Hatzitaki et al., 2004; Inukai et al., 2018; Pavlik, Inglis, Lauk, Oddsson, & Collins, 1999; H. Slijper & Latash, 2000). To assess CoP motion variability about the mean, the mean-removed root mean square (RMS) can be used (Christopher James Dakin, 2012; Forbes et al., 2016; Luu, Huryn, Van Der Loos, Croft, & Blouin, 2011). Some studies also compute a nonlinear metric known as approximate entropy (ApEn), which can assess the regularity of CoP motion (Cavanaugh et al. 2005a; Cavanaugh et al. 2005b; Cavanaugh et al. 2006; Cavanaugh et al. 2007; Dettmer et al. 2015; Lubetzky et al. 2018).

In addition to kinetic metrics, inertial measurement units (IMUs) or motion capture systems are often utilized to collect kinematic data such as joint angle positions and speeds to assess postural control (Goel et al., 2017; Kabbaligere et

al., 2017; B.-C. Lee, Martin, Ho, & Sienko, 2013; A. P. Mulavara et al., 2011). During static stance conditions, these metrics also generally indicate that less movement is associated with better postural control and balance maintenance (Haran & Keshner, 2009; Kabbaligere et al., 2017). Coordination between body segments can also be assessed with kinematic data, and the Anchoring Index (AI) is one such metric designed for this assessment (Assaiante & Amblard, 1993; B.-C. Lee, Martin, Ho, et al., 2013; Sveistrup, Schneiberg, McKinley, McFadyen, & Levin, 2008).

This project utilized kinetic data from a stationary treadmill platform instrumented with force plates (Bertec Fully Instrumented Treadmill, Columbus, OH, USA) and kinematics from a motion capture system (Vicon Nexus, Oxford, UK) to assess postural control during the various conditions of somatosensory stimulation. Somatosensory stimulation was provided by tendon vibrators (VB115, Techno-Concept, Cereste, France), and vibrotactile tactors (C-2, Engineering Acoustics Inc., Casselberry, FL, USA) provided the subthreshold noise designed to induce SR.

1.1 Problem Statement

Most studies observing sensory afferent contributions to postural control typically provide only one type of manipulation within the modality or modalities (visual, somatosensory, and vision) being studied. Thus, when results are explained within the context of sensory reweighting, the conclusions often tend to be stated in a holistic sense. It is often implied that entire modalities are reweighted according to the appropriateness of afferents resulting from the single

manipulation source within that modality. Such an implication however fails to take into account contributions from other sensory afferents within that modality that may be unaffected by the specific sensory manipulation or could be affected in an entirely different manner by another type of stimuli. For example, a strong proprioceptive stimulus like Achilles tendon vibration may perturb postural control, but when that stimuli is combined with conditions of FLT, the mechanoreceptor input from the finger could mitigate the detrimental postural effects of the tendon vibration (J R Lackner et al., 2000). Research is needed providing multiple, simultaneous manipulations within a single modality to further investigate the sensory reweighting concept and determine if weighting in each of the three sensory modalities contributing to balance does occur holistically, or if there is the potential for reweighting to occur from different receptors within the modalities themselves. Due to a gap in the literature about contributions to postural control from competing afferents within the same sensory modality, this dissertation attempts to address the following questions:

Question #1: Can the typical disrupted postural response observed with Achilles tendon vibration be modulated by a small amount of mechanical noise delivered to the bottom of the feet? This question concerns what happens to postural control when the delivery of competing afferent signals (with light foot noise potentially beneficial and Achilles vibration potentially detrimental to postural control) from two different types of somatosensory receptors (calf muscle spindles and foot mechanoreceptors), with both generated from the lower-body.

Question #2: Can the typical disrupted postural response observed with Achilles tendon vibration be modulated by FLT, as well as light touch combined with mechanical noise? This question concerns what happens to postural control when the delivery of competing afferent signals (fingertip touch and touch + noise being potentially beneficial, while Achilles vibration being potentially detrimental to postural control) occurs from two different types of somatosensory receptors (fingertip mechanoreceptors and calf muscle spindles), with one generated from the upper-body (fingertip) and one generated from the lower-body (calf).

Question #3: Are the postural effects of fingertip touch, as well as fingertip touch with mechanical noise, impacted by the potentially disruptive impact of arm tendon vibration? This question concerns what happens to postural control when the delivery of competing afferent signals (from the fingertip and arm musculature) occurs from two different types of somatosensory receptors (fingertip mechanoreceptors and muscle spindles within the biceps and triceps), with both being generated from the upper-body.

1.2 Research Objectives

The primary objective of this dissertation was to observe how postural control may be impacted when receiving competing afferent input from multiple receptor types and locations within the somatosensory system. Additionally, effects of the delivery method of competing somatosensory stimuli was also be compared. In other words, can a subthreshold mechanical noise stimulus meant to induce SR, modulate the postural effects of the strong, above threshold proprioceptive stimulus that is tendon vibration? Three separate, but

conceptually similar experiments, were designed to address the above research questions. Twenty-five healthy young adults were recruited to perform all three experiments within a single testing session. The purpose of each experiment is discussed in the following paragraphs.

Experiment 1

Experiment 1 was designed to answer Question #1. The subjects performed trials of quiet stance while standing on a pair of insoles fitted with small tactors capable of delivering subthreshold amounts of mechanical vibration noise to the bottom of the feet. Vibrators capable of providing a stronger vibration stimulus and activating type Ia afferents within the muscle spindles of the calf were also strapped to the Achilles tendons. Four trials consisting of two insole tactor conditions (off or on) by two Achilles tendon vibration conditions (off or on) were performed in a randomized order. Subjects performed all trials with their eyes closed to eliminate visual feedback. The only sensory manipulations provided occurred from somatosensory stimulation of the lower-body provided by the tactors and vibrators. Thus, any changes in postural control between conditions are assumed to be the result of sensory reweighting occurring within the somatosensory system, between mechanoreceptor afferents from the bottom of the feet stimulated by the noisy tactors and muscle spindle afferents from the calf muscles stimulated by the Achilles tendon vibration. CoP motion as well as kinematics were monitored and evaluated to determine if changes in the normal postural sway exhibited by Achilles tendon vibration is in any way modified by the addition of tactor noise. In other words, did an interaction effect of Achilles

tendon vibration and foot factor noise exist?

Experiment 2

Experiment 2 was conducted to address Question #2. In this experiment subjects performed static stance trials with the eyes closed. The Achilles tendon vibrators remained intact, but conditions of light fingertip touch as well as light fingertip touch with the addition of subsensory factor noise at the fingertip was introduced in certain conditions. The subjects performed six trials in a randomized order consisting of two Achilles tendon vibration conditions (off or on) by three fingertip touch conditions (no touch, light touch, or light touch with factor noise). Kinematic and CoP data were again evaluated to determine if the typical postural sway exhibited by Achilles tendon vibration was modified by light touch or by light touch with the addition of factor noise at the fingertip. In other words, did an interaction effect of Achilles tendon vibration and fingertip touch conditions exist? Again, only somatosensory afferents were being manipulated in these experimental conditions; therefore, changes to the typical Achilles tendon postural response could result from reweighting within the somatosensory system between lower-body calf muscle spindle afferents impacted by the Achilles tendon vibration and upper-body mechanoreceptor afferents at the fingertip stimulated by light touch and the noisy factor.

Experiment 3

The final experiment addressed Question #3. Static stance with the eyes closed was observed in the subjects under three arm tendon vibration conditions (biceps vibration, triceps vibration, or no vibration) by three fingertip touch

conditions (no touch, light touch, or light touch with tactor noise) for a total of nine randomized trial conditions. CoP and kinematic data were analyzed to observe changes in postural control during the competing upper-body somatosensory stimuli conditions affecting both muscle spindles of the biceps and triceps (arm tendon vibration) as well as mechanoreceptors in the fingertip (light touch and tactor noise). In other words, we tested if an interaction effect of arm tendon vibration conditions and fingertip touch conditions existed.

1.3 Specific Aims and Hypotheses

Aim 1: To determine if an interaction effect exists between small amounts of noisy vibration delivered to the bottom of the feet in the presence of Achilles tendon vibration.

As Achilles tendon vibration typically yields greater CoP motion in the A/P direction, A/P CoP motion metrics such as: mean position (POS), mean speed (SPD), RMS of CoP position, and approximate entropy (ApEn) of the CoP motion were considered as kinetic metrics of postural stability. Additionally, the kinematics of body segment angular motion in the sagittal plane for the head, torso, thigh, and shank were utilized as metrics of postural stability. Mean angular POS, SPD, and RMS were calculated for each of the four body segments, and AI of the head, torso, and thigh was computed to compare coordination of these three segments to their inferior segments. These kinetic and kinematic metrics of postural stability were utilized in assessing postural control for all the aims (1-3) of this study.

Hypothesis 1: Less motion (A/P CoP motion as well as less sagittal plane body segment motion) was expected, with less posterior POS values, decreased speeds, and reductions in variability when insole noisy tactor vibration was present along with Achilles tendon vibration, than when just Achilles tendon vibration was present. We expected disruptive muscle spindle afferents in the calf muscles from Achilles vibration to be down-weighted and mechanoreceptor afferents in the foot enhanced by SR to be up-weighted during balance maintenance, thus producing less sway and a shift in posture less posterior when the foot tactors were active.

Aim 2: To determine if an interaction effect exists between light fingertip touch conditions and Achilles tendon vibration.

Hypothesis 2.1: Less motion (A/P CoP motion as well as less sagittal plane body segment motion) was expected, with less posterior POS values, decreased speeds, and reductions in variability when light fingertip touch was present along with Achilles tendon vibration, than when just Achilles tendon vibration was present. We expected disruptive muscle spindle afferents in the calf muscles from Achilles vibration to be down-weighted and mechanoreceptor afferents in the fingertip to be up-weighted during balance maintenance, thus producing less sway and a shift in posture less posterior when utilizing light fingertip touch.

Hypothesis 2.2: We also expected less motion (A/P CoP motion as well as less sagittal plane body segment motion) with less posterior POS values, decreased speeds, and reductions in variability when light fingertip touch with

mechanical noise was present along with Achilles tendon vibration, than when just light touch and Achilles tendon vibration was present. We expected disruptive muscle spindle afferents in the calf muscles from Achilles vibration to be down-weighted and mechanoreceptor afferents in the fingertip enhanced by SR to be up-weighted during balance maintenance, thus producing less sway and a shift in posture less posterior when the fingertip factor was active.

Aim 3: To determine if an interaction effect exists between light fingertip touch and arm tendon vibration conditions.

In addition to the kinetic and kinematic metrics for aims 1 and 2, right elbow flexion/extension POS, SPD, and RMS were also evaluated for aim 3 in order to evaluate kinematic changes in the elbow of the arm that was performing the light touch conditions with combinations of arm vibration.

Hypothesis 3.1: Less motion (A/P CoP motion as well as less sagittal plane body segment motion) was expected, with less anterior POS values, decreased speeds, and reductions in variability when just biceps tendon vibration was present, than when light fingertip touch conditions (with and without mechanical noise) were present with biceps tendon vibration. We expected disruptive muscle spindle afferents in the biceps muscles from vibration to be down-weighted, in conditions without utilizing light touch to help provide a frame of reference for the body's position. However, when touch conditions were present (both with and without biceps vibration), we expected an up-weighting of arm muscle spindle afferent inputs, as under those conditions they provided input to body position while the fingertip was lightly touching a stable surface. When

both biceps vibration and touch conditions were present, we believed the up-weighting of inappropriate biceps muscle length input would generate more sway and in the anterior direction. In general, we expected biceps vibration to perturb postural control only when it was combined with conditions of light touch.

Hypothesis 3.2: Less motion (A/P CoP motion as well as less sagittal plane body segment motion) was expected, with less posterior POS values, decreased speeds, and reductions in variability when just triceps tendon vibration was present, than when light fingertip touch conditions (with and without mechanical noise) were present with triceps tendon vibration. We expected disruptive muscle spindle afferents in the triceps muscles from vibration to be down-weighted, under conditions without light touch to help provide a frame of reference for the body's position. However, when touch conditions were present (both with and without triceps vibration), we expected an up-weighting of arm muscle spindle afferent inputs, as under those conditions they provided input to body position when the fingertip was lightly touching a stable surface. When both triceps vibration and touch conditions were present, we believed the up-weighting of inappropriate triceps muscle length input would generate more sway and in the posterior direction. In general, we expected triceps vibration to perturb postural control only when it was combined with conditions of light touch.

1.4 Potential Contributions

The results from this study prove useful in providing further insight into how the sensory reweighting hypothesis accounts for human postural control. In particular, these experiments demonstrate what happens when a specific

sensory modality is receiving competing afferents from multiple stimuli acting upon different receptors within that modality. This contribution to the literature could lead to the design of new reweighting models used to predict postural control. Such models may need to be updated to account for the various types of receptors contributing to balance within the individual modalities.

Furthermore, results from this study have wide-standing implications for developing countermeasures to combat balance deficits in a multitude of populations at greater fall risk. For example, SR has been proposed as a means to potentially improve postural control in a number of populations with balance deficits, such as the elderly (Dettmer et al., 2015; Dhruv, Niemi, Harry, Lipsitz, & Collins, 2002; Fujimoto et al., 2016; Gravelle et al., 2002; Moss & Milton, 2003; A. Priplata et al., 2003), stroke patients (Harry, Niemi, Priplata, & Collins, 2005; A. Priplata et al., 2002), those with Parkinson's Disease (Pal, Rosengren, & Colebatch, 2009; Samoudi, Jivegård, Mulavara, & Bergquist, 2015), individuals with diabetic neuropathy (Hijmans, Geertzen, Zijlstra, Hof, & Postema, 2008; A. Priplata et al., 2006), those with vestibular dysfunction (Iwasaki et al., 2014), and astronauts returning to a terrestrial body from microgravity (Goel et al., 2015; A. Mulavara et al., 2015). These experiments tested the efficacy of utilizing mechanical noise to induce SR for balance enhancement under conditions where a strong proprioceptive stimulus like tendon vibration was also present, providing further insight into how these small amounts of noise may (or may not) benefit postural control.

II. LITERATURE REVIEW

2.1 Sensory Contributions to Upright Stance

Maintaining control of upright stance requires a complex interaction of dynamic sensorimotor processes (Fay B Horak, 2006). It has been proposed that the human body utilizes an array of sensory information and multimodal integration of these inputs within the CNS to generate a “body schema,” which provides a neural representation of the body (Gurfinkel, Levik, Popov, & Smetanin, 1988; Holmes & Spence, 2004; Shenton, Schwoebel, & Coslett, 2004). This body schema is dependent upon not only bodily sensations and their integration, but also their perception, and it does not require thought at a conscious level for its development (Gallagher, 2005; Gurfinkel et al., 1988; Laessoe, Barth, Skeie, & McGirr, 2017). During postural control, the body schema serves as a representation of the perceived vertical upright position of the body (Dettmer, 2014). Essentially, it forms a frame of reference or standard from which to measure changes in posture against in order to evoke required reactions (Head & Holmes, 1911; Mergner & Rosemeier, 1998). This body schema is expected to be plastic and constantly changing as new movements and incoming body sensations are brought into relation with it (Gallagher, 2005; Head & Holmes, 1911; Holmes & Spence, 2004), and there is evidence that forms of external afferent stimulation can also manipulate the body schema, potentially enhancing or perturbing the body perception utilized in maintaining postural control (Laessoe et al., 2017). For balance maintenance under both static and dynamic conditions, it is proposed that the body schema is updated

primarily by three sensory modalities providing afferent information: the visual, vestibular, and somatosensory systems (Chiba, Takakusaki, Ota, Yozu, & Haga, 2016; Holmes & Spence, 2004; Fay B. Horak & Macpherson, 1996; Mergner & Rosemeier, 1998; A. Mulavara et al., 2015; Shenton et al., 2004).

2.1.1 Contributions from Vision

Witkin and Asch performed some of the first rod and frame tests, which examined subjects' perceptions of the vertical and horizontal position when viewing a tilted scene with and without a frame as well as with and without body tilt present (Asch & Witkin, 1948a, 1948b; Witkin & Asch, 1948a, 1948b). Errors in the perceived verticality varied among subjects, but roughly half of the subjects incorrectly perceived body tilt in the direction of the frame when in a dark room providing no other reference (Witkin & Asch, 1948b). These experiments and other similar rod and frame tests have shown the importance of the visual system in the perception of verticality, an important aspect of the body schema providing reference for postural control (D J A Eikema, Hatzitaki, Konstantakos, & Papaxanthis, 2013; Einarsson et al., 2018; Hodgson, Christian, & McMorris, 2010; Isableu et al., 2010; Ruitenbergh et al., 2017). The degree to which individuals rely on visual input to judge their spatial position does vary however, and it has often been termed as their visual dependency (Brady et al., 2012; Cousins et al., 2014; Einarsson et al., 2018; Roberts, Da Silva Melo, Siddiqui, Arshad, & Patel, 2016).

Direct demonstrations of the importance of vision for postural control have been noted in publications for more than a century. Increased postural sway in

the absence of vision and in the blind are often observed (Edwards, 1946; Goldberg, Hernandez, & Alexander, 2005; J J Jeka & Lackner, 1994; John J. Jeka, Easton, Bentzen, & Lackner, 1996; McKay, Wu, & Angulo-Barroso, 2014). In 1895, Wood described postural instability in people when exposed to an optical illusion known as the 'Haunted Swing' where the surrounding room rotated separately from the ground (R. Wood, 1895). The illusion seemed to perturb visual input to the point that postural control was notably affected. Similarly, David Lee conducted experiments in the 1970s that demonstrated the importance vision played in the maintenance of postural control in both infants (David N. Lee & Aronson, 1974) and adults (D. N. Lee & Lishman, 1975). In these experiments it was demonstrated that a room with moving walls induced postural sway such that subjects tended to sway forward as the walls moved away from them and backwards as the walls moved towards them. Essentially, the room provided incorrect visual information that they were swaying in the opposite direction, and the subjects adjusted their posture accordingly, despite the input being erroneous (D. N. Lee & Lishman, 1975; David N. Lee & Aronson, 1974). Further supporting the importance of vision in postural control, studies have shown age induced declines in aspects of vision such as depth perception and contrast sensitivity to correspond to decreases in postural control and increased fall risk (S. Lord, Clark, & Webster, 1991; S. R. Lord, 2006; S. R. Lord & Dayhew, 2001).

2.1.2 Vestibular Contributions

In addition to visual input, the contributions of the vestibular system to postural control are also well noted. Vestibular afferents from the inner ear originate from the semicircular canals and the otolith organs (Herdman & Clendaniel, 2014). The semicircular canals detect head rotation, and their canal planes are oriented perpendicular to each other to provide roll, pitch, and yaw information. Hair cells in the ampulla of the semicircular canals are innervated by afferent neurons that fire when endolymph in the canals stimulate the hair cells. If rotation occurs, inertia causes the endolymph to lag behind and bend the hair cells, changing their firing rates and providing afferent input about the rotational direction. Thus the semicircular canals are vital for providing the sensory information needed in the vestibulo-ocular reflex, which allows eye movements to maintain gaze while the head is moving (Herdman & Clendaniel, 2014). The utricle and saccule of the otolith organs similarly use movement of hair cells to provide afferent input about the gravity vector and linear acceleration (Herdman & Clendaniel, 2014; Fay B. Horak & Macpherson, 1996). When a change in the gravity vector or linear acceleration occurs, movement of the otoconia occurs in these organs that stimulate hair cells to discharge afferents representative of the changes in linear acceleration detected.

For postural control, inputs from the vestibular system invoke vestibulo-spinal reflexes that trigger muscle responses utilized in the maintenance of balance (Angelaki & Cullen, 2008). In fact studies manipulating vestibular afferents with electrical currents have been able to demonstrate coherence between those currents and postural sway (Fitzpatrick, Burke, & Gandevia, 1996;

Forbes et al., 2016; Pavlik et al., 1999; A P Scinicariello, Eaton, Inglis, & Collins, 2001; Anthony P Scinicariello, Inglis, & Collins, 2002), as well as muscle activity in the lower-body (Christopher J Dakin, Son, Inglis, & Blouin, 2007).

Furthermore, vestibular input is utilized in the vestibulocollic reflex, contracting neck musculature in order to maintain a stable position of the head, providing the spatial orientation needed from which to make postural adjustments for balance maintenance (Angelaki & Cullen, 2008; Cullen, 2004; Herdman & Clendaniel, 2014). Like the visual system, declines in vestibular function have been shown to negatively impact postural control, such as in those experiencing vestibular loss (Fay B. Horak, Kluzik, & Hlavacka, 2016; James R. Lackner et al., 1999), vertigo (Cohen, Mulavara, Peters, Sangi-Haghpeykar, & Bloomberg, 2014), astronauts during gravitational transitions (W H Paloski, Reschke, Black, Doxey, & Harm, 1992; William H. Paloski, 2000), or even through means of purposefully disrupting vestibular afferents with mechanical or electrical stimulation (MacDougall et al. 2006; Chien et al. 2016; Chien et al. 2017)

2.1.3 Somatosensory Contributions

Somatosensory input is comprised of various receptors throughout the body that provide sensory input about pain, temperature, touch, and proprioception. Some examples of these various receptors generating feedback information about postural control include tactile receptors located in the skin, Golgi tendon organs at the origin and insertion points of skeletal muscle fibers to detect tension, and muscle spindles embodied within the muscles themselves.

Tactile mechanoreceptors within the skin have been shown to provide afferent input about pressure sensation that is utilized for postural control. In the bottom of the feet, mechanoreceptors can detect pressure, sensing the body's center of mass as it sways over the feet, providing feedback, and evoking postural reflexes (Inglis, Horak, Shupert, & Jones-Rycewicz, 1994; Strzalkowski, Ali, & Bent, 2017; Zehr & Stein, 1999). The mechanoreceptors at the bottom of the feet play an important role in balance maintenance, and numerous studies have noted that decreased sensitivity of these receptors and their input, such as through disease, natural aging, or other means can detrimentally impact postural control (Cohen et al., 2013; Diener et al., 1984; Hijmans et al., 2008; F B Horak et al., 1990; Machado, Bombach, Duysens, & Carpes, 2016; Magnusson, Enbom, Johansson, & Pyykkö, 1990).

Muscle spindles are located within the muscles and provide information about muscle length. This input on muscle length is important for postural control as it is utilized in the perception of joint position and movements (Goodwin et al., 1972; Roll & Vedel, 1982). For example, muscle spindles within musculature controlling movement about the ankles can detect body tilt occurring at the ankles, which is in turn utilized to adjust posture. If too much stretch is detected in the tibialis anterior muscles, they contract to pull body position forward. Likewise, if too much stretch is detected in the gastrocnemius, it contracts to pull body position backward. Input from the muscle spindles in some musculature can therefore be manipulated to induce changes in posture. One common means of manipulating the afferents from muscle spindles involved in postural

control is to utilize mechanical vibration, which is further discussed in section 2.2 of this dissertation.

Although somatosensory receptors from the lower-body are considered highly important for providing input about postural control (Fitzpatrick, Rogers, & McCloskey, 1994), the body is capable of utilizing somatosensory inputs throughout the body to update the body schema for balance maintenance. For example, making light contact with a stationary surface has been shown to improve postural control (Baldan et al., 2014; Holden, Ventura, & Lackner, 1994; John J. Jeka, 2016; Maaswinkel et al., 2014), which is further discussed in section 2.3 of this document. Likewise, vibration of neck musculature (Bove et al., 2006; Bove, Fenoglio, Tacchino, Pelosin, & Schieppati, 2009; Gomez et al., 2009; Ivanenko, Talis, & Kazennikov, 1999; Verrel, Cuisinier, Lindenberger, & Vuillerme, 2011) as well as tactile receptors around the torso (B.-C. Lee, Martin, Ho, et al., 2013; B.-C. Lee, Martin, & Sienko, 2013, 2012) has been shown to induce postural responses when standing, suggesting that somatosensory receptors from these areas can provide inputs utilized in balance maintenance (more information on these studies is in sections 2.2 and 2.3 of this document).

2.1.4 Sensory Reweighting Hypothesis

It has been proposed that as afferents from the three primary sources of sensory information utilized in postural control (visual, vestibular, and somatosensory) are generated, the body may weight reliance on these different afferents based upon their accuracy, a concept known as sensory reweighting. Specifically, the motor outcomes generated for postural control are hypothesized

to come from a weighted sum of the sensory input, and if a sensory cue is missing or inaccurate, inputs from more reliable sensory cues become more heavily weighted (Carver et al., 2006; Haran & Keshner, 2009; R J Peterka, 2002). For example, if information from vision becomes compromised or less accurate, such as by standing in a dark room, visual inputs may be down-weighted in favor of more accurate inputs from the vestibular and somatosensory systems to maintain postural control.

Reweighting is dynamic and seems capable of happening rather quickly in most individuals (Carver et al., 2006; Hwang et al., 2014; Temple et al., 2016; Herman van der Kooij & Peterka, 2011). It has been noted however that for some populations, such as the elderly, reweighting may take more time to occur, possibly due to degeneration of certain sensory systems (D J A Eikema et al., 2013; Diderik J A Eikema et al., 2014). For instance, greater reliance on vision may occur with age as vestibular afferents deteriorate (Pasma et al., 2015). Likewise individuals with a disease like neuropathy may consequently upweight vestibular afferents as well, due to deteriorating somatosensory input (F B Horak & Hlavacka, 2001). Having consistently higher reliance on one system such as vision in these instances however, could make it more difficult to reweight to other, perhaps less reliable modalities, if that system were suddenly compromised, such as by being in a dark room in our previous example. Such a greater reliance on one system would therefore not be ideal and potentially lead to slower reweighting (D J A Eikema et al., 2013; Diderik J A Eikema et al., 2014). Some studies suggest greater reweighting can occur and faster, as

threats to postural control become larger (A. D. Goodworth & Peterka, 2009; Honegger, Tielkens, & Allum, 2013; Oie et al., 2005; Herman van der Kooij & Peterka, 2011). Furthermore, there is evidence reweighting occurs over time in motor learning, as individuals learn which afferents are most appropriate for balance in specific tasks (Honegger et al., 2013). One study compared weighting of sensory inputs between tightrope walkers and those untrained in the skill when performing various tandem stance balancing skills. Tightrope walkers were found to have an increased reliance on neck and pelvis proprioception than untrained controls, suggesting training may have altered the way tightrope walkers weight these afferents during tandem balance tasks (Honegger et al., 2013).

When investigating how sensory reweighting of afferents occurs during balance maintenance, many studies probe the contributions of individual receptors or sensory systems by manipulating their afferent input in a manner providing information contrary to normal static stance conditions (Hwang et al., 2014; John J. Jeka, Oie, & Kiemel, 2008; MacDougall et al., 2006; R J Peterka, 2002; Harm Slijper & Latash, 2004). Likewise, this study examined the possibility for reweighting of multiple somatosensory inputs during postural control and did so by manipulating various types of somatosensory receptors in different ways. The next few sections (2.2, 2.3, and 2.4) address the various means of manipulating somatosensory inputs that were utilized in this study.

2.2 Mechanical Vibration

2.2.1 Mechanical Vibration Applications

Mechanical vibration has been utilized previously in a variety of contexts to stimulate somatosensory receptors. Studies of training with vibration have revealed increases in muscular strength and power (Luo, McNamara, & Moran, 2005) as well as enhanced balance and coordination (Cidem et al., 2017; Iwamoto, Sato, Takeda, & Matsumoto, 2012; S.-W. Lee, Cho, & Lee, 2013; Tihanyi et al., 2010). Often the vibration is applied indirectly to musculature through whole-body vibration (WBV), by standing on a vibrating platform. WBV is believed to improve synchronization and recruitment of motor units, leading to enhancements in neuromuscular performance, but the exact physiological mechanisms responsible for these improvements are still not well understood (Cidem et al., 2017).

In addition to indirect means of vibration such as WBV, increases in strength, muscle activation, and anabolic responses have also been observed when training with a more localized vibration stimulus on musculature and their corresponding tendons (Couto et al., 2013; Curry & Clelland, 1981; Lapole & Pérot, 2010; Pamukoff, Ryan, & Troy Blackburn, 2014). In fact, applying localized vibration to the heel, as well as the Achilles and tibialis anterior tendons during balance training has been shown to improve postural control and gait in chronic stroke patients (S.-W. Lee et al., 2013), presumably through increased proprioceptive input and muscle activation.

Vibration is also used as a means of augmenting sensory feedback such as providing biofeedback and cueing during postural control and gait. It has been demonstrated that mechanical vibration can be used to provide tactile

cueing for Parkinson's disease (PD) patients, meant to improve performance during both simple and complex motor control tasks (Ivkovic, Fisher, & Paloski, 2016). Vibrating insoles have been shown to improve gait in PD patients when utilizing constant frequencies at suprathreshold intensities (Novak & Novak, 2006). Subthreshold, noisy vibration stimuli delivered through shoe insoles has also been shown to improve gait in the elderly (Galica et al., 2009; Lipsitz et al., 2015; Stephen et al., 2012), which is further discussed in section 2.4 of this document. Several studies have utilized vibrotactile belts that provide body tilt feedback to enhance balance during both standing (B.-C. Lee, Kim, Chen, & Sienko, 2012; Sienko, Vichare, Balkwill, & Wall, 2010; Wall & Kentala, 2010, 2005) and locomotion tasks (Dozza, Wall, Peterka, Chiari, & Horak, 2007). Furthermore, the vibration from such devices has even been shown to generate natural postural sway responses through the activation of cutaneous receptors, without necessarily requiring voluntary adjustment of posture when the vibrators engage to provide tilt feedback (B.-C. Lee, Martin, Ho, et al., 2013; B.-C. Lee, Martin, & Sienko, 2013; B.-C. Lee, Martin, et al., 2012).

In addition to the above uses of vibration to potentially provide beneficial somatosensory stimulation to improve gait and posture, some sensory stimulations from mechanical vibration have been found to negatively impact balance. Mastoid vibration has been shown to perturb vestibular system afferents and can be utilized to investigate its contributions to postural control, as demonstrated in prior studies (Chien et al., 2017, 2016). Likewise, vibration is also capable of activating somatosensory muscle spindle receptors, which detect

muscle stretch. Such activation of these receptors can disrupt balance by generating erroneous proprioceptive feedback utilized in postural control. The following section specifically addresses muscle spindle activation from vibratory stimuli, as well as its effects on postural control.

2.2.2 Activation of Muscle Spindles with Vibration

When postural muscles are stimulated by vibration, a reflexive muscle contraction known as the tonic vibration reflex (TVR) commonly occurs (Cidem et al., 2017; G. Eklund & Hagbarth, 1966; Martin & Park, 1997; Park & Martin, 1993). TVRs are proposed to occur by the vibration preferentially activating muscle spindle primary afferent fibers (type Ia), thus generating the reflex (Ceyte et al., 2007; G. Eklund & Hagbarth, 1966; Goran Eklund, 1972; Lapole & Pérot, 2010; Lebedev & Polyakov, 1992; Martin & Park, 1997; Park & Martin, 1993). Interestingly, the H-reflex, which is commonly utilized as a measure of spinal cord excitability (Knikou, 2008; Kramer, Gollhofer, & Ritzmann, 2013), is suppressed during isolated tendon and muscle vibration that induces the TVR (Cakar, Cidem, Kara, & Karacan, 2014; Gillies, Lance, Neilson, & Tassinari, 1969). This H-reflex suppression is often explained by presynaptic inhibition of type Ia afferents (Gillies et al., 1969). However, it is paradoxical that suppression of the H-reflex is reported to exist along with a tonic reflex elicited by vibration (Cakar et al., 2014; Gillies et al., 1969) that is proposed to occur from stimulated type Ia afferents (Cidem et al., 2017; Martin & Park, 1997; Park & Martin, 1993; Rose & Christina, 2006; Harm Slijper & Latash, 2004; Thompson et al., 2007). This contradiction is known as the vibration paradox, and it has been shown not

to occur as a result of surface electrode recording error from movement induced by the vibration (Cakar et al., 2014).

In addition to the TVR, vibration also produces proprioceptive misinformation about the length of the vibrated muscle. This misinformation creates the perception that the muscle is longer than it actually is (Ceyte et al., 2007; Dettmer et al., 2013; J R Lackner & Levine, 1979; J R Lackner et al., 2000; Temple et al., 2016). Utilizing vibration in the range of 70-100 Hz have consistently been reported to produce the TVR and sensations of elongation in musculature at the neck (Bove et al., 2006, 2009; Verrel et al., 2011), ankle (Michel-Pellegrino et al. 2006; Caudron et al. 2010a; Caudron et al. 2010b; Temple et al. 2014), and elbow (Goodman & Tremblay, 2018; Goodwin et al., 1972; Rabin et al., 2008). Therefore, muscle vibration has commonly been utilized as a tool to probe contributions of muscle spindle proprioceptive input to postural control. Achilles tendon vibration has commonly been utilized to preferentially stimulate muscle spindles of the attached calf musculature in the triceps surae. The vibratory stimulation causes the calf to contract and produce a backwards sway, which can usually be observed with more posterior CoP and COM locations (Ivanenko et al. 1999; Ceyte et al. 2007; Caudron et al. 2010a; Caudron et al. 2010b; Duclos et al. 2014). Likewise, vibration stimulation of the anterior side ankle musculature, such as the tibialis anterior, causes a similar postural response in the opposite direction, where sway is observably more forward (Michel-Pellegrino et al. 2006; Caudron et al. 2010a; Temple et al. 2014).

Arm vibration causes illusory joint motion and muscle contractions too. Vibration of the distal biceps tendon has been shown to result in an illusory elbow extension sensation, often generating bicep contraction and elbow flexion. Likewise, vibration of the triceps causes an illusory elbow flexion sensation, and often generates elbow extension with contraction of the muscle (Roll & Vedel, 1982). Although arm vibration alone may not largely affect postural control, when combined with FLT, it may have an enhanced effect. Biceps vibration combined with FLT has previously been shown to increase center of pressure and head motion than in conditions of biceps vibration without touch (Rabin et al., 2008). The following section addresses more specifically the effects of FLT on postural control.

2.3 Fingertip Light Touch (FLT)

FLT of a fixed, rigid surface has been utilized in numerous studies and shown to reduce postural sway by roughly 50% during quiet stance in healthy individuals (Baldan et al., 2014; Holden et al., 1994; John J. Jeka, 2016). Under conditions when planned movement occurs, anticipatory postural adjustments have also been found to be reduced with the presence of FLT (H. Slijper & Latash, 2000; Harm Slijper & Latash, 2004). Why do such stabilizing effects on postural control occur with FLT? Fingertip contact with the surface provides additional information about body position through fingertip mechanoreceptors as well as other somatosensory receptors providing input on arm position, which can be utilized for improved postural control (Baldan et al., 2014; John J. Jeka, 2016). Not much force for the fingertip contact is needed to provide the postural

benefits, as most studies ensure the contact force is less than one Newton (N), and contact forces less than four N are generally considered to indicate that the improvements are due to the additional sensory input alone, rather than mechanical support of the body through touching a stationary surface (Baldan et al., 2014; Holden et al., 1994).

The benefits of FLT on postural control have been tested under various stance conditions, such as natural bipedal stance (Bove et al., 2006; Clapp & Wing, 1999; Dickstein, 2005; Dickstein et al., 2001), single leg stance (Holden et al., 1994), and tandem Romberg stance (J. Lackner, Rabin, & DiZio, 2001; J R Lackner et al., 2000; James R. Lackner et al., 1999; Rabin et al., 2008), yet all stance conditions seem to reveal about the same benefit of reduced sway with FLT (Baldan et al., 2014). The experiments of this study utilized a more common, natural bipedal stance, as Achilles tendon vibration has been shown to produce A/P postural sway from this stance.

As for the configuration of the surface to be lightly touched, the effect of FLT reducing postural sway seems to be more pronounced when touching a horizontal surface (Houser, 2007) in the same plane as the greatest body sway (Bove et al., 2006). We utilized a configuration designed to maximize FLT postural control benefits in these experiments by touching a horizontal surface located directly in front of subjects in the sagittal plane, where the greatest sway would be expected. Height of the surface to touch in these experiments was set at each subject's shoulder height, which is higher than many prior studies. However, if the body behaves as an inverted pendulum during quiet stance with

movement primarily occurring about the ankles, then body sway is more pronounced the further (higher) you travel from the ankles, as Figure 2.3 demonstrates below. Therefore, a higher point of contact for FLT was utilized to allow more precise detection of sway, as movements would be larger the further from the ankles they are, and the fingertip might be able to detect the body sway earlier than receptors at the feet or ankles (John J. Jeka, 2016).

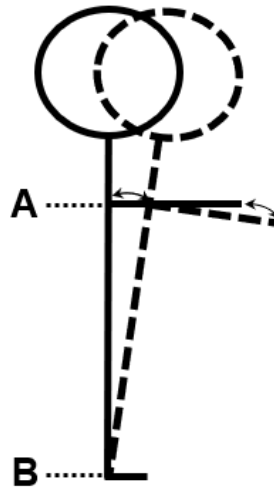


Figure 2.3 Depiction of greater motion occurring at level A than level B, with body sway acting as an inverted pendulum.

2.3.1 Light Touch Combined with Vibration

Some studies have tested effects of FLT on postural control when performed in combination with certain forms of mechanical vibration. Neck vibration can induce changes to postural control where the body tends to sway in a direction away from the side of the neck being vibrated (Gomez et al., 2009; Ivanenko et al., 1999; Verrel et al., 2011); however, under conditions with FLT, the effect of the neck vibration on postural control is reduced (Bove et al., 2006).

Similarly, trunk sway induced by lower back muscle vibration has also been shown to be reduced when touch conditions are also present in a seated position (Maaswinkel et al., 2014). One study did observe combinations of FLT with bicep muscle vibration while standing in a tandem Romberg stance. Instead of FLT reducing postural sway however, lateral drift of both the CoP and head were found to increase when FLT was combined with biceps vibration (Rabin et al., 2008).

A few studies have looked at the effects of FLT when standing postural control is perturbed by Achilles tendon vibration. Lackner and associates utilized a tandem Romberg stance with a vibrator attached to only the rear Achilles tendon. When both the Achilles vibration and FLT of a bar on the right side of the body were present, mediolateral (M/L) CoP motion was significantly reduced over conditions with the Achilles vibrator on without FLT. In fact, M/L CoP motion with the Achilles vibration and FLT was not significantly different from the condition with just FLT and no vibration present, suggesting that FLT was suppressing the abnormal postural effects of the Achilles vibration (J R Lackner et al., 2000). Consistent with Lackner's findings, one study found reduced A/P CoP excursions and velocities while standing in a normal tandem stance when conditions of FLT were provided along with vibration of both Achilles tendons, compared to conditions with Achilles vibration and no FLT (Houser, 2007). Another study also found similar results when utilizing a normal tandem stance, vibrators on both Achilles tendons, and light touch of a surface in front of subjects. Although the study was primarily focused on changes in anticipatory

postural adjustments when performing an arm raise and load dropping task, the effects of Achilles vibration on the anticipatory postural adjustments were reduced when FLT was also provided (Harm Slijper & Latash, 2004). These prior studies observed the combined effects of both Achilles vibration and FLT on postural control, but none have investigated how the postural effects of Achilles vibration might be mitigated by light touch with the addition of small amounts of mechanical noise added to the fingertip.

Two studies have demonstrated that the enhancement of postural control in young healthy subjects by FLT can be further improved by adding small amounts of mechanical noise to the fingertip that is lightly touching the stable surface (Kimura et al., 2012; Magalhães & Kohn, 2011b). Moreover, it is hypothesized that the added postural benefits from the fingertip noise could be even more prevalent in blind individuals (Magalhães & Kohn, 2011a). These findings of fingertip noise further enhancing the postural benefits of FLT have been attributed to a phenomenon known as stochastic resonance, which is discussed in the next section.

2.4 Stochastic Resonance

The term “stochastic resonance” (SR) was coined more than three decades ago as a means to explain the periodic jumping between climate states (R Benzi, Sutera, & Vulpiani, 1981; Roberto Benzi, Parisi, Sutera, & Vulpiani, 1982; Nicolis, 1981, 1982). SR can essentially be thought of as a phenomenon where a certain amount of noise in a nonlinear system is capable of enhancing signal transmission or detection (McDonnell & Abbott, 2009). Nomenclature of

the term has come under some scrutiny in debate as to whether the occurrence should be classified as a true, “bona-fide” resonance (Choi, Fox, & Jung, 1998; Evstigneev, Reimann, Schmitt, & Bechinger, 2005; L Gammaitoni, Marchesoni, & Santucci, 1995; Luca Gammaitoni, 1995; Giacomelli, Marin, & Rabbiosi, 1999; Marchesoni, Gammaitoni, Apostolico, & Santucci, 2000). Semantics aside however, numerous studies have found the detection of weak periodic signals to be enhanced by the addition of certain non-zero levels of noise (McDonnell & Abbott, 2009).

The idea of “good noise” is somewhat counterintuitive for signal detection, as noise is commonly viewed as a nuisance that degrades signal characteristics. The SR phenomenon clearly suggests though that a certain optimal amount of noise can improve signal detection. Thus, SR is commonly observed in results where an inverted-U shape can be seen in performance measures (Douglass, Wilkens, Pantazelou, & Moss, 1993; Lin, Liu, Liu, & Gao, 2015; E Manjarrez, Diez-Martínez, Méndez, & Flores, 2002; McDonnell & Abbott, 2009; Mendez-Balbuena et al., 2012). Findings with this occurrence indicate that with too little noise, signal detection is not improved. Likewise, a level of noise too high will degrade detectability of a signal; however, there lies an amount of noise in the middle where maximum enhancement of performance or signal detection can be achieved, as Figure 2.4 shows below.

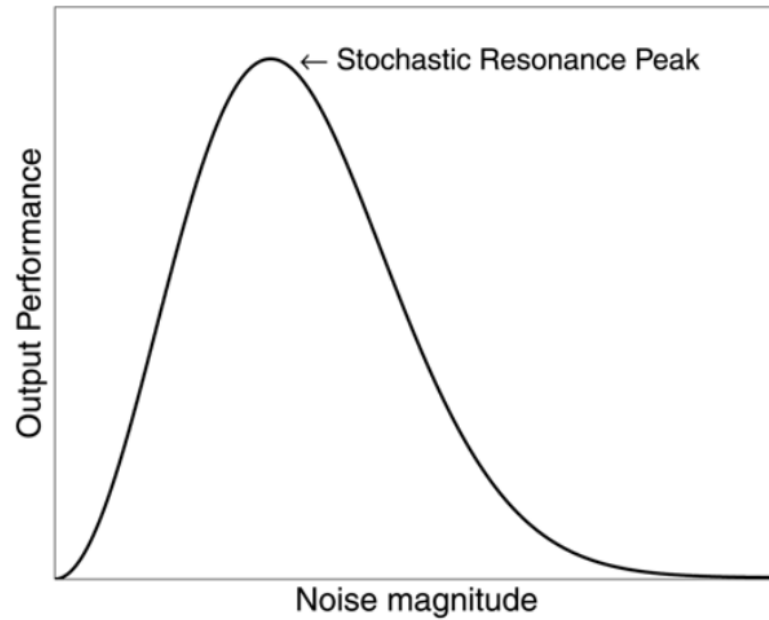


Figure 2.4 Depiction of a typical inverted-U shaped curve seen in performance measures from several SR studies (McDonnell & Abbott, 2009).

SR is believed to occur in a multitude of physical systems, including several physiological applications that have been studied, such as sensory neurons and receptors (Douglass et al. 1993; Moss et al. 1994; Wiesenfeld and Moss 1995; Collins et al. 1996a; Levin and Miller 1996). The remainder of this section primarily focuses on physiological studies, which seem to observe the SR phenomenon occurring in human sensory systems, especially those that are attributed to postural control (vision, vestibular, and somatosensory).

2.4.1 Sympathetic Nervous Responses to Noise

A few studies have noticed improved sympathetic nervous system responses within the human body attributed to SR (Hidaka et al., 2001; Hidaka, Nozaki, & Yamamoto, 2000; Yamamoto et al., 2002). The sympathetic nervous

system is responsible for regulation of blood pressure during periods of hypovolemic stress, such as transitions from a lying to a standing position. Without an appropriate sympathetic response, such a transition could lead to orthostatic intolerance and syncope, causing a fall. Therefore, populations commonly experiencing orthostatic intolerance, such as astronauts returning from spaceflight (Convertino, 1996; Karmali & Shelhamer, 2010; Spiering et al., 2011; S. J. Wood, Loehr, & Guilliams, 2011), may stand to benefit from a countermeasure utilizing SR to improve sympathetic nervous system responses.

In 2000, Hidaka demonstrated that noise could improve human blood pressure regulation. Various levels of noise were added to carotid artery baroreceptors through a pneumatic neck chamber while healthy subjects were subjected to oscillating tilts ranging 0°-20° from a supine position. The oscillatory tilts were designed slow enough to not induce heart rate and blood pressure responses; however, at certain levels of added noise the baroreflex response was shown to be enhanced. The enhancements to the response seemed to reveal the conventional inverted-U shaped curve for performance as the noise levels increase, which is commonly noted during SR. Important to note in this study was their conclusion that SR was likely occurring in the brainstem rather than the peripheral baroreceptors, suggesting that added noise is capable of evoking SR within the central nervous system as opposed to just the periphery where it is introduced (Hidaka et al., 2000).

Hidaka again reported improved sympathetic responses to baroreceptor noise added via a pneumatic neck chamber in 2001. Instead of a tilt platform

however, oscillator lower-body negative pressure (LBNP) was utilized to perturb central venous pressure while providing negligible effects on the arterial system. Muscle sympathetic nerve activity (MSNA) was also recorded from the peroneal nerve. The results indicated that heart rate, cardiac interbeat interval, and MSNA were all significantly improved by the addition of certain amounts of noise. It was again concluded that SR was likely occurring in the brain stem, where inputs from arterial and cardiopulmonary baroreceptors first interact (Hidaka et al., 2001).

In 2002, Yamamoto reported that patients with primary autonomic failure (PAF), a condition where the autonomic nervous system does not function properly, were able to improve postural hypotension to head-up tilts. Five subjects with PAF and eight healthy controls were exposed to 30° or 60° head-up tilts while also receiving Gaussian white noise delivered through a pneumatic neck chamber. A condition of continuous positive pressure was also introduced to examine potential hypertensive effects. Although the noise did not restore systolic blood pressure (SBP), diastolic blood pressure (DBP), and heart rate metrics in PAF patients to levels of healthy controls during head-up tilt, the metrics were significantly improved over the noise and often compression conditions. Thus, this research became one of the first studies to demonstrate that externally applying noise to sensory receptors can be used for therapeutic purposes in treating individuals with neurological disorders (Yamamoto et al., 2002).

2.4.2 Visual Responses to Noise

Several studies have observed how the addition of external noise can potentially impact the visual system. Although to date there is little evidence of noise enhanced vision via SR leading to balance improvements, the importance of the visual system in postural control is well established (Berthoz, Lacour, Soechting, & Vidal, 1979; D. N. Lee & Lishman, 1975; David N. Lee & Aronson, 1974; S. R. Lord, 2006; S. R. Lord & Dayhew, 2001; L. Nashner & Berthoz, 1978; R. Wood, 1895). It is noted that some individuals may place a greater importance on visual information than others during balance maintenance and perception of verticality (Brady et al., 2012; Einarsson et al., 2018; Fay B. Horak et al., 2016; Isableu et al., 2010; Witkin & Asch, 1948b; Woollacott, Debú, & Mowatt, 1987). Visual impairments can surely impact postural control in a negative manner (Coleman et al., 2004; Edwards, 1946; Kasuga, Aruga, Ono, Hiratsuka, & Murakami, 2017; Skalska et al., 2013). Therefore, it is reasonable to assume that balance could be improved by enhancing visual input through the SR phenomenon.

In 1997, Simonotto published a series of experiments adding noise to the visual systems of humans. An image digitized on a grey scale was depressed so that subjects couldn't perceive the picture. When various levels of noise were added to the gray value of every pixel, a certain level of optimal noise allowed the subjects to best see the picture's detail. They also found that a mid-range level of grayscale noise improved the threshold at which subjects could distinguish a pattern of lines. The tests used were found to be fairly robust and repeatable, and it was concluded that their experiments demonstrated how SR could serve

as a measure of efficiency of how the visual system processes noise (Simonotto et al., 1997).

Kitajo and associates in 2003 used the eyes as a means to provide a double receptor design to explore SR effects. Most SR studies have utilized a single receptor design in that both the noise and the weak stimuli that is trying to be detected are applied to the same receptor area. Kitajo's double receptor design added visual noise to one eye, while the other eye received a weak visual stimulus (a slowly changing gray level). This design ensures that interaction between the noise and the stimulus should not occur until afferents from both reach the central nervous system (similar to the logic used in Hidaka's studies with baroreceptors). Subjects were able to improve detection of the stimulus when noise was applied to the opposite eye, thus it was concluded that binocular interaction of the noise and stimuli must have been occurring at higher levels in the visual system within the brain, rather than in the periphery (Keiichi Kitajo et al., 2003). Kitajo used his visual double receptor design again in other studies. With an analogous protocol in 2007, it was found that phase synchronization of electroencephalogram (EEG) signals was increased along with the noise-enhanced performance in visual detection, providing further evidence of SR occurring within the brain (K Kitajo et al., 2007). In 2008, internal noise levels of subjects were assessed when using the dual receptor design. Internal noise was defined as being fluctuations in behavior in the absence of the externally added noise. It was found that the internal noise levels of subjects seemed to also have an effect on SR occurring through the addition of external noise. Indeed, it

seemed that the lower the internal noise rate, the higher the effect of SR when external noise was applied (Aihara, Kitajo, Nozaki, & Yamamoto, 2008).

Kitajo's studies may not necessarily be the first evidence of SR occurring within the brain. Prior to the discovery of SR it had been noted that critical flicker frequency (CFF), the frequency at which the blink of a flickering light is no longer distinguishable, can be altered when listening to concurrent auditory noise (Harper, 1979). Moreover, peak flicker sensitivity when listening to various levels of audible white noise has been reported to occur in an inverted-U shape (Harper, 1979), a noted characteristic of the SR phenomenon. If these earlier findings were attributable to SR, they would also be evidence of it occurring within the brain as well as across different modalities (i.e. vision and hearing). In 2007 Manjarrez expanded on this earlier research. He had subjects indicate when they could detect brief light flashes (stimuli) while listening to various levels of auditory white noise. When the stimuli were subthreshold, a certain middle range of auditory noise seemed to improve detection of the stimuli in subjects the best. The inverted-U shape of these results was indicative of the SR phenomenon. However, when three additional subjects were tested while receiving the stimuli at suprathreshold levels, detection of the flashes seemed to degrade as auditory noise levels increased. It was concluded that SR likely caused improvements in detection when the signal was delivered at subthreshold levels; however, for a strong visual signal the auditory noise may act in detriment to detection (Elias Manjarrez, Mendez, Martinez, Flores, & Mirasso, 2007). Interestingly to note though, the addition of auditory noise has also been shown

to improve postural control in subjects, even under conditions without vision (J. M. Ross, Will, McGann, & Balasubramaniam, 2016; Jessica Marie Ross & Balasubramaniam, 2015), suggesting that the possible SR induced benefits are not limited to auditory and visual connections within the brain.

Sasaki published research in 2006 that added both Gaussian white noise and weak periodic light signals to the dominant eyes of twenty-two male subjects. Results showed that a certain amount of noise, usually around the threshold detection level for the noise, improved the threshold detection of the weak periodic signals (Sasaki et al., 2006). Sasaki later examined these findings again in 2008 and confirmed that detection rates of the periodic signals were indeed improved beyond the level of chance when a certain subthreshold level of white noise was added; however, too much noise degraded detection (inverted-U shape in performance). Furthermore the subthreshold noise was also found to facilitate visual pattern discrimination in a second experiment (Sasaki et al., 2008). Occurrence of SR in the CNS or periphery cannot be determined in these experiments due to the single receptor design; however, they do provide evidence of noise enhancing visual perception, even if that noise is delivered at a subthreshold level.

In 2007 Funke used image jitter noise to improve visual input in cats. The cats were anesthetized and action potentials from the primary visual cortex were recorded while they looked at a stimulus with image jitter added (noise). Results tended to show visual responses (recorded action potentials) were enhanced by low to moderate levels of tremor noise, suggesting enhanced signal detection

reaching the brain that was attributed to SR (Funke, Kerscher, & Wörgötter, 2007).

Loader also utilized stochastically moving visual stimuli to potentially evoke SR in 2007. He took twenty-four subjects with unilateral vestibular hypofunction and divided them into a group receiving computerized optokinetic therapy (COKT) and a control group receiving no therapy. The COKT group received ten training sessions where they read text that was stochastically moving. Sensory organization tests (SOTs), a common posturography test that manipulates visual and somatosensory inputs during quiet stance, were performed on both groups pre- and post-training. Within the COKT group, subjects significantly improved some pre- to post-training SOT scores, and between group analysis also revealed significantly better post-test SOTs in the COKT group over the control (Loader et al., 2007). The exact mechanisms by which the postural improvements occurred in this study could use further investigation, but it does seem to be one of the first studies suggesting that balance maintenance can be improved by evoking SR through the introduction of visual noise.

More recently one study has noted the potential for SR to help the visually impaired. Itzcovich had fourteen subjects with severe visual impairment identify alphabet characters with various levels of grayscale noise added. The fraction of recognized letters significantly increased with the addition of noise. Moreover, improvements seemed to show a clear inverted-U shape where letter recognition increased along with increasing noise levels to a peak, at which point greater

amounts of noise diminished performance. Results suggest that SR may be capable of improving vision aids utilized by the visually impaired (Itzcovich, Riani, & Sannita, 2017). As future studies continue to test the use of SR for vision enhancement in the visually impaired, the impact on postural control should also be assessed.

2.4.3 Vestibular Responses to Noise

Externally added noise has also been shown to impact the effect of vestibular afferents contributing to balance maintenance. Most commonly, noise is delivered to the inner ear through stochastic electrical current passing between two electrodes placed over the mastoid bones, as has been done with several Galvanic vestibular stimulation (GVS) studies (Dilda, Morris, Yungheer, MacDougall, & Moore, 2014; Fitzpatrick et al., 1996; Pavlik et al., 1999). A few studies have shown higher stochastic currents can decrease postural control (MacDougall et al., 2006; Moore et al., 2006; Pavlik et al., 1999; Anthony P Scinicariello et al., 2002). However, it has also been observed that prolonged exposure to certain levels of noisy GVS might lead to improved adaptive capabilities, where the importance of vestibular afferents may be down-weighted in favor of more reliable inputs (Dilda et al. 2014), especially under novel vestibular challenges (Moore, Dilda, Morris, Yungheer, & MacDougall, 2015). There are also a few studies that have noted improved postural control under non-noisy GVS conditions (Krizková & Hlavacka, 1994; A P Scinicariello et al., 2001). Of particular interest in this section however, are GVS studies that have suggested applying small amounts of noisy current (often below what can be

perceptually detected) as a means of evoking SR induced improvements in vestibular afferent signals, which seem to contribute to better balance performance and potentially an up-weighting of vestibular afferents.

One of the first studies to note balance improvements when introducing stochastic GVS was done by Pal in 2009. PD patients and healthy subjects balanced on a foam compliant surface while being exposed to noisy GVS at various amplitudes. Configuration of the GVS electrodes allowed for current flow in the A/P direction with two cathodal electrodes on the mastoid process and one anode placed on the C7 vertebra. When PD patients' eyes were closed, a small (4.5%) but significant reduction in A/P sway was noted with low intensity stochastic GVS. Effects of the GVS noise on postural control were not noted in the healthy subjects however. It was concluded that the sway reductions seen in the PD patients may have been due to SR enhancing the detection of weak vestibular stimuli (Pal et al., 2009). Pal's study was not the only one to find improved postural control in PD patients utilizing stochastic GVS. In 2015 Samoudi and associates found a broadband (0-30 Hz), noisy GVS signal to significantly improve static postural control in PD patients (especially when off LDOPA medication), as well as significantly decrease correction time when posture was perturbed by the release of a weight. Maximum amplitude of the GVS signals was individually set to the perceptual threshold level for subjects, thus making the majority of the noise signal subthreshold and consequently undetectable to the patients (Samoudi et al., 2015).

In 2011, Mulavara also tested the effects of noisy GVS on postural control in healthy subjects with their eyes closed on a foam compliant surface. The more common binaural configuration with two electrodes placed on the mastoid processes was utilized though to provide current in the M/L direction. Furthermore, two separate bipolar stochastic electrical signals (0-30 Hz and 1-2 Hz) were utilized at seven different amplitude levels (all at current peaks less than 1 mA). The broadband (0-30 Hz) and narrowband (1-2 Hz) signals were both examined because although most studies investigating contributions of the vestibular system to postural sway have found greater coherence between the two at frequencies less than 5 Hz (Pavlik et al., 1999; Anthony P Scinicariello et al., 2002), coherence between stochastic vestibular stimulation and lower-body muscle activation has also been found to occur in the 0-20 Hz frequency range (Christopher J Dakin et al., 2007). Contrary to Pal's findings, Mulavara was able to find significantly less M/L sway at an optimal amount of noisy GVS than without for the healthy subjects. Both the broadband and narrowband signals revealed postural improvements, with a slightly larger effect seen using the broadband (0-30 Hz) signal (A. P. Mulavara et al., 2011). Interestingly though, later analysis including some of this data showed that the common M/L current configuration was capable of improving postural control sway metrics in the A/P direction as well (Goel et al., 2015). Improved postural control has also been recently observed on a stable surface with the eyes open, when healthy subjects were exposed to noisy GVS (Inukai et al., 2018).

Improvements in postural control for more dynamic tasks such as locomotion have also been observed when utilizing stochastic GVS. Using the same configuration and broadband noise as in his previous studies (Goel et al., 2015; A. P. Mulavara et al., 2011), in 2015 Mulavara reported significantly improved walking stability in healthy subjects. Furthermore, the improvements seemed to follow the typical inverted-U shape indicative of the SR phenomenon, where a certain middle levels of noise most improved the dynamic stability and higher amplitudes of noise revealed less enhancement (A. Mulavara et al., 2015). A year later, Wuehr reported significantly improved bilateral coordination and less gait variability when healthy subjects used noisy GVS and walked with their eyes closed. The improvements were seen at slower walking speeds, as they were hypothesized to be more reliant on sensory feedback control (Wuehr et al., 2016).

Both Wuehr and Mulavara showed that immediate improvements in locomotor performance are capable when applying subthreshold, broadband stochastic GVS, seemingly through the improvement of weak vestibular afferent detection and signal transfer occurring via SR (A. Mulavara et al., 2015; Wuehr et al., 2016). There is also evidence of subthreshold GVS noise enhancing locomotor performance in an adaptation paradigm. We had healthy subjects perform repeated functional mobility tests (FMTs) on a foam compliant surface under conditions of visual discordance, thereby ensuring that the only fully intact sensory system contributing to dynamic balance performance was the vestibular system. Subjects receiving subthreshold noisy GVS had significantly faster adaptation rate increases for time to complete the FMT course than did those receiving sham GVS.

These results suggest that stochastic GVS may be a means by which to improve short-term locomotor adaptation to a novel sensory discordant environment through SR induced vestibular enhancement (Temple et al., 2018).

2.4.4 Somatosensory Responses to Noise

Many studies have noted improvements in the detection of weak somatosensory stimuli attributed to the addition of noise. Improved cutaneous mechanoreceptor transmission with the addition of small amounts of noise has been attributed to SR in animal studies on crayfish (Douglass et al., 1993), rats (Collins et al. 1996a; Ivey et al. 1998), and cats (Elías Manjarrez, Rojas-Piloni, Méndez, & Flores, 2003). Indeed, human studies have also reported improved detection and transmission of weak somatosensory stimuli with the addition of noise applied to muscle spindles (Cordo et al., 1996), mechanoreceptors (Collins et al. 1996b; Collins et al. 1997; Dhruv et al. 2002; Liu et al. 2002; Manjarrez et al. 2002; Richardson et al. 1998; Wells 2002), and peripheral nerves (Iliopoulos, Nierhaus, & Villringer, 2014). Of particular interest in this dissertation however, are somatosensation enhancements attributed to SR that result in improved human postural control.

The efficacy of externally delivered noise's capability to improve balance in the elderly is somewhat established. Numerous studies have reported improved postural control in elderly subjects when noisy stimuli have been provided to somatosensory afferents through both mechanical (Dettmer et al., 2015; Lipsitz et al., 2015; A. Priplata et al., 2003; A. Priplata et al., 2002; Zhou, Lipsitz, Habtemariam, & Manor, 2016) and electrical (Gravelle et al., 2002)

means. In 2002, Priplata used a matrix of nylon indentors to deliver mechanical white noise to the bottom of healthy elderly and young subjects' feet, at levels just below perceptual threshold. The subthreshold noise was found to significantly improve postural control metrics in both the elderly and younger subjects (A. Priplata et al., 2002). A year later, similar results were found when Priplata used subthreshold vibratory noise on the plantar foot surface at the forefoot and heel with insoles. Although reductions in sway parameters were noted in both younger and elderly subjects, the greater SR response resulting in enhanced postural control tended to occur with the elderly population (A. Priplata et al., 2003). Similar findings have been more recently reported by Dettmer during a sensory conflict postural control task. Dettmer too used vibrating foot insoles and found them to significantly improve sway metrics in the elderly under conditions where the visual scene provided erroneous postural orientation input. No significant changes in metrics were noted for the younger population however, when utilizing the noisy vibration stimulus (Dettmer et al., 2015). A few other studies have noted improved postural control in elderly subjects when exposed to vibration noise delivered to the plantar surface of the feet (Costa et al., 2007; Lipsitz et al., 2015; Wanderley, Albuquerque-Sendon, Parizotto, & Rebelatto, 2011; Wang & Yang, 2012; Wei et al., 2012; Zhou et al., 2016). Moreover there is evidence such mechanical noise devices can also provide improved postural control during locomotion as well in the elderly (Galica et al., 2009; Lipsitz et al., 2015; Stephen et al., 2012), perhaps even allowing for better capacity to adapt to physical and cognitive stressors through increased

postural complexity (Costa et al., 2007; Zhou et al., 2016). A systematic review on the beneficial effects of foot vibration on balance and locomotion in the elderly has recently been published (Aboutorabi, Arazpour, Bahramizadeh, Farahmand, & Fadayevatan, 2017).

Along with the elderly, improved postural control has also been noted in populations known to exhibit diminished cutaneous sensation with the addition of small amounts of mechanical noise delivered through vibrating insoles (A. A. Priplata et al., 2006). In 2002, Liu demonstrated that elderly, patients with stroke, and those with diabetic neuropathy were able to improve fingertip tactile sensitivity with the addition of small amounts of mechanical noise. Furthermore, diabetic neuropathy patients were also tested at the metatarsal head of the foot and found similar enhanced sensory detection with the noise (Liu et al., 2002). A year later Khaodhri additionally reported improved tactile sensation on the bottom of the foot in those with diabetic neuropathy when subthreshold vibration noise was also present (Khaodhri et al., 2003). These findings led Priplata to test the effect of vibrating insoles on postural control in stroke and diabetic patients in 2006. When application of the vibratory noise was set at 90% of the subjects' sensory thresholds for the stimulus, several sway parameters were significantly improved. Moreover, a relationship between sensory impairment and balance enhancement was also observed, where those who had greater sensory impairment experienced more improvement in postural sway when the subthreshold noise was present (A. A. Priplata et al., 2006).

In addition to mechanical noise added at the foot, improved postural control has also been noted with noise added to the fingertip. In 2011, Fernando Magalhães reported significantly improved postural control when certain levels of vertically oriented vibration noise were added to the fingertip of healthy subjects, compared to conditions where just light touch was utilized. Multiple levels of fingertip vibration were utilized and tended to reveal the typical inverted-U shape indicative of the SR phenomenon, where sway metrics improved as noise amplitude increased to a peak, after which the addition of higher levels of external noise was associated with poorer sway metrics. Furthermore, as the levels of fingertip vibration noise were not delivered at subthreshold levels, the idea that postural control could have been improved by increased attention to the task was also tested. They had three subjects perform the task with the vibration stimulus fixed to their body. In these conditions the postural improvements were not noted, indicating that SR was the more likely cause of the noise induced postural improvements observed (Magalhães & Kohn, 2011b). The following year, a similar postural control study was published that utilized subthreshold vibration noise delivered to the fingertips of healthy individuals. Noise delivered at about 50% of the perceptual threshold significantly improved the sway metrics over both the no noise and noise delivered at the level of perceptual threshold conditions (Kimura et al., 2012). Along with these standing postural control tasks, Mendez-Balbuena also reported improved sensorimotor performance in a seated task where subjects were required to compensate static forces generated by a manipulandum. Certain levels of fingertip vibration noise significantly

reduced mean variation in the task over both zero and high noise conditions. It was concluded that SR induced increases in peripheral sensory receptor sensitivity as well as corticomuscular synchronization likely led to the better sensorimotor integration and improved performance (Mendez-Balbuena et al., 2012).

Despite the numerous studies showing SR induced somatosensory enhancement leading to improved postural control, SR effects under conditions where another sensory conflict is present is rarely investigated. Even for those studies that have (Dettmer et al., 2015; Temple et al., 2018), the conflicting stimuli detrimental to balance performance usually does not occur within the same sensory modality attempting to utilize SR induced sensory enhancement. Further studies need to be done to see if small amounts of external noise that usually evoke SR enhance, counteract, or have no effect on balance perturbing stimuli delivered within the same sensory modality.

2.4.5 Factors Potentially Influencing SR Response

Although most published research investigating SR in the nervous system tends to demonstrate enhancement of weak sensory stimuli with the addition of small amounts of noise, these results are not always the case. Some subjects do not appear to be responsive to external noise or demonstrate improvements in performance attributable to SR (Collins et al. 1996b; Mulavara et al. 2011; Dettmer et al. 2015; Goel et al. 2015; Temple et al. 2018). Even when testing individual receptors, increased sensitivity to stimuli with a certain level of nonzero noise is commonly unobserved in a few cells (Douglass et al. 1993; Collins et al.

1996a). As a result, there is speculation as to why the SR effect is not always evident with the addition of noise.

It is possible that in a few studies, the optimal amount of external noise was not delivered to induce SR. As has been stated, performance attributed to SR often follows an inverted-U shape as the level of added noise increases. Therefore, if the external noise is applied at a level too small, it may not be enough to see performance improvements. Likewise, if the level of noise is too high, it can degrade the ability to detect weak stimuli. Finding the right level of noise to induce SR is often a challenge researchers face, as optimal levels of noise seem to vary between subjects (Goel et al., 2015).

One of the challenges to finding the optimal level of noise may lie within the neurons themselves. Neurons are inherently noisy, and this level of internal noise already within the system may impact the ideal level of external noise needed to observe the SR phenomenon. It has been noted that a negative relationship may exist between internal noise levels and the SR effect when external noise is added. That is, higher internal noise levels tend to correspond to less of an SR effect (Aihara et al., 2008; Douglass et al., 1993). It has been proposed that the decreased SR effects seen when internal noise is high may be due to the internal noise levels already being at or above the optimal level of noise for SR, thus adding more external noise can degrade signal detection and performance (Aihara, Kitajo, Nozaki, & Yamamoto, 2010).

In addition to internal noise levels potentially impacting SR effects, the type of subjects recruited for a study can also play a role. Often the SR response

is noted to occur on somewhat of a continuum with the sensory capabilities of the subjects. That is, those with diminished sensory capabilities tend to exhibit a greater SR response than individuals with normal functioning sensory systems (Bieze, 2004; Inukai et al., 2018; Liu et al., 2002). This finding has been evident in numerous balance studies where patients with stroke, diabetic neuropathy, or the elderly have seen greater SR induced postural control improvements than healthy controls (Dettmer et al., 2015; A. A. Priplata et al., 2006; A. a Priplata et al., 2003; A. Priplata et al., 2002). It has also been shown that healthy subjects with less stable balance tend to show a greater SR effect than healthy subjects with more stable balance (Inukai et al., 2018).

2.5 Metrics to Assess Postural Control

To assess postural control, studies often utilize kinetic metrics that deal with the forces causing body motion, as well as kinematic metrics that describe characteristics of the motion (Hamill, Knutzen, & Derrick, 2015). The next two sections of this proposal (2.5.1 and 2.5.2) will discuss the different kinetic and kinematic metrics that were utilized to assess postural control during this study.

2.5.1 Kinetic Metrics

One commonly utilized method to evaluate the postural control and balance capabilities of individuals during quiet stance involves using force plates to calculate CoP as the exerted forces of the feet on the support surface. Typically, research indicates that increased CoP displacements and motion across time is associated with poorer balance and postural control (Hatzitaki et al., 2004; Thompson et al., 2011). It is also commonly assumed that with greater

metrics of CoP, fall risk is likewise increased (Melzer, Benjuya, & Kaplanski, 2004). This study utilized four metrics taken from the A/P CoP data: 1) mean position (POS_{CoP}), 2) mean-removed root mean square (RMS_{CoP}), 3) mean speed (SPD_{CoP}), and 4) approximate entropy ($ApEn_{CoP}$).

POS_{CoP} indicated on average where the CoP was positioned, with positive values indicating more anterior positions and negative values indicating more posterior positions. Greater values (whether positive or negative) tended to be indicative of poorer postural control, in that they show the CoP was shifting away from the central position in which it began.

RMS was computed as the square root of the average squared deviations of A/P position from the origin (Temple et al., 2014, 2016), which then had the absolute value of the mean position ($|POS_{CoP}|$) subtracted from it to provide the demeaned value. It was calculated by equation 2.5.1 below where x was the A/P displacement from the origin at time point t , N was the total number of time points, and \bar{x} was the mean A/P displacement position (POS_{CoP}).

Equation 2.5.1

$$RMS = \sqrt{\frac{\sum_{t=1}^N x_t^2}{N}} - |\bar{x}|$$

RMS provided information about the amplitude of the A/P CoP motion about the POS_{CoP} . In general, the greater the RMS, the more A/P CoP motion there was about the POS_{CoP} , and the poorer the postural control.

Mean speed (SPD) is a common metric utilized to assess CoP motion (Goel, 2017; Houser, 2007; Kabbaligere et al., 2017). In general, with more

speed, the faster the CoP motion was, which also could indicate poorer postural control (Goel, 2017; Houser, 2007). SPD_{CoP} served as an indicator of how fast CoP motion was occurring on average during a trial.

Approximate entropy (ApEn) is a nonlinear metric of regularity that has previously been utilized to characterize CoP motion (Cavanaugh et al. 2005a; Cavanaugh et al. 2005b; Cavanaugh et al. 2006; Cavanaugh et al. 2007; Dettmer et al. 2015). ApEn utilizes a moving window procedure to determine the probability of data sequences being repeated. The unit-less numbers produced range from zero to two, with a value of zero indicating a completely regular time series (e.g. sine wave), and a value of two indicating a completely random time series (e.g. Gaussian noise) (Cavanaugh et al. 2005b). We utilized ApEn to determine if CoP motion regularity was affected by the various somatosensory stimuli imposed in these three experiments. Previous studies calculating ApEn of CoP motion have suggested that the metric can detect complexity of the postural control system as a whole, such that greater values revealing more irregularity are also indicative of greater complexity, less system constraint, and more degrees of freedom being utilized in system control (Haran & Keshner, 2009; Newell & Molenaar, 1998). Essentially, more complex postural control systems have more interactions between system components and the environment, which can manifest in a more chaotic system output signal, such as the CoP trajectory (Haran & Keshner, 2009). CoP ApEn has been shown to be reduced in concussed patients (Cavanaugh et al., 2006; Cavanaugh, Guskiewicz, Giuliani, et al., 2005; Cavanaugh, Guskiewicz, & Stergiou, 2005), as well as in children

and the elderly who exhibit greater freezing of degrees of freedom (Newell & Molenaar, 1998). Thus, it is inferred that higher CoP ApEn values in the current study might reveal greater postural complexity and be indicative of better postural control.

2.5.2 Kinematic Metrics

In addition to CoP data, kinematics of body segment A/P angular motion was utilized in analysis of postural control. Specifically, changes in sagittal plane angles at the head [flexion(+)/extension(-)], torso [flexion(+)/extension(-)], thigh [knee flexion(-)/extension(+)], and shank [ankle dorsiflexion(+)/plantarflexion(-)] with respect to vertical were computed and normalized by demeaning the values by the POS of the first five seconds of baseline data for each trial. Additionally, flexion(+)/extension(-) angles of the right elbow were also utilized for experiment #3, as this joint was likely to see changes in angle positions with combinations of FLT and arm tendon vibration, as previously noted (Rabin et al., 2008; Roll & Vedel, 1982). POS, SPD, and RMS of the displacements of these angles were calculated and utilized as metrics of postural control, as has been done previously (Kabbaligere et al., 2017; B.-C. Lee, Martin, Ho, et al., 2013). As with the other metrics, greater values tend to indicate more angular movement and consequently poorer postural control.

Means and standard deviations of the four A/P body segments angles with respect to vertical (head, torso, thigh, and shank) were also utilized to calculate the Anchoring Index (AI) for the head, torso, and thigh. AI is a metric that can be used to quantify the stabilization of a given body segment with respect to its

inferior segment, as well as to external space (Amblard et al., 2001; Assaiante & Amblard, 1993; B.-C. Lee, Martin, Ho, et al., 2013; Sveistrup et al., 2008). It was calculated by equation 2.5.2 below where $\sigma_{(s-i)}$ is the standard deviation of the angular displacements achieved when subtracting the angular displacements of the inferior segment (*i*) from the angular displacements of the superior segment (*s*), and σ_s is the standard deviation of the angular displacements of the superior segment.

Equation 2.5.2

$$AI = \frac{\sigma_{(s-i)} - \sigma_s}{\sigma_{(s-i)} + \sigma_s}$$

Negative AI values indicated more predominant stabilization of the superior segment to the inferior segment than to external space, while positive values showed more predominant stabilization of the superior segment to external space than to the inferior segment (Amblard et al., 2001; Assaiante & Amblard, 1993; B.-C. Lee, Martin, Ho, et al., 2013; Sveistrup et al., 2008).

III. EXPERIMENTAL METHODS

The experiments in this study were conducted in compliance with policies and practices concerning the protection of human subjects at the University of Houston (UH). The study gained approval from the UH IRB Committee 1, which is responsible for the protection of human subjects and conforms to the Declaration of Helsinki. Written informed consent was obtained from each subject prior to testing, and subjects were free to leave or cease testing at any time during the study, should they have chosen to.

3.1 Subjects

Twenty-five healthy, young adults between the ages of 18 and 35 years were recruited as subjects for these experiments. Subjects were recruited with flyers (see Appendix A) and by word of mouth from the UH main campus, including students, faculty, and staff members. All subjects were free of any known neurological or musculoskeletal impairments that may have affected postural control. Attempts were made to recruit equal numbers of male and female subjects in order to minimize any effect of gender bias on results. Every subject performed all conditions for all three experiments within a single, two-and-a-half-hour testing session.

Sample size for the study was calculated a priori using G*Power. Based on previous research investigating the effect of FLT combined with noise on postural control (Kimura et al., 2012), with significance set at $\alpha = 0.05$, 94% power, and an effect size of $f = 0.34$, a total of twenty subjects were calculated to be required for this study to see a significant main effect of fingertip noise (SR)

on postural control (SPD_{CoP} and RMS_{CoP}). An additional five subjects were recruited for this study in case individuals choose to drop out from participating, which none did.

3.1.1 Inclusion Criteria

Subjects for this study needed to have intact, normal neurological and musculoskeletal functioning to perform the experimental protocols. All subjects had to be free of lower-body muscle and joint injuries and abnormalities. A modified Physical Activity Readiness Questionnaire (PAR-Q) was utilized for subjects to self-report any known neurological and musculoskeletal issues that might have affected balance maintenance (see Appendix B).

The subjects also needed to be within the age range of 18-35 years. They had to be at least 18 years of age and a legal adult capable of providing written informed consent to participate in the study. The age limit of 35 years was to mitigate the chance of age-related sensory declines or physical performance capabilities affecting balance maintenance in subjects.

Finally, subjects had to display the typical postural response observed when Achilles tendon vibration was present during quiet stance, in order to ensure that the tendon vibration was providing a perturbing stimulus so that the body might reweight afferents accordingly in conjunction with other types of somatosensory stimuli. Each potential subject was exposed to a few seconds of Achilles tendon vibration with their eyes closed. If there was a noticeable, typical posterior sway produced when the vibration was turned on, then the person was

allowed to participate as a subject. All subjects recruited displayed this typical Achilles tendon response.

3.1.2 Exclusion Criteria

Subjects were excluded if they self-reported any physiological problems on the modified PAR-Q that might have led to alterations in typical balance performance. If a potential subject answered “Yes” to any questions on the PAR-Q, other than question #7, they were excluded from participating. Question #7 of the PAR-Q regarded knowledge of vision problems. Potential subjects that indicated they had vision acuity problems that were corrected with contacts or glasses on question #7 of the PAR-Q were still be eligible to participate in the study. Subjects had to fall within the established 18-35 year age range at the time of testing, or they were excluded from the study. Finally, the noticeable postural response to Achilles tendon vibration, which produced posterior sway when the vibration was activated with the eyes closed had to be present. Any potential subject who did not display this typical postural response to Achilles tendon vibration would have been excluded; however, all potential subjects did display this response.

3.2 Environment

Each subject was tested during a single, two-and-a-half-hour session at the Center for Neuromotor and Biomechanical Research (CNBR), located in the Health 2 building at UH.

3.3 Instrumentation

3.3.1 Force Plates

A split-belt treadmill instrumented with two force plates (Bertec Fully Instrumented Treadmill, Columbus, OH, USA) was utilized to collect ground reaction forces for the computation of CoP. The treadmill was kept stationary during the entire testing session. Thus, it served as just a platform for subjects to stand on and gain kinetic data from during the static stance conditions of the three experiments. Kinetic data from the force plates was sampled at 1000 Hz and synchronously recorded by the motion capture system, which also collected kinematics.

3.3.2 Motion Capture System

Subjects were fitted with 27 reflective markers utilized by a common passive motion capture system (Vicon Nexus, Oxford, UK) for kinematic data collection. The reflective markers were used to generate a model of each subject composed from a modified full body “Plug-in-Gait” model, which encompassed the lower body, torso, right arm, and head. Twelve cameras mounted around the testing area (the Bertec treadmill platform) were used to capture the position of the reflective markers in three-dimensional space, and thus collect the subjects’ kinematic data, which was sampled at 100 Hz.

3.3.3 Tendon Vibrators

Two cylindrical shaped portable vibrators (VB115; Techno-Concept, Cereste, France) were utilized to provide bilateral Achilles tendon vibration during certain conditions in experiments #1 and #2. The postural response to vibration of musculature controlling the ventral side of the ankle, such as the tibialis anterior, has been shown to reproduce the same type of body sway associated

with Achilles tendon vibration, just in the opposite direction (Michel-Pellegrino et al. 2006; Caudron et al. 2010a; Temple et al. 2014; Temple et al. 2016).

Accordingly, we had no reason to believe that interactions between tibialis anterior vibration and our other somatosensory stimuli conditions would produce any different postural responses than those produced with Achilles tendon vibration, other than the direction of sway. Therefore, we decided to only utilize these vibrators on the Achilles tendons in experiments #1 and #2. The cylindrical vibrators were utilized to provide biceps and triceps tendon vibration to the upper right arm during specified conditions in experiment #3. When activated, they produced a continuous 1mm amplitude vibration stimulus at 80 Hz, which has been shown to preferentially activate muscle spindle type Ia primary afferent fibers (Michel-Pellegrino et al., 2006; Harm Slijper & Latash, 2004; Temple et al., 2016; Thompson et al., 2007, 2011).

3.3.4 Vibrating Insoles

Six small round, mechanical vibration tactor stimulators (C-2, Engineering Acoustics Inc., Casselberry, FL, USA) were imbedded into custom-made corkboard insoles for experiment #1. It has previously been described that cork material provides an ideal hardness for creating insoles in which to embed tactors into for purposes of evoking SR via the feet (Hijmans, Geertzen, Schokker, & Postema, 2007). Three tactors were imbedded into each foot insole: one at the heel, one at the first metatarsal-phalangeal joint (MTP1), and one at the fifth metatarsal-phalangeal joint (MTP5). A representation of the location of the tactors in the insoles can be seen below in Figure 3.3.4.

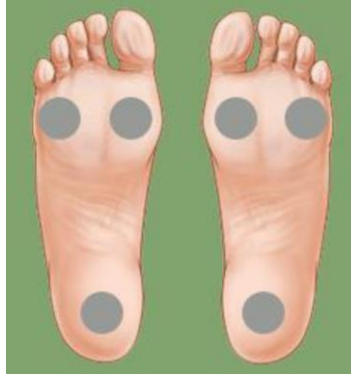


Figure 3.3.4 depicts the location of the six, round mechanical tactors placed in the cork insoles (Dettmer, 2014).

The tactors were capable of producing a maximum displacement amplitude of 0.635 mm (Hijmans et al., 2007). Custom designed software was used to generate a pseudo-random white noise vibration signal at an amplitude that was roughly 50% of each subject's perceptual threshold level. A frequency modulated white noise signal comparable to the features of the noise-generation module that was part of the firmware provided with the tactor controller, was generated by the custom software. The custom software was created to allow users to manipulate stimulus magnitude as a percentage of the maximum vibration output. Thus, whatever percentage of maximal output was determined to be the perceptual threshold level of a subject during threshold testing, half of that percentage was utilized in attempts to evoke SR for conditions when the tactors were turned on. Adding noise at amplitudes of roughly half the perceptual threshold are believed to have induced SR in previous studies and subsequently improved postural control (Goel et al., 2015; Kimura et al., 2012; Temple et al., 2018).

3.3.5 Light Touch Device

The setup for FLT conditions consisted of a board securely fastened to a metal frame that protruded out to a point where subjects touched parallel to the floor. Secured to the board was a plastic, custom-built miniature force plate containing a six-dimensional force/torque transducer (Nano-25; ATI Industrial Automation, Garner, NC, USA). Fastened to the top of the finger force plate was a single C-2 tactor, with a piece of sock fabric attached on top of the tactor. The sock fabric was placed on the tactor as a means of dampening the tactor vibration at roughly the same amount that the socks worn by subjects dampened the tactors in the foot insoles from experiment #1. Subjects lightly touched the sock fabric with their right index fingertip. The tactor was turned on with the same noise signal used in the vibrating foot insoles, at roughly 50% of the fingertip tactile threshold level during conditions with light touch and mechanical noise. Under conditions of FLT without noise, the tactor was turned off.

The same custom-designed software that controlled the tactors while imbedded in the foot insoles was utilized to control the tactor that subjects placed their fingertip on. Control of the finger force plate was provided by custom-built software designed to produce a real-time graph of the vertical forces (in Newtons) imparted on the load sensor. Test administrators utilized this vertical force data to ensure that subjects did not touch the tactor during FLT conditions with a force greater than one Newton, thus keeping subjects from being able to gain mechanical support from the light touch device (Baldan et al., 2014).

3.4 Procedures

Individuals recruited for the study as subjects were given the modified PAR-Q document prior to arriving for testing to determine their eligibility as a potential subject. If the subject was determined to meet the inclusion criteria from the PAR-Q, a date and time for testing was scheduled.

Calibration of the data collection instruments (i.e. Vicon, Bertec treadmill force plates, and the finger force plate) was conducted prior to the arrival of each subject. Once a potential subject arrived at the CNBR, they began the informed consent process. Potential subjects first read the informed consent document themselves. A test administrator then went through the informed consent document individually with the potential subject, answering any questions that they may have had about participating in the study. Once all of a potential subject's questions and concerns had been addressed, they were asked to sign the informed consent document, agreeing to participate in the study.

After obtaining informed consent, subjects removed their shoes and were given a clean pair of athletic socks to wear throughout the testing. Subjects were then tested for their response to Achilles tendon vibration. Two cylindrical vibrators (VB115, Techno-Concept, Cereste, France) were attached bilaterally to the Achilles tendons of the subject and secured with an elastic strap. Subjects then stood upright with their arms across their chest and close their eyes. The test administrator would then turn the vibrators on for a few seconds at 80 Hz and observe the subject's posture from a sagittal viewpoint. After a few seconds of observation, the vibrators were turned off, and the subject was instructed to open their eyes and relax. If the Achilles vibration produced a noticeable

posterior sway in the subject (this occurred in all subjects), the test administrator then proceeded to collect anthropometrics and fit the subject with the reflective markers utilized for kinematic data collection.

Anthropometric data measured for each subject included: height, weight, shoulder height, leg length, knee width, ankle width, shoulder offset, elbow width, and wrist width. A common physicians' scale, tape measure, and anthropometer were used to collect these metrics. Subjects were also fitted for the correct insole size to be used during experiment #1.

Subjects were fitted with a safety harness and a total of 27 reflective markers (Vicon Nexus, Oxford, UK) used to generate a full-body model of each subject. The model was composed from markers on the feet, heels, ankles, tibias, knees, thighs, anterior superior iliac spines, sacrum, T10 vertebra, C7 vertebra, sternum, clavicle, right shoulder, right elbow, right wrist (two markers), and head (four markers). The markers were attached to skin and clothing with double-sided medical tape, and a headband with four markers attached was placed around the head. Once a subject was fitted with all the markers, the subject stepped onto the treadmill with each foot standing over a force plate. The safety harness was secured by an overhanging strap that was loose enough to not provide any mechanical support to the subject when standing upright, yet it was still able to catch the subject if a fall were to occur.

Once harnessed, subjects would then step onto the vibrating insoles, ensuring that the MTP1, MTP5, and the heel were directly placed on top of their corresponding factor. Their stance was positioned with the feet at should width,

and tape was applied to the ground to mark the position of the feet. The stance width remained constant throughout all trials in all the experiments, as there is evidence changes in stance width between trials could impact utilization of sensory feedback in postural control (Adam D. Goodworth, Mellodge, & Peterka, 2014). After setting the correct stance position, vibration perceptual threshold testing for the feet began.

3.4.1 Foot Perceptual Threshold Testing

Perceptual threshold of the noisy foot vibration was tested using a type of two alternative forced-choice, method of levels procedure commonly used when estimating vibration and thermal perceptual thresholds (Aaserud, Juntunen, & Matikainen, 1990; Arezzo, Schaumburg, & Laudadio, 1986; Bartlett, Stewart, Tamblyn, & Abrahamowicz, 1998; Chong & Cros, 2004; Perry, 2006; Shy et al., 2003). The amplitude of the noise signal was raised or lowered during threshold testing as a percentage of the maximal tactor output. During perceptual threshold testing of the foot tactor vibration, subjects stood on the vibrating insoles in the same manner they were going to during the conditions of experiment #1. The subjects stood as opposed to sitting, as perceptual thresholds to the tactors could change under sitting conditions (Mildren, Strzalkowski, & Bent, 2016).

To begin the threshold testing, subjects were given two separate sequential stimulus presentations lasting about two seconds each. One presentation contained a large noisy vibration stimulus that the subject should have felt, and the other contained no vibration stimulation. The subject was

asked to indicate which presentation contained the actual stimulus (“one or two”). If the subject identified the stimulus correctly, the two-choice procedure was then repeated with the next stimulus profile occurring at half the amplitude of the prior. As the subject continued to correctly identify the stimulus, amplitudes of the stimulus presentation continued to decrease by half until an incorrect response was given or the subject admitted to guessing. When that occurred, the amplitude of the next stimulus profile was given halfway between the amplitude of the incorrect response and the last correct response (lowest correct amplitude). These procedures continued until the lowest percentage of maximum tactor output was found where the subject could consistently identify when the vibration was present. Once this vibration perceptual threshold was determined, the amplitude of the mechanical noise signal to be given by the tactors during certain conditions in experiment #1 was set at half the vibration perceptual threshold amplitude. If a subject could detect the tactor stimulus at the lowest output possible and a true perceptual threshold could not be determined, then the lowest noise output the tactors could produce was utilized.

3.4.2 Experiment #1

Aim 1: To determine if an interaction effect exists between small amounts of noisy vibration delivered to the bottom of the feet in the presence of Achilles tendon vibration.

Upon completing the foot perceptual threshold testing, subjects were then asked to complete a series of two Achilles tendon vibration (Off or On) by two foot tactor noise (Off or On) conditions, yielding a total of four separate trials, with each

lasting 40 seconds in duration. Figure 3.4.2 below depicts the four trial conditions (A-D) that each subject was exposed to in experiment #1.

		Tendon Vibration	
		Off	On
Tactor Noise	Off	(A) Foot Tactors Off, Achilles Vibration Off	(B) Foot Tactors Off, Achilles Vibration On
	On	(C) Foot Tactors On, Achilles Vibration Off	(D) Foot Tactors On, Achilles Vibration On

Figure 3.4.2 Table depicting the four somatosensory stimulation conditions (A-D) that each subject was tested on during experiment #1. Completion order of these four conditions was randomized for each subject.

Before each trial, subjects were instructed to close their eyes with their arms at their sides and remain as upright and still as possible for the duration of the trial. Once the subject complied, the test administrator began the trial. During each trial, the first five seconds of the trial consisted of quiet stance without any additional somatosensory stimuli being applied (i.e. Achilles tendon vibration and foot tactors turned off). This five second duration served as a baseline measure for which the data from each trial was normalized to. At the five second mark of each trial, all corresponding somatosensory stimuli conditions (Achilles tendon vibration, or foot tactor noise, or both) were turned on simultaneously and remained on until the end of the trial. Once the trial ended,

the somatosensory stimuli was turned off, and the subject was instructed to open their eyes and relax. After completion of each trial, the subject was asked if they felt the Achilles tendon vibrators, the foot tactors, or both stimuli at any point in the trial. This information served to confirm activation of the tendon vibrators when applicable and provided input as to whether or not the noisy tactor stimuli was indeed subthreshold. Upon completion of all four trial conditions, subjects stepped off the corkboard insoles. The test administrator then removed one of the tactors from an insole and mounted it to the top of the finger force plate on the light touch device, for use in fingertip perceptual threshold testing.

3.4.3 Fingertip Perceptual Threshold Testing

During fingertip perceptual threshold testing, subjects raised their right arm straight out forward to a position where the hand was at shoulder height. From this position, they then flex the elbow to roughly a 45-degree angle, which was measured by a goniometer. Subjects pointed with their right index finger, and the fingertip was positioned directly in front of the midline of the body. All conditions performed in experiments #2 and #3 were completed with the right arm in this position.

To test fingertip perceptual threshold of the tactor vibration, a single tactor attached to the finger force plate was positioned directly below the fingertip. Subjects lightly touched the tactor with their fingertip, and the test administrator ensured that the force pressed on the tactor remained less than one Newton by periodically checking the real-time force indicator graph. Subjects performed the same two alternative forced-choice, method of levels procedure completed for

the foot perceptual threshold testing for the fingertip, utilizing the same frequency modulated noise signal as before. Once the fingertip vibration perceptual threshold was determined, the amplitude of the mechanical noise signal to be given by the tactor during certain conditions in experiments #2 and #3 was again set at half the threshold amplitude, or the lowest noise setting possible was utilized if a subject could detect the lowest stimuli produced by the tactor.

3.4.4 Experiment #2

Aim 2: To determine if an interaction effect exists between light fingertip touch conditions and Achilles tendon vibration.

Upon completing the fingertip perceptual threshold testing, subjects then completed a series of two Achilles tendon vibration (Off or On) by three fingertip light touch (No FLT, FLT Only, or FLT + Noise) conditions, yielding a total of six trials, with each lasting 40 seconds in duration. Figure 3.4.4 below depicts the six trial conditions (E-J) that each subject was exposed to in experiment #2.

		Fingertip Light Touch (FLT) Conditions		
		No FLT	FLT Only	FLT + Noise
Tendon Vibration	Off	(E) No FLT, Achilles Vibration Off	(F) FLT Only, Achilles Vibration Off	(G) FLT + Noise, Achilles Vibration Off
	On	(H) No FLT, Achilles Vibration On	(I) FLT Only, Achilles Vibration On	(J) FLT + Noise, Achilles Vibration On

Figure 3.4.4 Table depicting the six somatosensory stimulation conditions (E-J) that each subject was tested on during experiment #2. Completion order of these six conditions was randomized for each subject.

Like experiment #1, all six conditions of experiment #2 were completed with the subjects' eyes closed. Before each trial, subjects were instructed to remain as upright and still as possible for the duration of the trial, with their left arm at their side and their right arm held out in front of them or lightly touching the tactor, depending on the condition. Once the subject complied, the test administrator began the trial. The first five seconds of each trial again served as a baseline metric from which to normalize each condition to, and the corresponding somatosensory stimuli for each trial was turned on at the five second mark. Conditions with light touch (FLT Only and FLT + Noise) were performed with the subjects lightly touching the tactor for the entire 40-second duration of the trial; however, during the FLT + Noise conditions, the tactor noise did not begin until five seconds into the trial. The two trials performed without FLT were performed with the right arm in the same outstretched position parallel to the ground, as in the FLT conditions. However, without FLT the light touch device was removed before those trials began, and the fingertip did not have contact with a tactor or surface. At the end of each 40-second trial, the somatosensory stimuli were turned off, and the subject was instructed to open their eyes and relax. Subjects were allowed to rest their right arm at their side until the beginning of the next trial. As in experiment #1, after each trial subjects were asked if they felt the Achilles tendon vibrators, the fingertip tactor, or both stimuli at any point in the trial. Upon completion of all six trial conditions for experiment #2, subjects relaxed while the cylindrical Achilles tendon vibrators

were removed. The test administrator then strapped the cylindrical tendon vibrators to the biceps and triceps tendons of the right upper arm in preparation for experiment #3.

3.4.5 Experiment #3

Aim 3: To determine if an interaction effect exists between light fingertip touch and arm tendon vibration conditions.

In experiment #3, the tendon vibrators were attached to the upper right arm of the subject with elastic straps, at a position roughly 4 cm above the elbow joint. This previously utilized positioning (Rabin et al., 2008), ensured that no external reference cues were provided by the vibrator coming into contact with the forearm. In experiment #3, subjects completed a series of three arm tendon vibration (None, Biceps, or Triceps) by three fingertip light touch (No FLT, FLT Only, or FLT + Noise) conditions, yielding a total of nine 40-second trials. Figure 3.4.5 below depicts the nine trial conditions (K-S) that each subject was exposed to in experiment #3.

		Fingertip Light Touch (FLT) Conditions		
		No FLT	FLT Only	FLT + Noise
Tendon Vibration	None	(K) No FLT, No Arm Vibration	(N) FLT Only, No Arm Vibration	(Q) FLT + Noise, No Arm Vibration
	Biceps	(L) No FLT, Biceps Vibration	(O) FLT Only, Biceps Vibration	(R) FLT + Noise, Biceps Vibration
	Triceps	(M) No FLT, Triceps Vibration	(P) FLT Only, Triceps Vibration	(S) FLT + Noise, Triceps Vibration

Figure 3.4.5 Table depicting the nine somatosensory stimulation conditions (K-S) that each subject was tested on during Experiment #3. Completion order of these nine conditions was randomized for each subject.

Before each trial, subjects were again instructed to close their eyes and remain as upright and still as possible for the duration of the trial, with their left arm at their side and their right arm held out in front of them, either pointing or lightly touching the tactor (depending on the condition). The test administrator then began the trial and turned the corresponding somatosensory stimuli conditions on at the five second mark of each trial, again using the first five seconds of data to serve as a baseline from which to normalize each condition to. As in experiment #2, conditions with light touch (FLT Only and FLT + Noise) were performed with the subjects lightly touching the tactor for the entire 40-

second duration of the trial, with the additional somatosensory stimuli (arm vibration, tactor noise, or both) beginning five seconds into the trial when applicable. At the end of each 40-second trial, the somatosensory stimuli were turned off, and the subject opened their eyes and rested their right arm at their side until the beginning of the next trial. At the end of each trial, subjects were again asked if they felt the arm tendon vibrators, the fingertip tactor, or both stimuli at any point in the trial.

Once all nine trial conditions for experiment #3 were done, testing was completed. Tendon vibrators, testing socks, the safety harness, and the reflective markers were all removed from the subject, and the test session was complete.

3.5 Data Reduction and Outcome Measures

3.5.1 Kinetic Data Processing

The kinetic data from the treadmill force plates was processed utilizing MATLAB (Mathworks, Natick, USA). The CoP for the A/P direction was calculated and filtered in MATLAB using a second-order Butterworth filter with a 10-Hz cutoff frequency, as previously utilized (Temple et al., 2014, 2016). The cutoff frequency was determined by the fact that frequency bandwidths of body sway are below 10 Hz during quiet stance (B.-C. Lee, Martin, Ho, et al., 2013; Sienko et al., 2010; Winter, 1995). A/P CoP data was also down sampled to 100 Hz. Differences between the filtered 1000 Hz and filtered down sampled (100 Hz) A/P CoP trajectories were negligible, as a representative sample in Figure

3.5.1 shows below, thus we chose to utilize the 100 Hz A/P CoP data for analysis in order to match the sampling frequency of the kinematic data.

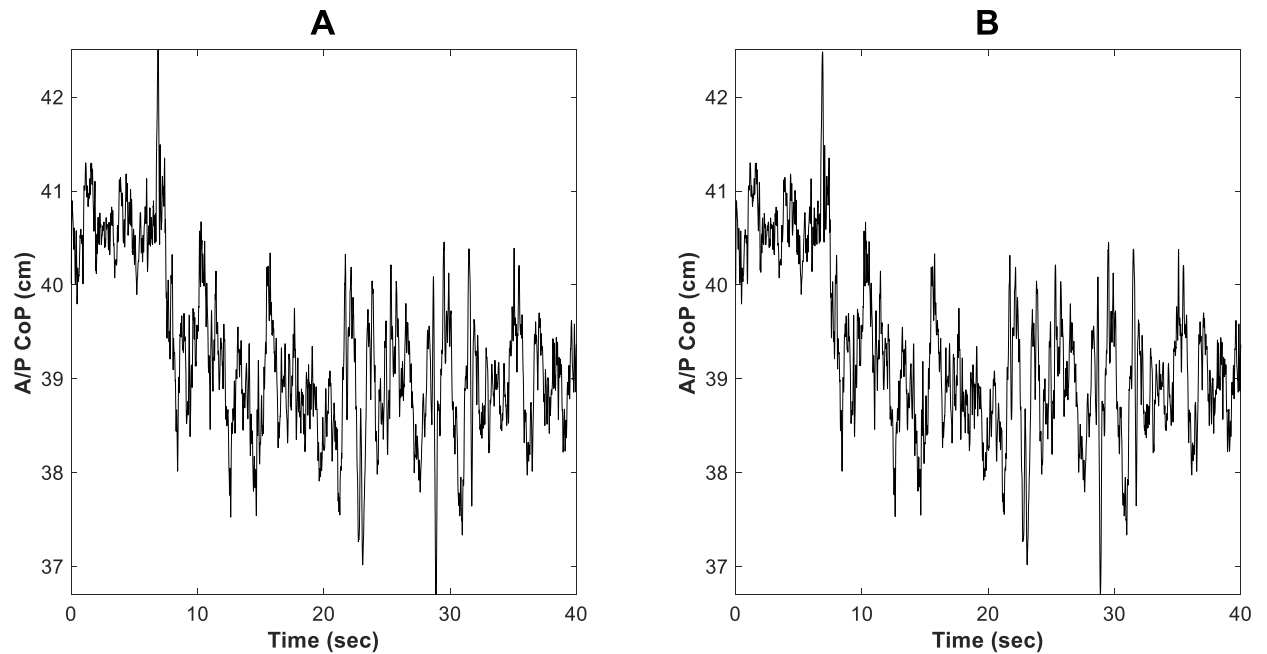


Figure 3.5.1 Depiction of a sample A/P CoP trajectory for one subject at the original filtered 1000 Hz frequency (A) and the down sampled filtered 100 Hz frequency (B).

The filtered and down sampled A/P CoP data was then demeaned to a zero-point based on the mean position of the first five seconds of data for each trial. In other words, the mean of the first five seconds of CoP data for each trial served as the origin. As previously mentioned, these first five seconds of baseline data to reference the remainder of the data to, were completed under conditions with no tactor noise or tendon vibration.

Once the filtered A/P CoP data was referenced to an origin based from the mean of the first five seconds of trial data, the last 30 seconds of trial data was

utilized to calculate the kinetic dependent variables of CoP: POS_{CoP} , SPD_{CoP} , RMS_{CoP} , and $ApEn_{CoP}$. Seconds five through ten of each trial were not utilized in the calculation of dependent variables and subsequent data analysis, as this time period was when the somatosensory stimuli (tendon vibration and/or tactor noise) were turned on. The tendon vibrators take roughly one to two seconds to ramp up to their full amplitude and frequency, making synchronizing of the somatosensory stimuli initiations at full force impossible. Furthermore, the purpose of the current study did not entail observing the initial postural responses to these stimuli, but rather the general effects that combinations of these stimuli had on postural control. Therefore, data from seconds five through ten of each trial were dismissed as the ramp up period when the somatosensory stimuli began and initial postural responses occurred. The time period of interest for analysis was the thirty seconds afterward (seconds 10 – 40), in which the general postural responses to the stimuli at full force could be observed.

Force data from the custom-built miniature force plate for the fingertip was likewise checked in Matlab to confirm subject compliance of not touching during FLT conditions with greater than one Newton of force. The finger force data was filtered using a second-order Butterworth filter with a 10-Hz cutoff frequency. Although a few subjects did seem to exceed one Newton for very brief instances in some trials, none exceeded four Newtons worth of force during any of the trials, which is a level that has previously shown can provide some mechanical support (Baldan et al., 2014). We are therefore confident that none of the subjects in this study received mechanical support from the FLT device.

3.5.2 Kinematic Data Processing

The kinematic data collected by the motion capture system (Vicon Nexus, Oxford, UK) was also processed using MATLAB. Reflective marker positions were filtered using a second-order Butterworth filter with a 10-Hz cutoff frequency, like the kinetic data. Filtered marker positions were then used to calculate angular position trajectories in the A/P direction for the head, torso, thigh, and shank with respect to vertical. The flexion/extension angular trajectories of the right elbow, which were computed in the Vicon “Plug-in-Gait” model, were also filtered with a second-order Butterworth filter using a 10-Hz cutoff frequency and utilized for experiment #3 analysis. These angular trajectories were then demeaned to a zero-point based on the mean position of the first five seconds of data for each trial, like the CoP data, and used to calculate the kinematic variables of POS, SPD, RMS, and AI during the same 30-second data analysis time period as the kinetic data (trial seconds 10 – 40).

3.6 Statistical Analyses

Statistical analyses were conducted using SPSS 20 (IBM Corp, Aramont, NY, USA). All variables (POSSs, SPDs, RMSs, ApEns, and AIs) were checked for normality and homogeneity of variance using the Shapiro-Wilk and Levene’s tests. Several variables appeared to violate these assumptions (normality and homogeneity of variance) with significant Shapiro-Wilk and Levene’s tests ($p < 0.001$), thus nonparametric (Kruskal-Wallis tests) in addition to the traditional parametric (factorial MANOVA) analyses were performed on all the variables from experiment #1 to compare results between the two analyses. All the

dependent variables that showed significant ($p < 0.05$) Achilles vibration by foot noise interaction pairwise comparisons for the parametric factorial MANOVA, likewise indicated to be significantly different ($p < 0.05$) when analyzed with the nonparametric Kruskal-Wallis tests. However, the Kruskal-Wallis tests also indicated there were significant comparisons between some conditions in a few metrics that were not likewise supported by the factorial MANOVA analysis. In essence, the Kruskal-Wallis tests confirmed all the significant findings from the factorial MANOVA; however, the Kruskal-Wallis tests were also less conservative than the factorial MANOVA. One example of a variable that was shown to have significantly different comparisons by the Kruskal-Wallis tests but not the factorial MANOVA is depicted below in Figure 3.6 with a mean data graph.

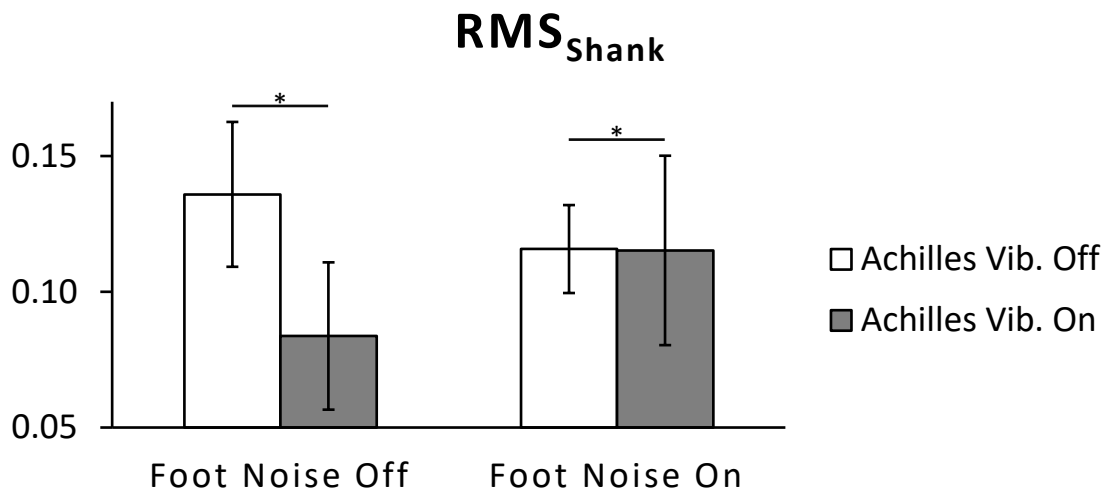


Figure 3.6 Shank demeaned RMS for subjects during the four conditions experienced in experiment #1 ($n = 24$). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration conditions that were indicated by Kruskal-Wallis tests; however, a factorial MANOVA did not yield any significant comparisons for this metric.

Due to the fact that the Kruskal-Wallis tests showed some significant comparisons between conditions where the means were essentially the same (see Figure 3.6 above), we believe the nonparametric analysis was inappropriate and risked the chance of inflating type I error. We therefore decided to report significant findings in this study from the more conservative parametric factorial MANOVA.

IV. RESULTS

4.1 Participants

A total of twenty-five young adults participated in this study (11 males, 14 females, age range 19-34 years, mean height 166.9 ± 8.3 cm, mean weight 66.4 ± 13 kg). All subjects had a body mass index (BMI) less than 30. Although 25 subjects completed all three experiments, one subject was eliminated from analysis for inaccurate kinematic data likely due to electromagnetic interference. Additionally, two more subjects were eliminated from analysis in experiment #2 and one subject was eliminated from analysis in experiment #3 due to missing marker data, which did not allow for accurate kinematic analysis.

4.2 Experiment #1

The MANOVA revealed significant main effects of Achilles vibration [$F(19, 74) = 19.176, p < 0.001$] and foot noise [$F(19, 74) = 1.795, p = 0.039$]. A significant interaction effect of Achilles vibration and foot noise was not found [$F(19, 74) = 0.327, p = 0.996$]; however, Achilles vibration by foot noise pairwise comparisons did yield numerous significant findings in several of the metrics.

4.2.1 Experiment #1 CoP Findings

Mean CoP position (POS_{CoP}) was significantly more posterior with Achilles vibration on than with Achilles vibration off during conditions with both foot noise off [$F(1, 92) = 60.789, p < 0.001$] and foot noise on [$F(1, 92) = 67.088, p < 0.001$]. These findings indicate that the typical POS_{CoP} response to Achilles vibration in this study remained intact, regardless of foot noise condition. Foot noise however did not significantly impact POS_{CoP} , during either conditions with

Achilles vibration off [$F(1, 92) = 0.012, p = 0.914$], or on [$F(1, 92) = 0.253, p = 0.616$], indicating that foot noise did not modulate POS_{CoP} or the typical POS_{CoP} response observed with Achilles vibration on (see Figure 4.2.1.1 below).

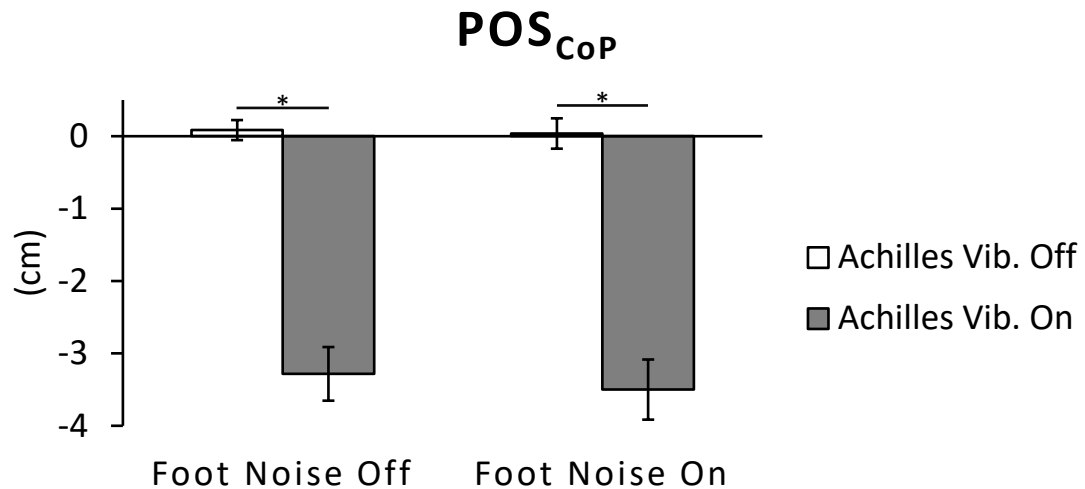


Figure 4.2.1.1 A/P CoP mean position for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in centimeters (cm) with error bars of ± 1 SE. Positive values depict a more anterior position, while negative values depict a more posterior position. Asterisks denote significant comparisons between Achilles vibration conditions.

Mean CoP speed (SPD_{CoP}) was significantly faster with Achilles vibration on than with Achilles vibration off during conditions with foot noise off [$F(1, 92) = 8.31, p = 0.005$], but not with foot noise on [$F(1, 92) = 3.29, p = 0.073$]. These findings indicate that Achilles vibration increased SPD_{CoP}, but when foot noise was also present, the increase in speed from Achilles vibration was less prominent (and no longer statistically significant). SPD_{CoP} was significantly faster

with foot noise on than with foot noise off during conditions with both Achilles vibration off [$F(1, 92) = 10.942, p = 0.001$] and Achilles vibration on [$F(1, 92) = 5.012, p = 0.028$]. These findings indicate that foot noise generally had a greater effect on increasing SPD_{CoP} than Achilles vibration (see Figure 4.2.1.2 below).

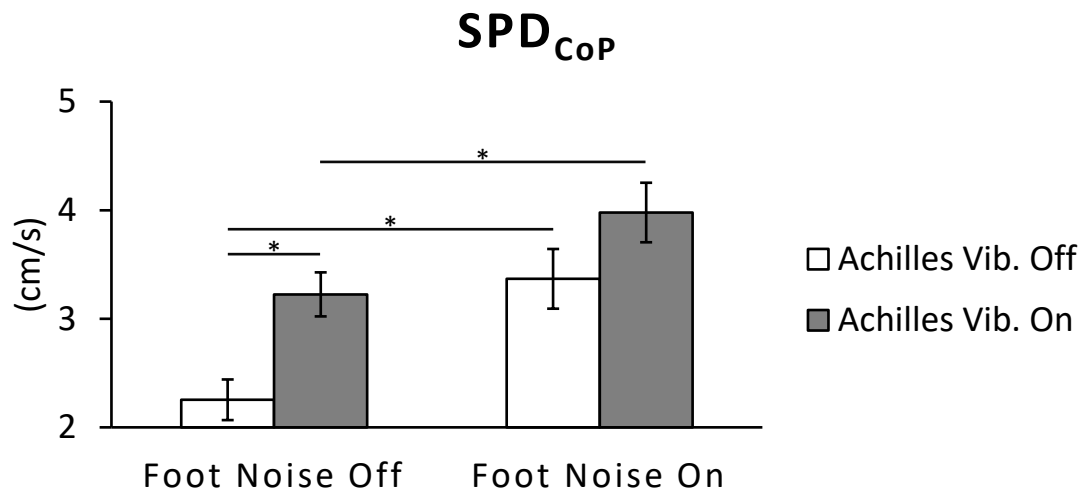


Figure 4.2.1.2 A/P CoP mean speed for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in centimeters per second (cm/s) with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and foot noise conditions.

Demeaned CoP RMS (RMS_{CoP}) was generally less during conditions with Achilles vibration than without vibration; however, it was not significantly less during either of the conditions with [$F(1, 92) = 0.941, p = 0.334$] or without [$F(1, 92) = 0.504, p = 0.480$] foot noise. Likewise, the effect of foot noise (on vs off) did not significantly change RMS_{CoP} during conditions with [$F(1, 92) = 0.059, p = 0.809$] or without [$F(1, 92) < 0.001, p = 0.985$] Achilles vibration. These findings

indicate that both Achilles vibration and foot noise did not have a significant effect on the variability of CoP motion about the mean (see Figure 4.2.1.3 below).

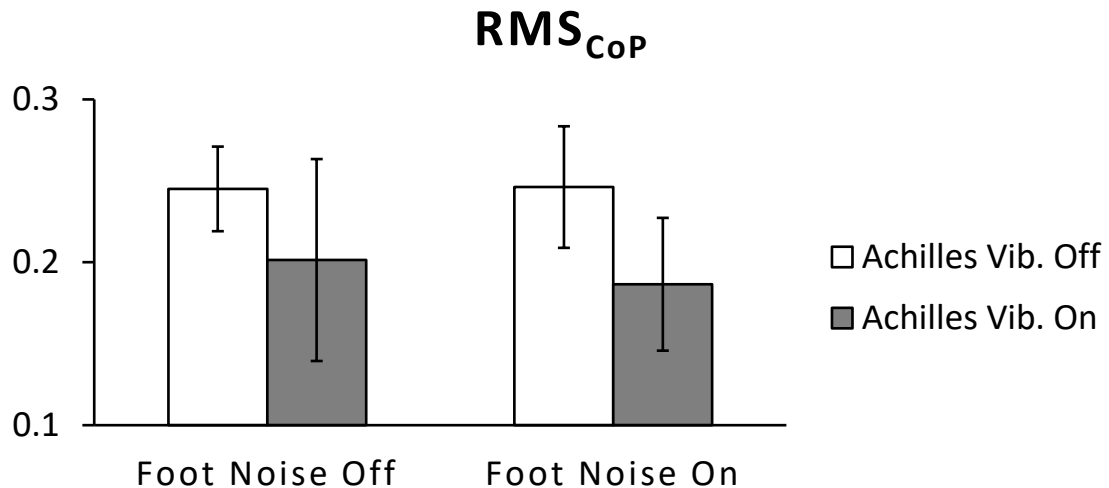


Figure 4.2.1.3 A/P CoP demeaned RMS for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE.

Approximate entropy of the CoP position ($ApEn_{CoP}$) was generally less with Achilles vibration present than without Achilles vibration, but it was significantly less during conditions with both Achilles vibration and foot noise present than during conditions with Achilles vibration off and foot noise on [$F(1, 92) = 7.297, p = 0.008$]. This finding indicates that the presence of foot noise significantly decreased the regularity of CoP motion during conditions when Achilles vibration was also present compared to conditions when Achilles vibration was not present. The other three $ApEn_{CoP}$ pairwise comparisons were not significant (see Figure 4.2.1.4 below).

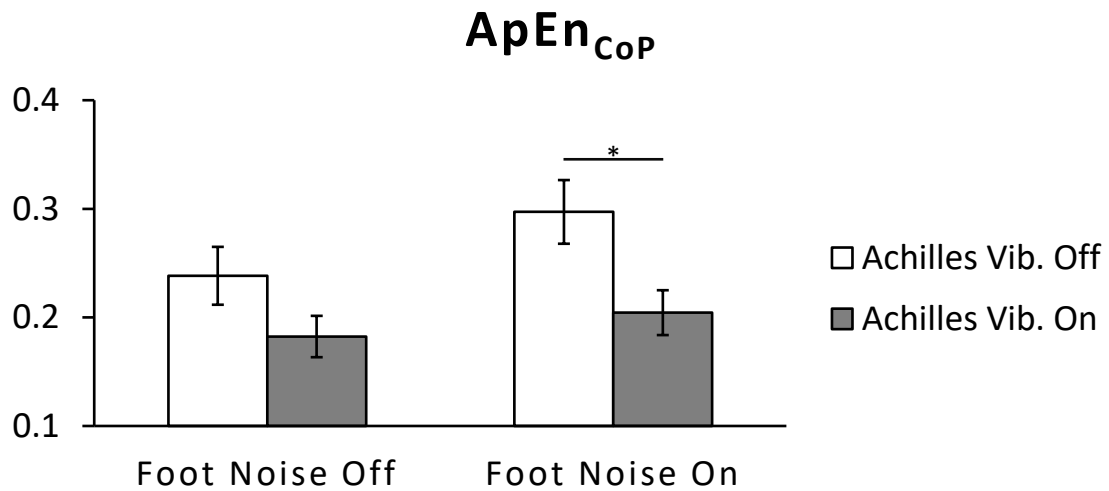


Figure 4.2.1.4 A/P CoP approximate entropy for subjects during the four conditions experienced in experiment #1 (n = 24). Means are represented with error bars of ± 1 SE. Asterisk denotes the significant comparison between Achilles vibration conditions with foot noise on.

4.2.2 Experiment #1 Mean Angular Position Findings

Mean angular position of the head (POS_{Head}) was significantly less anterior with Achilles vibration on than with vibration off, during conditions with foot noise off [$F(1, 92) = 4.85, p = 0.03$], but not during conditions with foot noise on [$F(1, 92) = 3.612, p = 0.06$]. POS_{Head} also tended to be less anterior with foot noise on than with foot noise off; however, neither of the foot noise (on vs. off) comparisons with the Achilles vibration on [$F(1, 92) = 0.174, p = 0.678$] or off [$F(1, 92) = 0.517, p = 0.474$] conditions were significantly different. These findings indicate that both Achilles vibration and foot noise tended to move POS_{Head} less anterior, but this change was only significant as an effect of Achilles vibration (on vs off) when foot noise was not present (see Figure 4.2.2.1 below).

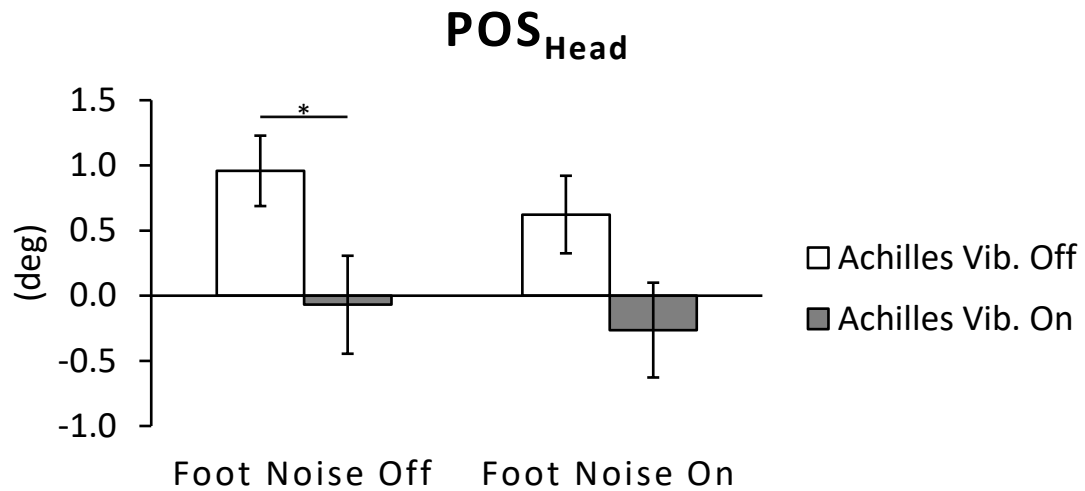


Figure 4.2.2.1 Head mean angular position for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in degrees with error bars of ± 1 SE. Positive values depict head flexion, while negative values depict head extension. Asterisk denotes the significant comparisons between Achilles vibration conditions with foot noise off.

Mean angular positions of the torso (POS_{Torso}), thigh (POS_{Thigh}), and shank (POS_{Shank}) revealed significantly more posterior positions with Achilles vibration on than with Achilles vibration off, regardless of foot noise condition [POS_{Torso} : foot noise off $F(1, 92) = 16.745, p < 0.001$; foot noise on $F(1, 92) = 17.129, p < 0.001$; POS_{Thigh} : foot noise off $F(1, 92) = 49.744, p < 0.001$; foot noise on $F(1, 92) = 55.205, p < 0.001$; POS_{Shank} : foot noise off $F(1, 92) = 33.316, p < 0.001$; foot noise on $F(1, 92) = 35.197, p < 0.001$]. These findings indicate a strong effect of Achilles vibration in pulling body position backward, which was also confirmed by the POS_{CoP} findings. Foot noise did not produce any significant effects on mean angular positions (head, torso, thigh or shank) regardless of Achilles vibration

condition, thus Achilles vibration was the stimulus primarily responsible for postural changes in position (see Figures 4.2.2.2 - 4.2.2.4 below).

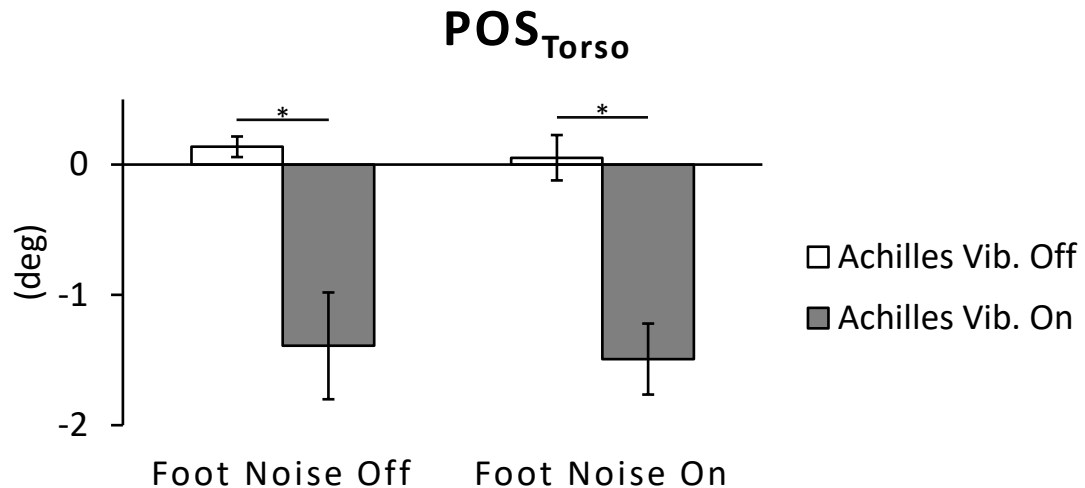


Figure 4.2.2.2 Torso mean angular position for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in degrees with error bars of ± 1 SE. Positive values depict torso flexion, while negative values depict torso extension. Asterisks denote significant comparisons between Achilles vibration conditions.

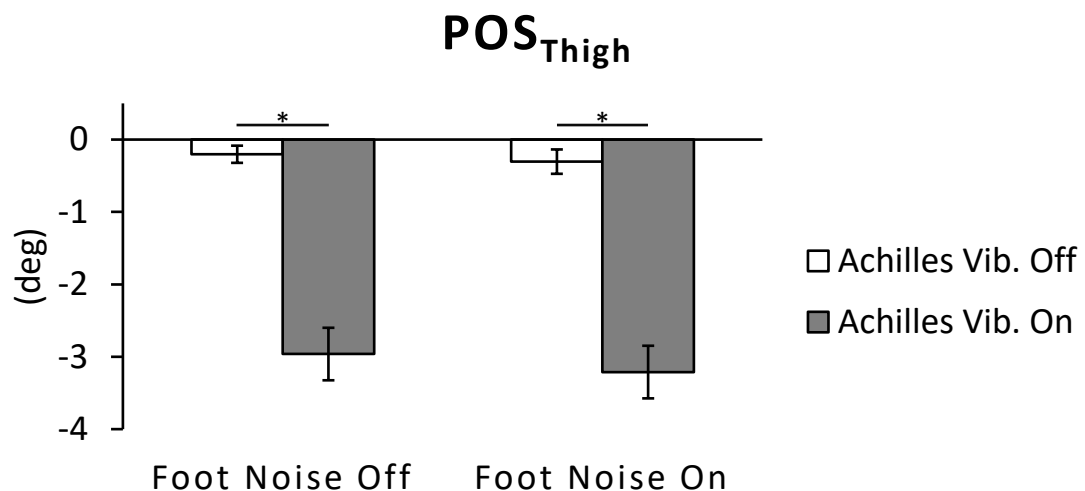


Figure 4.2.2.3 Thigh mean angular position for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in degrees with error bars of ± 1 SE. Negative values depict knee flexion, while positive values depict knee extension. Asterisks denote significant comparisons between Achilles vibration conditions.

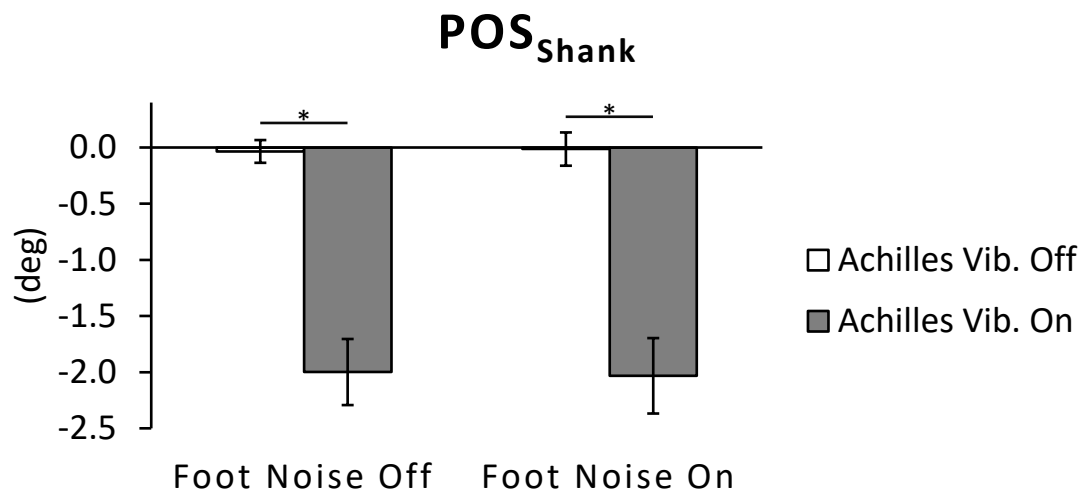


Figure 4.2.2.4 Shank mean angular position for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in degrees with error bars of ± 1 SE. Positive values depict ankle dorsiflexion, while negative values depict ankle plantarflexion. Asterisks denote significant comparisons between Achilles vibration conditions.

4.2.3 Experiment #1 Mean Angular Speed Findings

Mean angular speeds of the head (SPD_{Head}), torso (SPD_{Torso}), thigh (SPD_{Thigh}), and shank (SPD_{Shank}) were significantly faster with Achilles vibration on than with Achilles vibration off, regardless of foot noise condition [SPD_{Head} : foot noise off $F(1, 92) = 7.766$, $p = 0.006$; foot noise on $F(1, 92) = 6.581$, $p = 0.012$; SPD_{Torso} : foot noise off $F(1, 92) = 43.06$, $p < 0.001$; foot noise on $F(1, 92) = 37.632$, $p < 0.001$; SPD_{Thigh} : foot noise off $F(1, 92) = 64.487$, $p < 0.001$; foot noise on $F(1, 92) = 57.429$, $p < 0.001$; SPD_{Shank} : foot noise off $F(1, 92) = 64.591$, $p < 0.001$; foot noise on $F(1, 92) = 48.626$, $p < 0.001$]. These findings indicate a strong effect of Achilles vibration toward increasing body sway speed. Foot noise did not produce any significant effects on body segment mean angular speeds (head, torso, thigh or shank) regardless of Achilles vibration condition, thus Achilles vibration was the stimulus primarily responsible for increases in postural speed (see Figures 4.2.3.1 - 4.2.3.4 below).

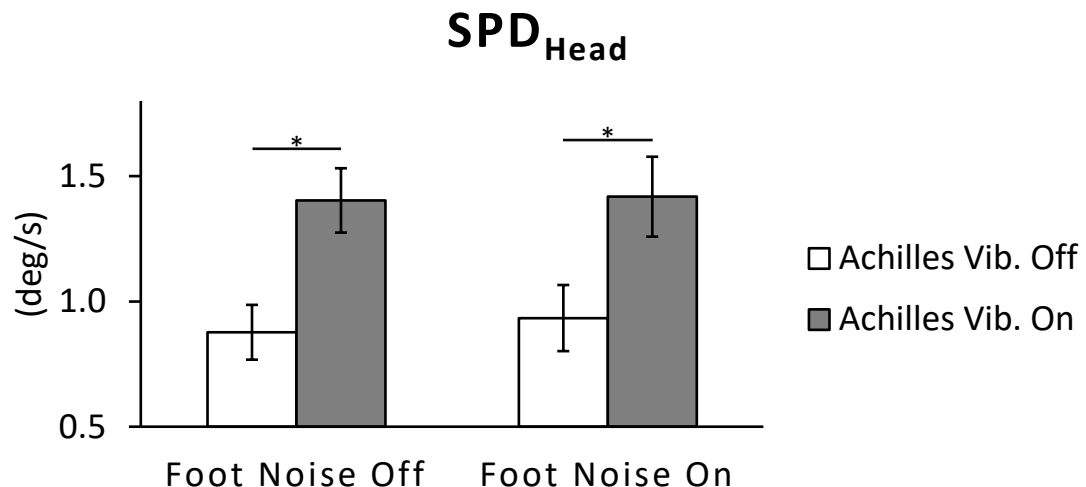


Figure 4.2.3.1 Head mean angular speed for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in degrees

per second (deg/s) with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration conditions.

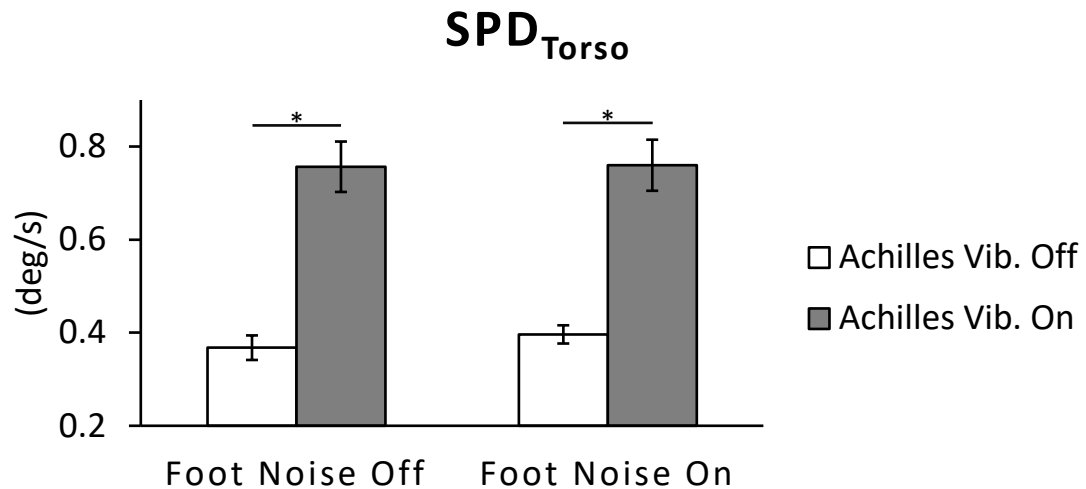


Figure 4.2.3.2 Torso mean angular speed for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration conditions.

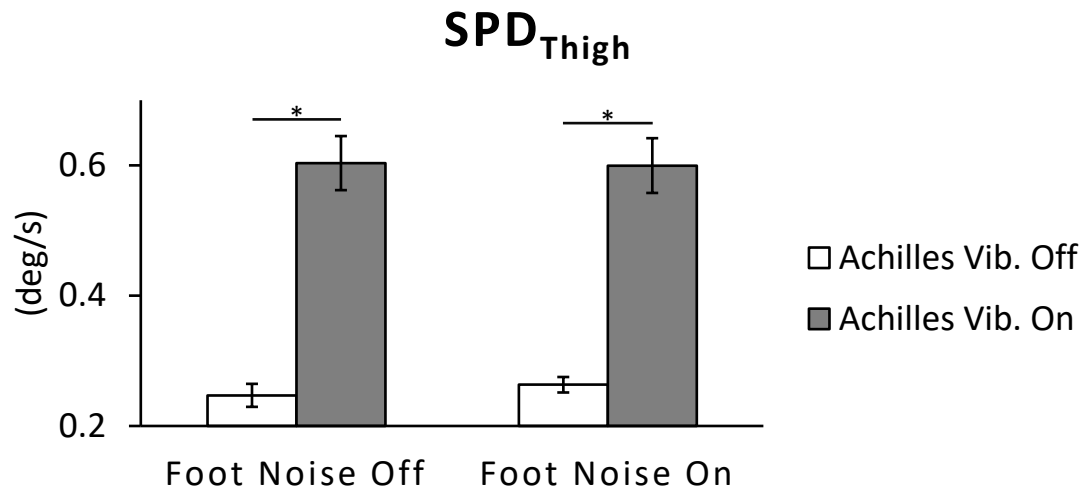


Figure 4.2.3.3 Thigh mean angular speed for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration conditions.

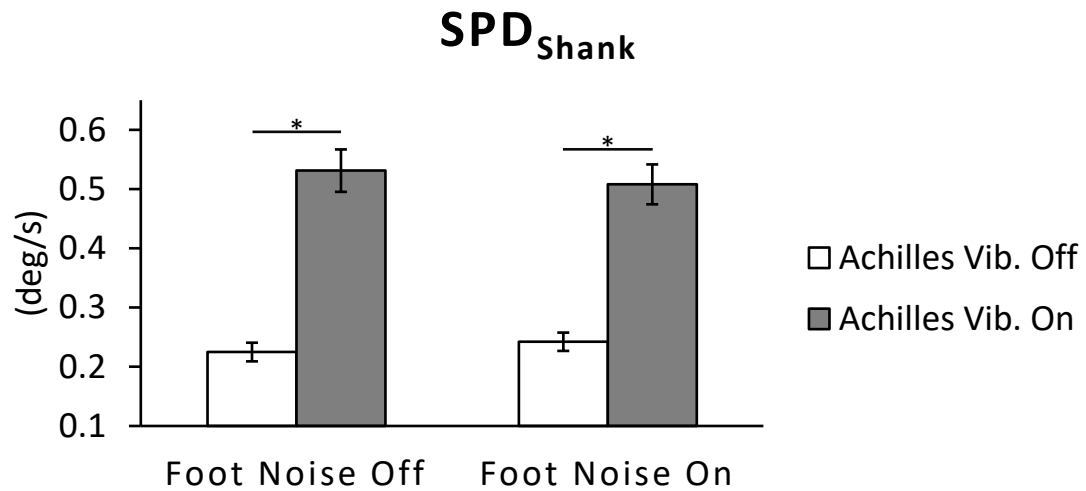


Figure 4.2.3.4 Shank mean angular speed for subjects during the four conditions experienced in experiment #1 (n = 24). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration conditions.

4.2.4 Experiment #1 Angular Demeaned RMS Findings

Achilles vibration by foot noise pairwise comparisons of angular demeaned RMS for the head (RMS_{Head}), torso (RMS_{Torso}), thigh (RMS_{Thigh}), and shank (RMS_{Shank}) were not significantly different for either Achilles vibration comparisons or foot noise comparisons. This indicates that angular variability about the mean angular positions was not significantly impacted by either Achilles vibration or foot noise (see Figures 4.2.4.1 - 4.2.4.4 below).

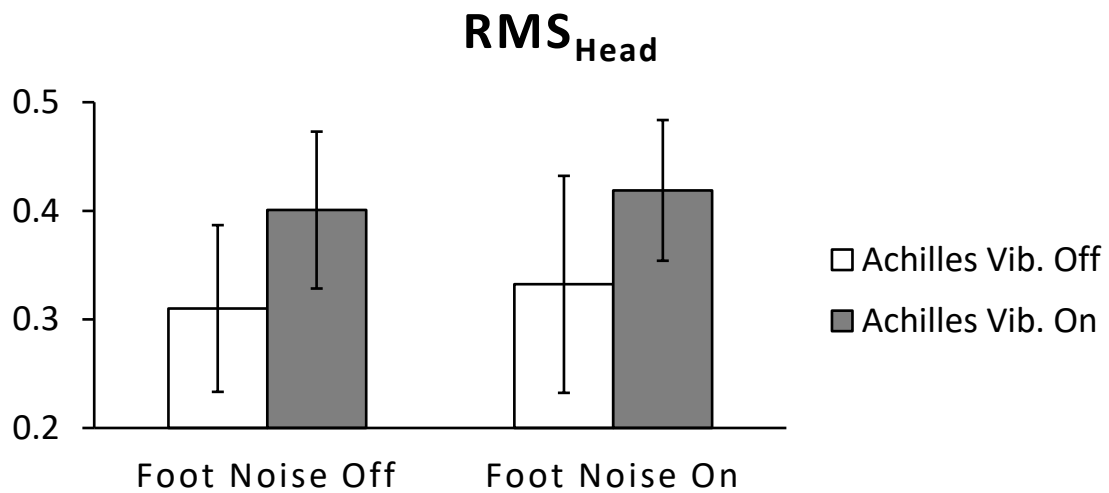


Figure 4.2.4.1 Head demeaned RMS for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE.

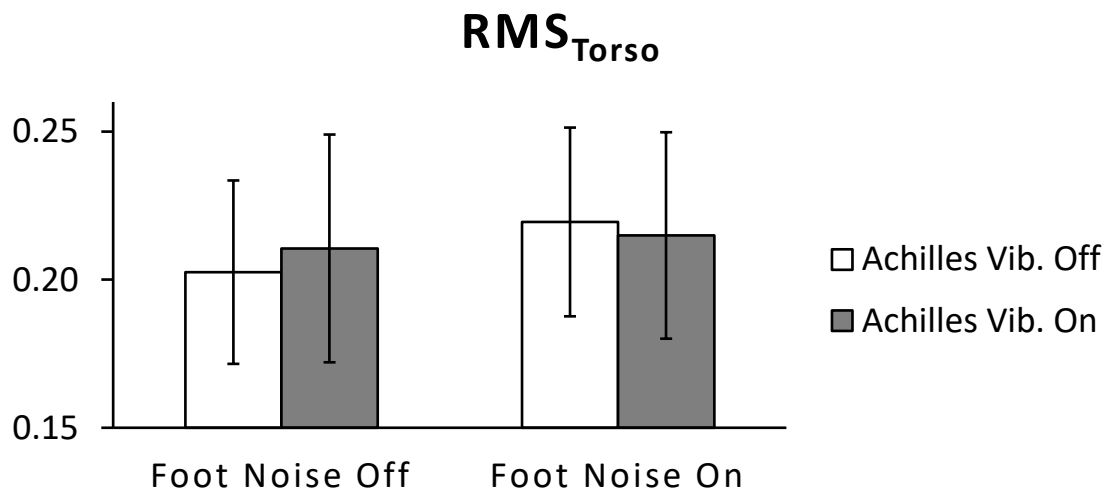


Figure 4.2.4.2 Torso demeaned RMS for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE.

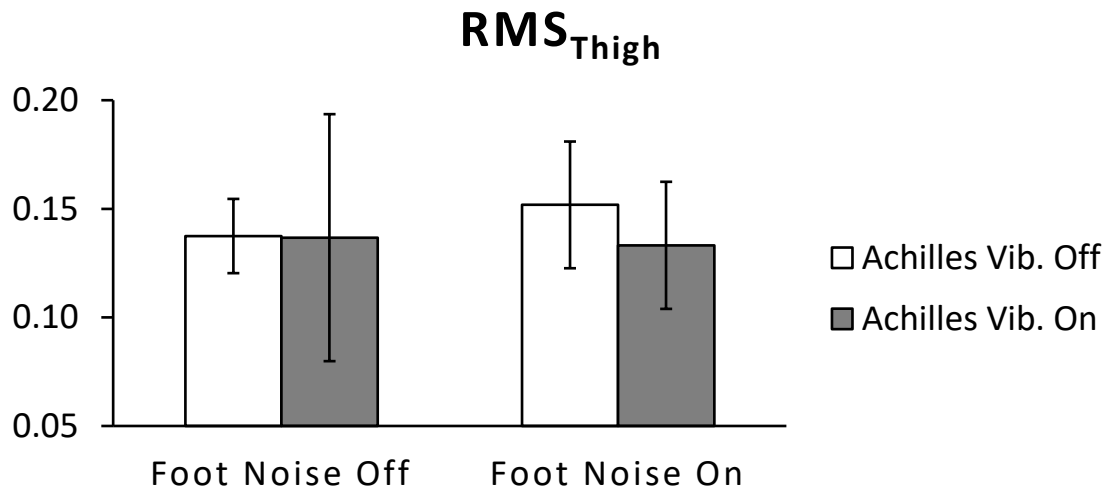


Figure 4.2.4.3 Thigh demeaned RMS for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE.

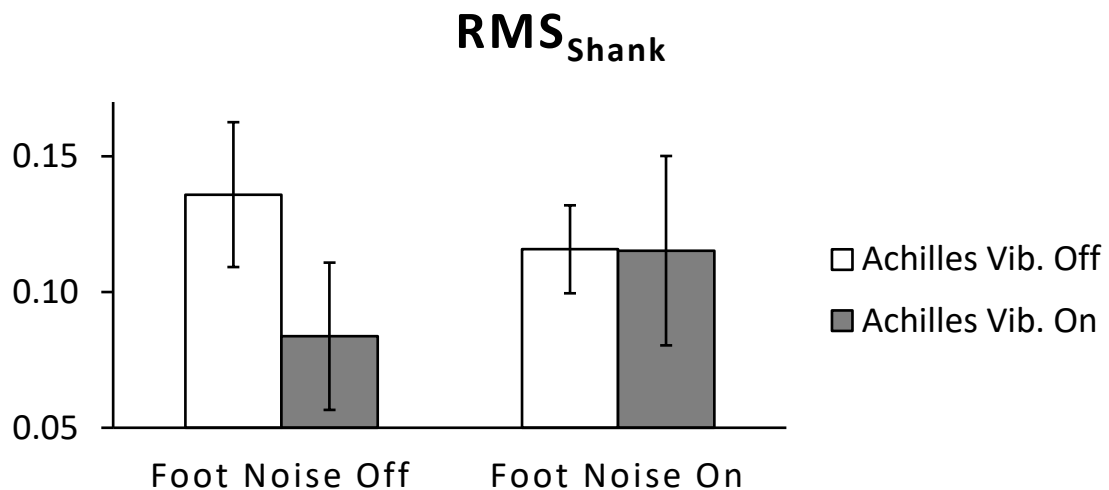


Figure 4.2.4.4 Shank demeaned RMS for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE.

4.2.5 Experiment #1 Anchoring Index Findings

Anchoring Index of the head (AI_{Head}) and torso (AI_{Torso}) were not significantly different for either Achilles vibration comparisons or foot noise comparisons. This indicates that the “anchoring” or angular motion of both the head and the torso with respect to their inferior segment was not significantly impacted by either Achilles vibration or foot noise (see Figures 4.2.5.1 and 4.2.5.2 below).

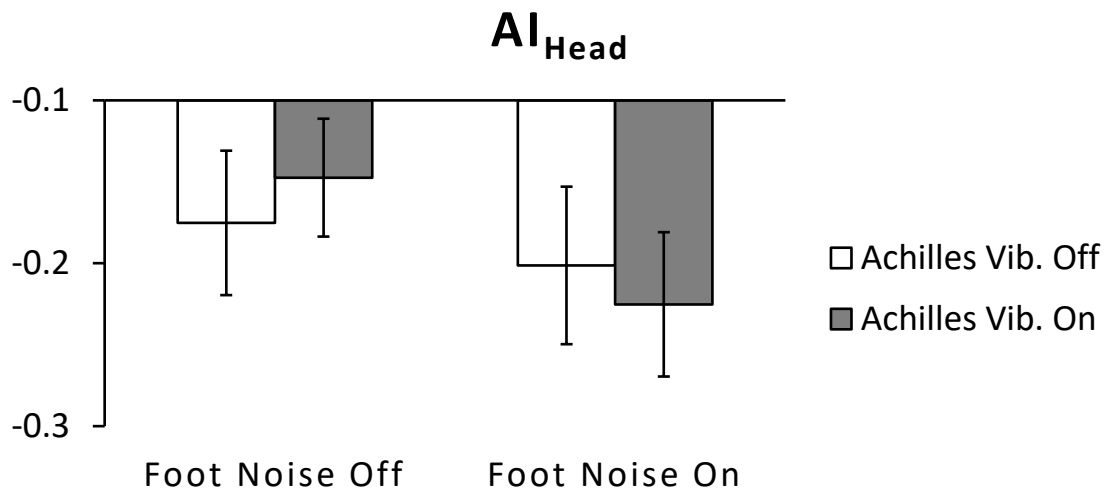


Figure 4.2.5.1 AI_{Head} for subjects during the four conditions experienced in experiment #1 ($n = 24$). Mean values are depicted with error bars of ± 1 SE.

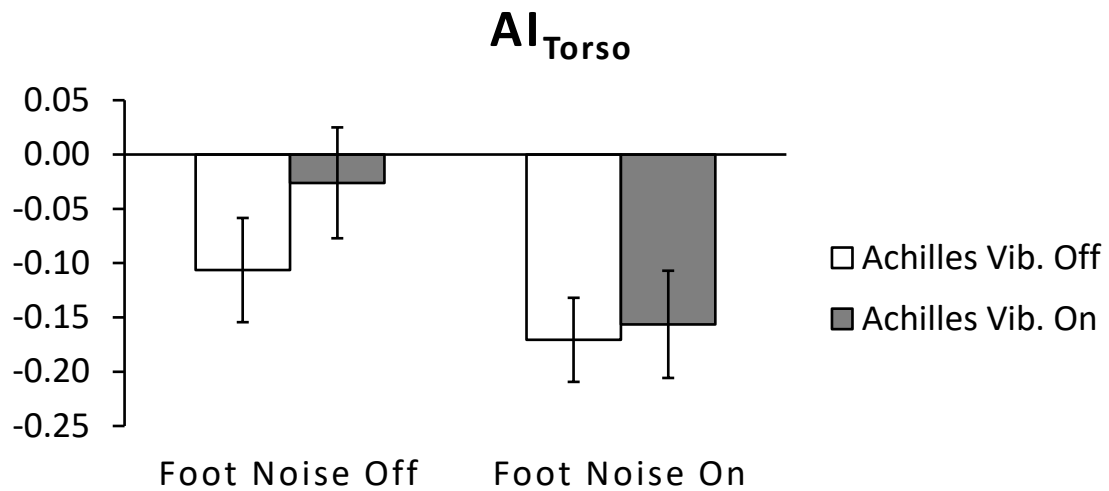


Figure 4.2.5.2 Al_{Torso} for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE.

Anchoring Index of the thigh (Al_{Thigh}) however, was significantly greater with Achilles vibration on than with Achilles vibration off, during conditions when foot noise was off [$F(1, 92) = 13.172, p < 0.001$] as well as when foot noise was on [$F(1, 92) = 7.616, p = 0.007$]. These findings indicate that anterior/posterior angular motion of the thigh with respect to the shank was significantly impacted by Achilles vibration, with anchoring of the thigh to the shank being decreased during stance with the vibration present (see Figure 4.2.5.3 below).

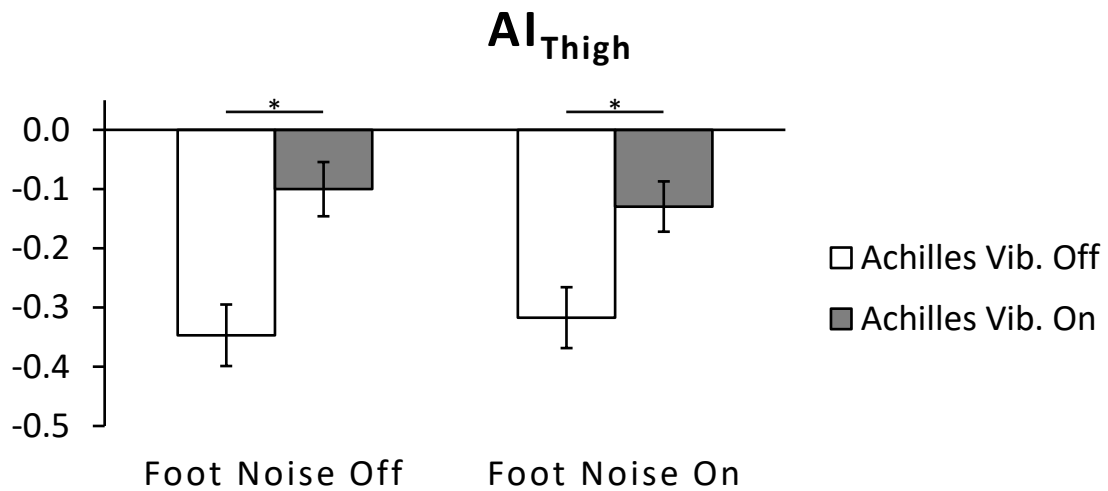


Figure 4.2.5.3 Al_{Thigh} for subjects during the four conditions experienced in experiment #1 (n = 24). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration conditions.

4.3 Experiment #2

In experiment #2, the MANOVA revealed significant main effects of Achilles vibration [$F(19, 108) = 29.963, p < 0.001$] and FLT condition [$F(38, 218) = 3.797, p < 0.001$]. A significant interaction effect of Achilles vibration and FLT condition was also found [$F(38, 218) = 2.46, p < 0.001$], and several Achilles vibration by FLT condition pairwise comparisons did yield significant findings in the metrics.

4.3.1 Experiment #2 CoP Findings

Achilles vibration caused the POS_{CoP} to be significantly more posterior when on compared to when it was off, for all FLT conditions (No FLT: $p < 0.001$; FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$). These findings indicate that much

like in experiment #1, the typical response of Achilles vibration moving CoP posterior remained intact during all FLT conditions. However, FLT did have a significant effect on the Achilles vibration response, as both FLT Only and FLT + Noise were significantly less posterior than No FLT when Achilles vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). No significant changes in POS_{CoP} occurred as a result of FLT condition when no Achilles vibration was present. These findings indicate that although Achilles vibration significantly moved POS_{CoP} more posterior, FLT did modulate that effect, causing POS_{CoP} to be significantly less posterior when both Achilles vibration and FLT (FLT Only and FLT + Noise) conditions were present compared to the Achilles vibration condition with No FLT present. The effect of adding noise to the light touch (FLT + Noise), did not significantly change POS_{CoP} beyond the effect of FLT itself though (see Figure 4.3.1.1).

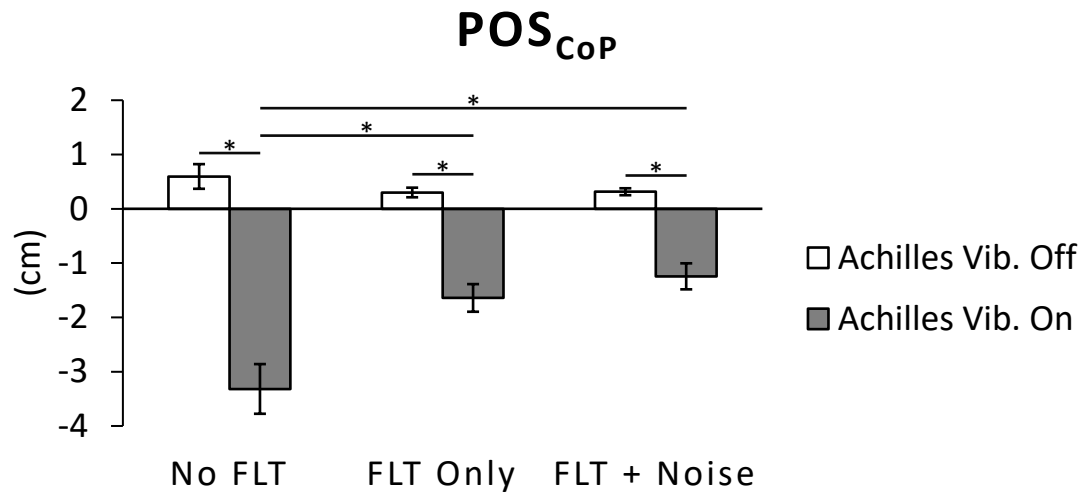


Figure 4.3.1.1 A/P CoP mean position for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in cm with error bars of ± 1 SE. Positive values depict a more anterior position, while negative values depict a more posterior position. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

SPD_{CoP} was significantly faster with Achilles vibration on than with Achilles vibration off with No FLT ($p = 0.002$) present, but it was not during the FLT Only ($p = 0.07$) and FLT + Noise ($p = 0.053$) conditions. FLT conditions were not significantly different during either the Achilles vibration Off or On conditions. These findings indicate that Achilles vibration increased SPD_{CoP}, but when FLT was also present, the increase in speed from Achilles vibration was no longer statistically significant. The effect of adding noise to the light touch, did not significantly change SPD_{CoP} beyond the effect of FLT itself though (see Figure 4.3.1.2 below).

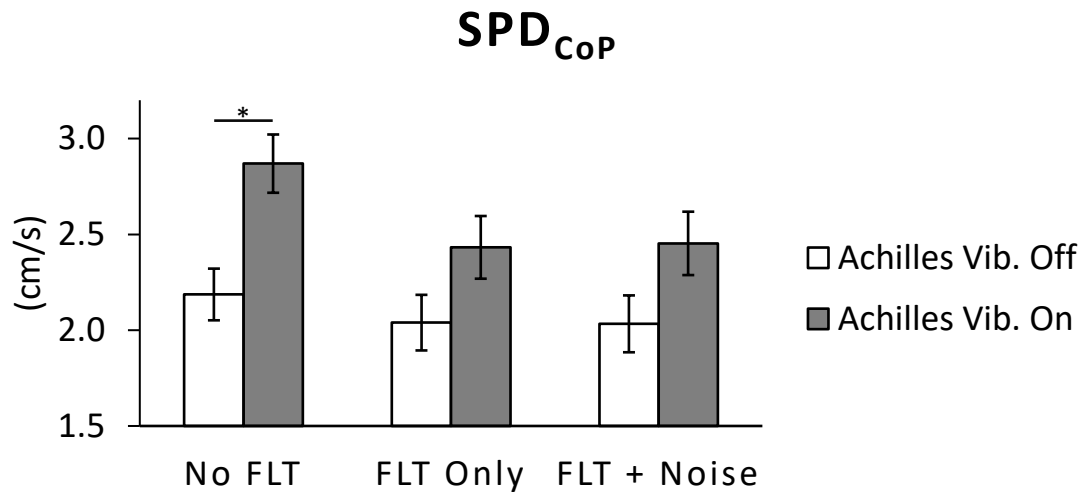


Figure 4.3.1.2 A/P CoP mean speed for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in cm/s with error bars of ± 1 SE. Asterisk denotes significant comparison between Achilles vibration conditions during no FLT.

RMS_{CoP} was not significantly different between Achilles vibration conditions (on vs. off) during any of the FLT conditions; however, it was significantly less during conditions with FLT than with No FLT when no Achilles vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p = 0.011$). These findings indicate that like in experiment #1, Achilles vibration did not have a significant effect on the variability of CoP motion about the mean. When no Achilles vibration was present, variability of CoP motion about the mean was significantly decreased by FLT conditions; however, the addition of noise to the light touch did not significantly affect RMS_{CoP} beyond the effect of FLT itself (see Figure 4.3.1.3 below).

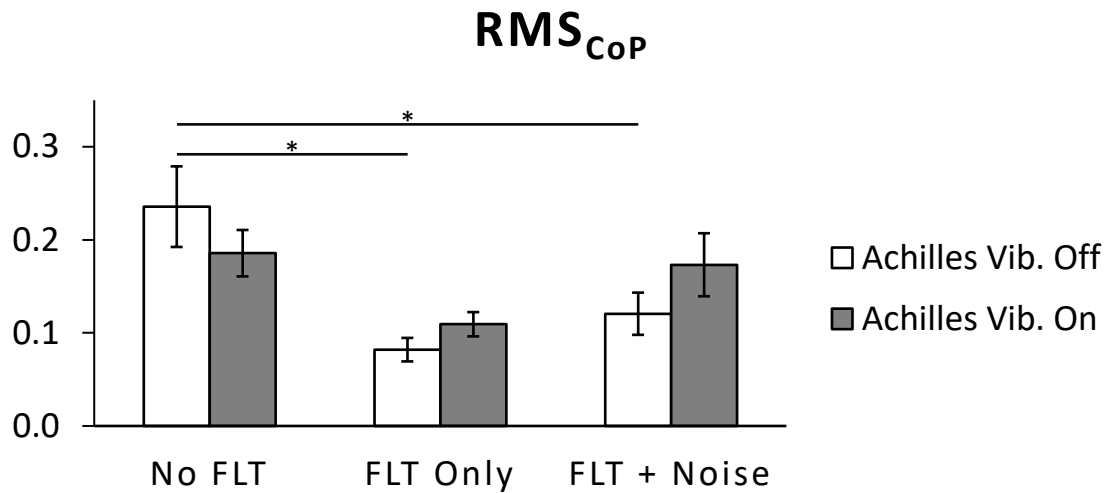


Figure 4.3.1.3 A/P CoP demeaned RMS for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions when Achilles vibration is not present.

ApEn_{CoP} was significantly less with Achilles vibration present than without Achilles vibration during both conditions with FLT present (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$). When no Achilles vibration was present, ApEn_{CoP} was significantly greater during both conditions with FLT than the condition with No FLT present (FLT Only vs No FLT: $p < 0.001$; FLT + Noise vs No FLT: $p < 0.001$). These findings indicate that when no Achilles vibration was present, the regularity of CoP motion was significantly decreased by FLT conditions. Furthermore, both FLT conditions significantly decreased the regularity of CoP motion when no Achilles vibration was present compared to when Achilles vibration was present. However, the addition of noise to the light touch did not

significantly affect regularity of CoP motion beyond the effect of FLT itself (see Figure 4.3.1.4 below).

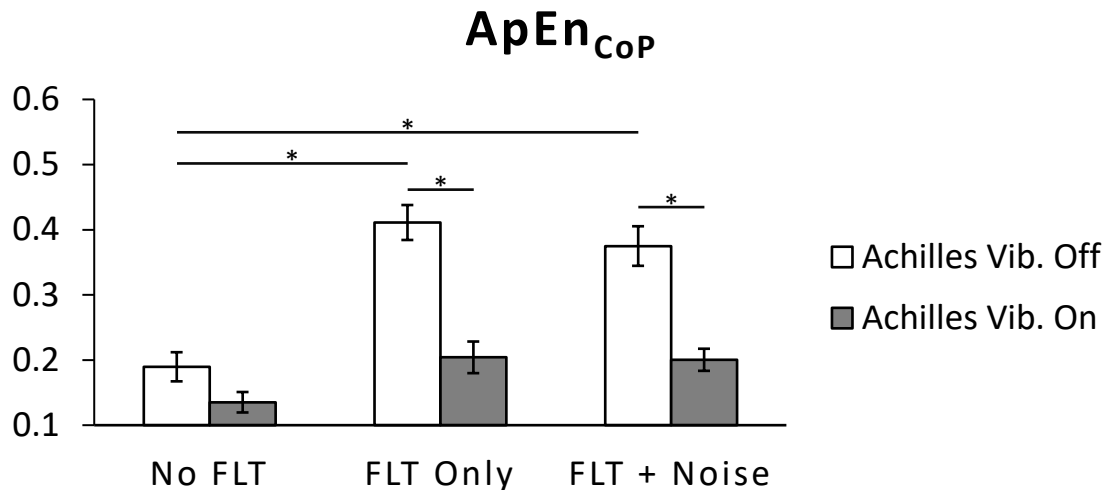


Figure 4.3.1.4 A/P CoP approximate entropy for subjects during the six conditions experienced in experiment #2 (n = 22). Means are represented with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

4.3.2 Experiment #2 Mean Angular Position Findings

POS_{Head} was significantly less anterior with Achilles vibration on than with vibration off, during conditions with No FLT ($p = 0.031$). POS_{Head} had a tendency to shift more anterior during FLT conditions (both FLT Only and FLT + Noise) with Achilles vibration on and tended to move more posterior during FLT conditions with Achilles vibration off; however, none of the interaction pairwise comparisons beyond Achilles vibration (on vs. off) during No FLT were significantly different. These results confirmed experiment #1 findings that Achilles vibration significantly moved POS_{Head} less anterior when No FLT was

present. However, there was not a significant effect of Achilles vibration (on vs. off) on POS_{Head} during both the FLT and FLT + Noise conditions, thus FLT helped stabilize POS_{Head} to reduce the typical Achilles vibration effect when No FLT was present. Adding noise to the light touch did not seem to affect POS_{Head} beyond FLT itself though (see Figure 4.3.2.1 below).

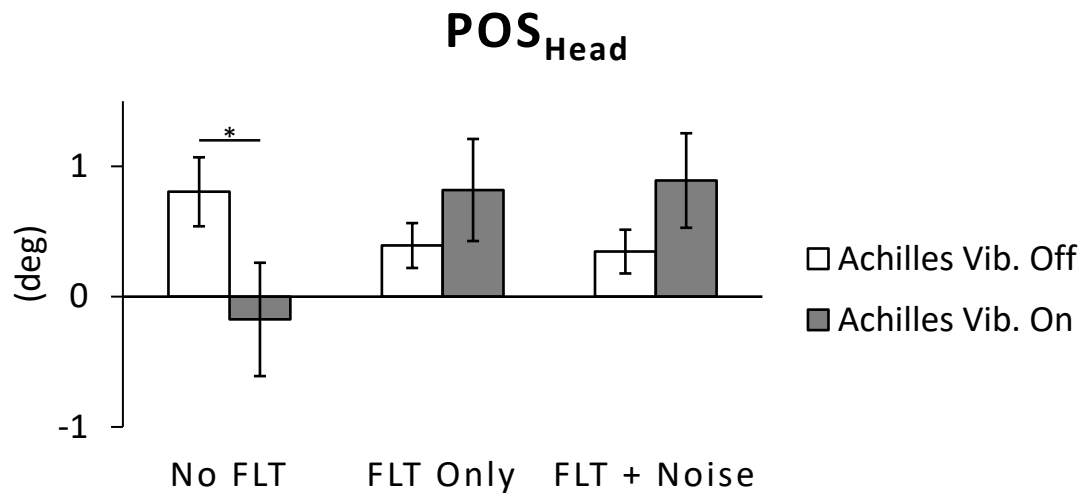


Figure 4.3.2.1 Head mean angular position for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in degrees with error bars of ± 1 SE. Positive values depict head flexion, while negative values depict head extension. Asterisk denotes significant comparison between Achilles vibration conditions during No FLT.

POS_{Torso} revealed significant extension with Achilles vibration on compared to Achilles vibration off during the No FLT condition ($p < 0.001$). During conditions with FLT, POS_{Torso} was significantly more flexed with Achilles vibration on than with Achilles vibration off (FLT Only: $p = 0.032$; FLT + Noise: $p = 0.026$). When Achilles vibration was present, POS_{Torso} was also significantly

more flexed during FLT conditions than during No FLT (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). These findings confirmed the effect from experiment #1 of Achilles vibration pulling POS_{Torso} significantly more posterior. When FLT was present with Achilles vibration on however, POS_{Torso} was shifted significantly more anterior than No FLT with the vibration on, as well as significantly more anterior than Achilles vibration off conditions with FLT. Adding noise to the light touch did not significantly affect POS_{Torso} beyond the effect of FLT itself though (see Figure 4.3.2.2 below).

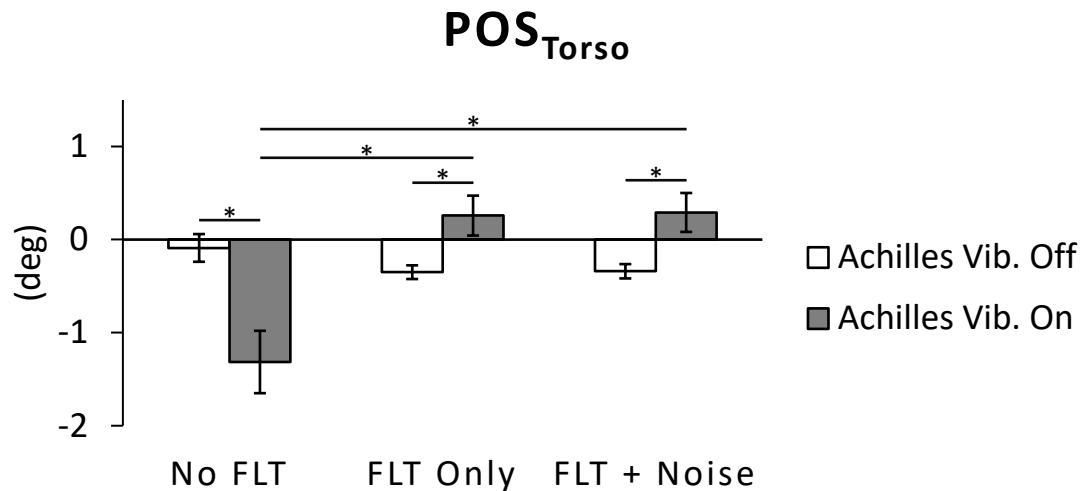


Figure 4.3.2.2 Torso mean angular position for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in degrees with error bars of ± 1 SE. Positive values depict torso flexion, while negative values depict torso extension. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

POS_{Thigh} revealed significant knee flexion with Achilles vibration on compared to Achilles vibration off, regardless of FLT condition (No FLT: $p <$

0.001; FLT Only: $p < 0.001$; FLT + Noise: $p = 0.004$). With Achilles vibration present however, POS_{Thigh} also showed significantly less knee flexion during conditions with FLT than during No FLT (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). These findings confirmed the effect from experiment #1 of Achilles vibration pulling POS_{Thigh} significantly more posterior. When FLT was present with Achilles vibration on however, POS_{Thigh} was significantly less posterior than No FLT with the vibration on. Adding noise to the light touch did not significantly affect POS_{Thigh} beyond the effect of FLT itself though (see Figure 4.3.2.3 below).

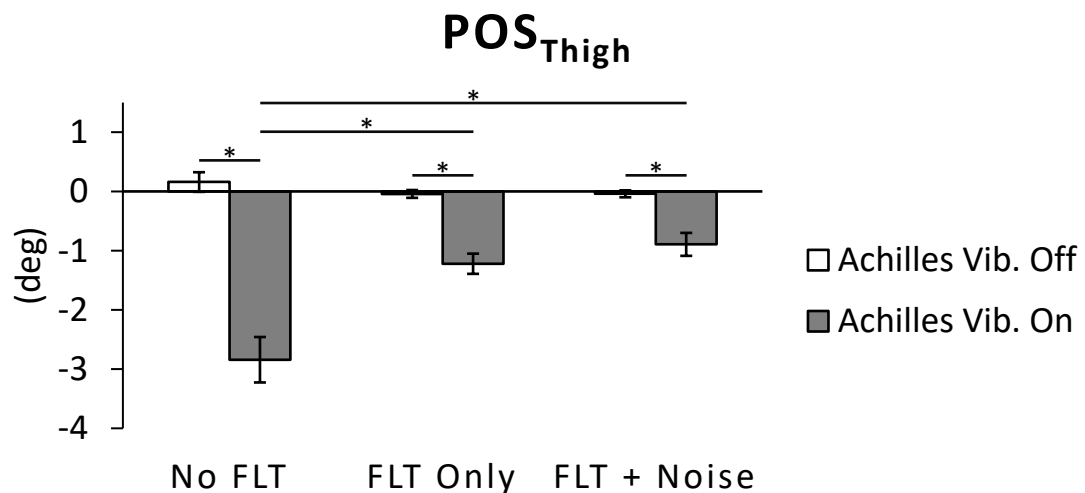


Figure 4.3.2.3 Thigh mean angular position for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in degrees with error bars of ± 1 SE. Negative values depict knee flexion, while positive values depict knee extension. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

POS_{Shank} revealed significant ankle plantarflexion with Achilles vibration on compared to Achilles vibration off, regardless of FLT condition (No FLT: $p < 0.001$; FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$). No other pairwise comparisons for the POS_{Shank} reached the level of significance though. These findings confirmed the effect from experiment #1 of Achilles vibration pulling POS_{Shank} significantly more posterior. FLT however, was unable to significantly reduce this posterior motion, as it did at the levels of the thigh and torso (see Figure 4.3.2.4 below).

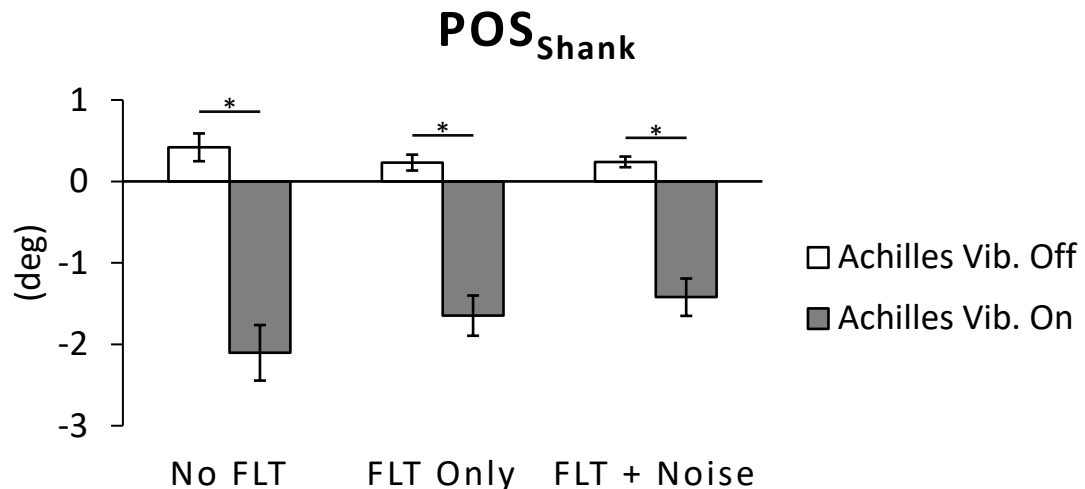


Figure 4.3.2.4 Shank mean angular position for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in degrees with error bars of ± 1 SE. Positive values depict ankle dorsiflexion, while negative values depict ankle plantarflexion. Asterisks denote significant comparisons between Achilles vibration conditions.

4.3.3 Experiment #2 Mean Angular Speed Findings

SPD_{Head} was significantly faster with Achilles vibration on compared to Achilles vibration off, regardless of FLT condition (No FLT: $p < 0.001$; FLT Only: $p = 0.003$; FLT + Noise: $p < 0.001$). With Achilles vibration present however, SPD_{Head} was also significantly slower during conditions with FLT than during No FLT (FLT Only vs. No FLT: $p = 0.005$; FLT + Noise vs. No FLT: $p = 0.042$). These findings confirmed the effect from experiment #1 of Achilles vibration significantly increasing SPD_{Head}. When FLT was present along with the Achilles vibration however, SPD_{Head} was significantly slower than the No FLT condition with vibration on. Adding noise to the light touch did not significantly affect SPD_{Head} beyond the effect of FLT itself though (see Figure 4.3.3.1 below).

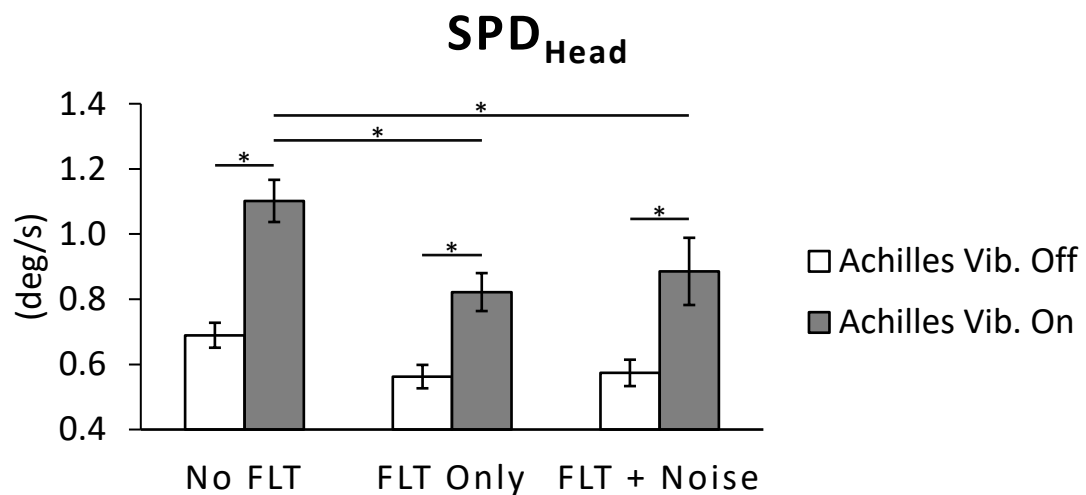


Figure 4.3.3.1 Head mean angular speed for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

The three lower segments also showed significant increases in mean angular speed when Achilles vibration was present (on vs. off), regardless of FLT condition (SPD_{Torso}: No FLT, $p < 0.001$; FLT Only, $p = 0.004$; FLT + Noise, $p = 0.001$; SPD_{Thigh}: No FLT, $p < 0.001$; FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$; SPD_{Shank}: No FLT, $p < 0.001$; FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$). Likewise, the increased speed was again significantly reduced in the three lower segments by the presence of FLT compared to No FLT when Achilles vibration was present (SPD_{Torso}: FLT Only vs. No FLT, $p < 0.001$; FLT + Noise vs. No FLT, $p < 0.001$; SPD_{Thigh}: FLT Only vs. No FLT, $p < 0.001$; FLT + Noise vs. No FLT, $p < 0.001$; SPD_{Shank}: FLT Only vs. No FLT, $p < 0.001$; FLT + Noise vs. No FLT, $p < 0.001$). However, mean angular speed in the three lower segments was also significantly reduced by the presence of FLT compared to No FLT when Achilles vibration was not present (SPD_{Torso}: FLT Only vs. No FLT, $p = 0.028$; FLT + Noise vs. No FLT, $p = 0.045$; SPD_{Thigh}: FLT Only vs. No FLT, $p < 0.001$; FLT + Noise vs. No FLT, $p < 0.001$; SPD_{Shank}: FLT Only vs. No FLT, $p = 0.012$; FLT + Noise vs. No FLT, $p = 0.033$). These results confirmed the findings of experiment #1 that Achilles vibration significantly increased mean angular speeds in the A/P directions for all body segments evaluated. The increases in body segment angular speeds caused by Achilles vibration were significantly reduced by the presence of FLT, and the three lower segments even saw a significant effect of FLT reducing speed when Achilles vibration was not present. Adding noise to the light touch did not significantly affect segment mean angular

speeds beyond the effects of FLT itself though (see Figures 4.3.3.2 – 4.3.3.4 below).

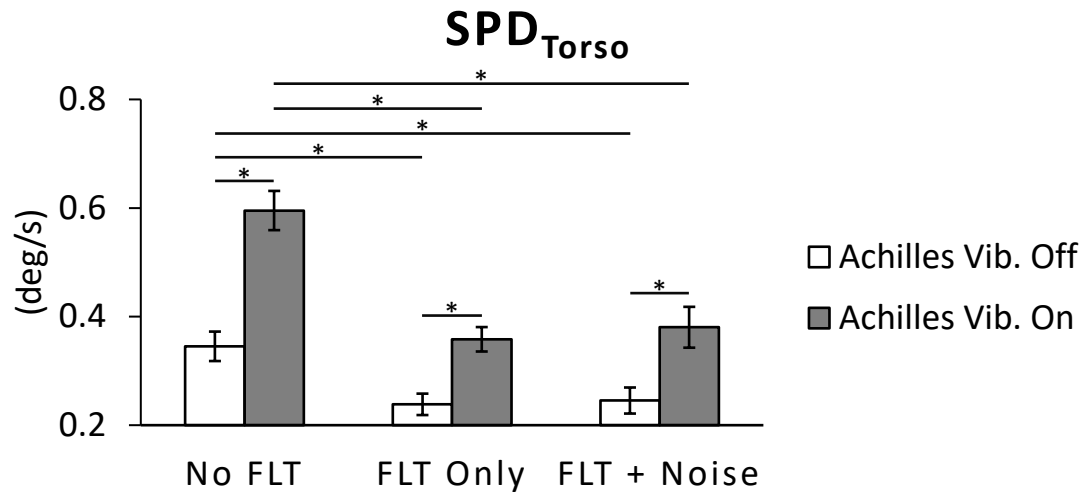


Figure 4.3.3.2 Torso mean angular speed for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

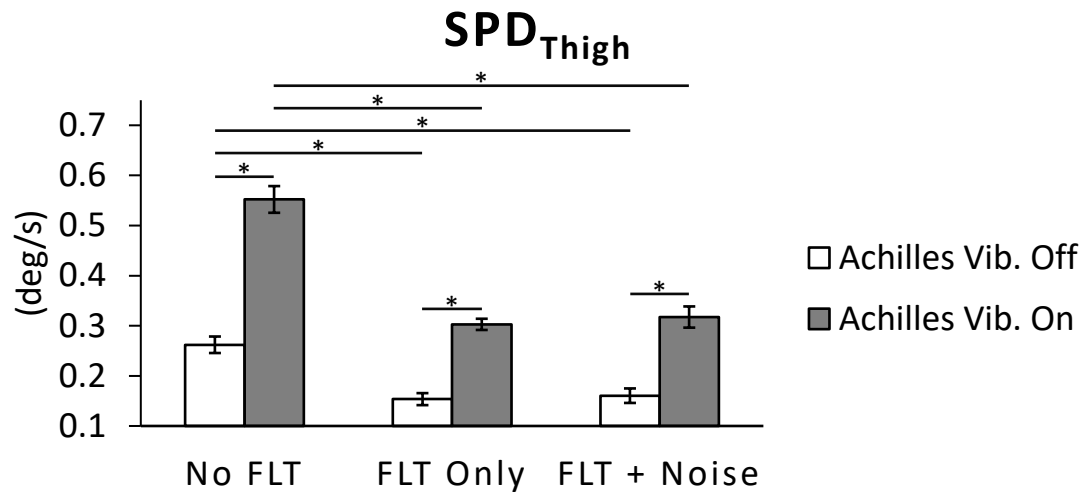


Figure 4.3.3.3 Thigh mean angular speed for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

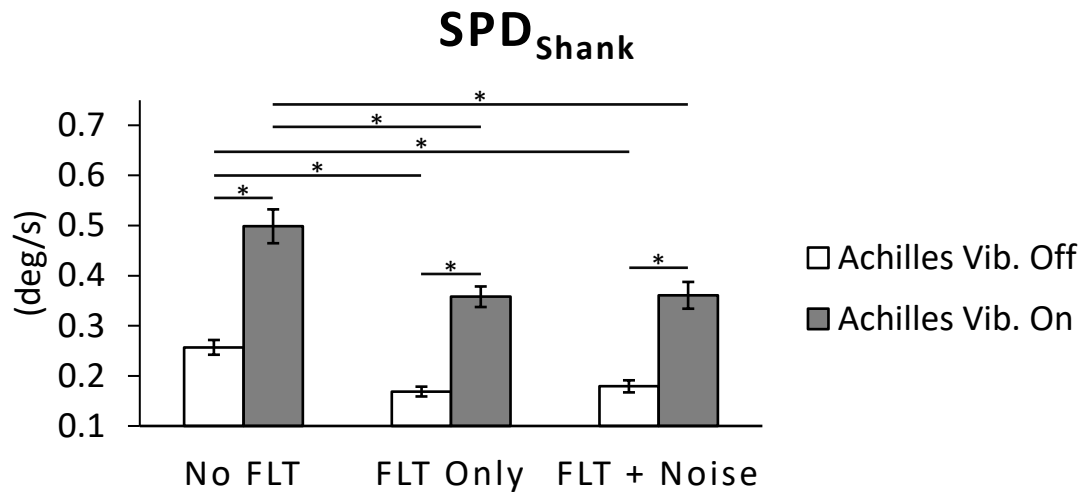


Figure 4.3.3.4 Shank mean angular speed for subjects during the six conditions experienced in experiment #2 (n = 22). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

4.3.4 Experiment #2 Angular Demeaned RMS Findings

Achilles vibration by FLT condition pairwise comparisons of RMS_{Head} and RMS_{Shank} were not significantly different for either Achilles vibration comparisons or FLT comparisons. This indicates that angular variability about the mean angular positions for the head and shank were not significantly impacted by either Achilles vibration or FLT conditions (see Figures 4.3.4.1 and 4.3.4.2 below).

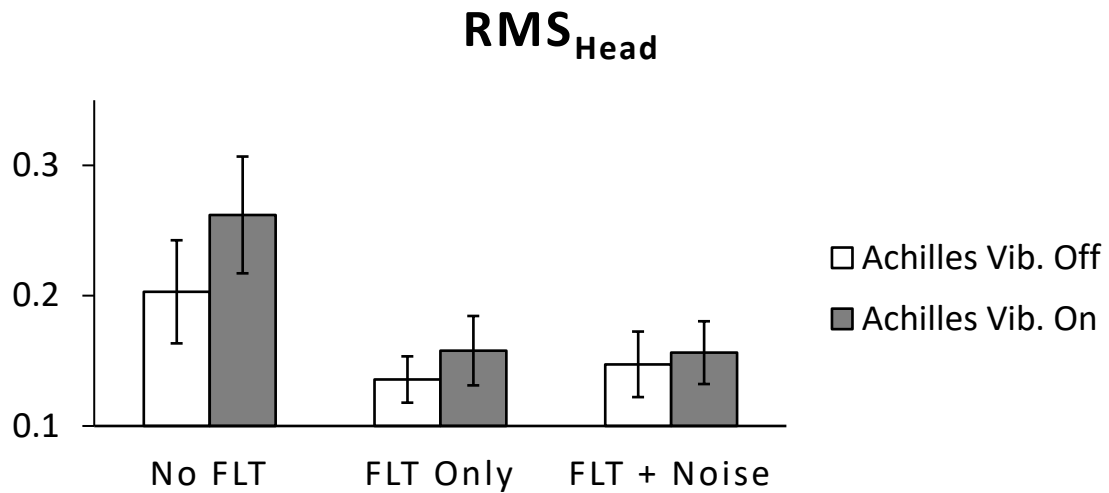


Figure 4.3.4.1 Head demeaned RMS for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE.

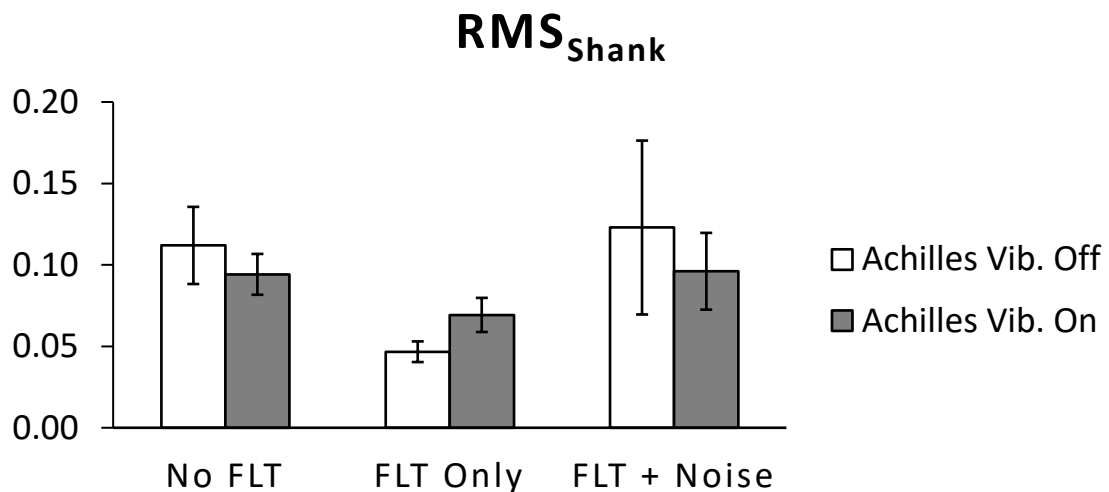


Figure 4.3.4.2 Shank demeaned RMS for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE.

RMS_{Torso} was significantly less with Achilles vibration on than with vibration off, during conditions with No FLT ($p = 0.014$), but not during the conditions with FLT (FLT Only or FLT + Noise). FLT significantly reduced RMS_{Torso} compared to No FLT when Achilles vibration was off (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$), but no significant effects of FLT condition on RMS_{Torso} were observed when Achilles vibration was on. These results indicate that angular variability about the torso mean angular position was significantly less when Achilles vibration was on, but the greater variability seen with Achilles vibration off was significantly reduced by FLT (both FLT Only and FLT + Noise) such that comparisons between Achilles vibration conditions were not significantly different when FLT was present. Adding noise to the light touch did not significantly affect RMS_{Torso} beyond that effect of FLT itself though (see Figure 4.3.4.3 below).

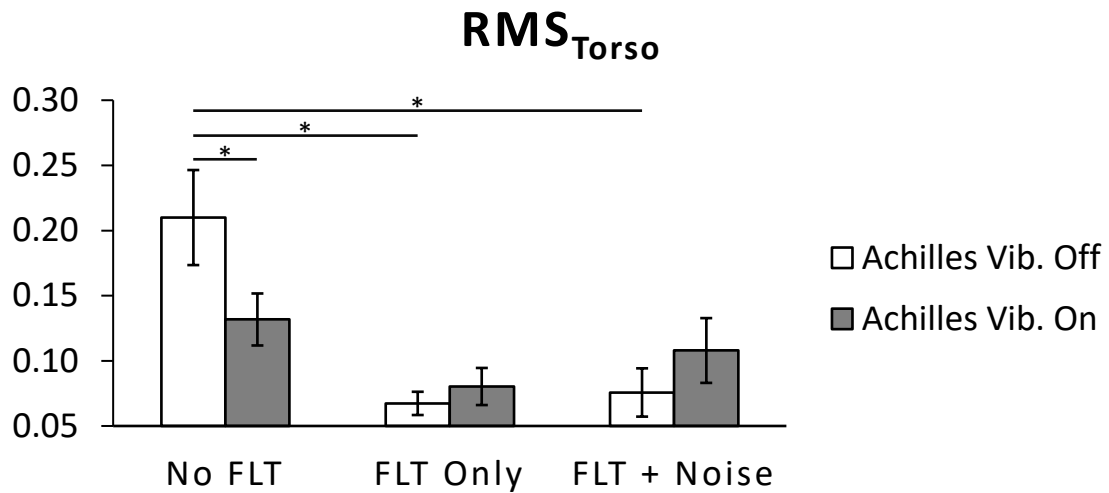


Figure 4.3.4.3 Torso demeaned RMS for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

No significant comparisons between Achilles vibration conditions (on vs. off) were observed for $\text{RMS}_{\text{Thigh}}$; however, FLT did significantly reduce $\text{RMS}_{\text{Thigh}}$ compared to No FLT when Achilles vibration was off (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p = 0.005$). Like the other three segments evaluated, $\text{RMS}_{\text{Thigh}}$ was not significantly impacted by FLT conditions when Achilles vibration was on. These results indicate that angular variability about the thigh mean angular position was significantly reduced by FLT when no Achilles vibration was present, much like it was with the torso. Adding noise to the light touch did not significantly affect $\text{RMS}_{\text{Thigh}}$ beyond that effect of FLT itself though (see Figure 4.3.4.4 below).

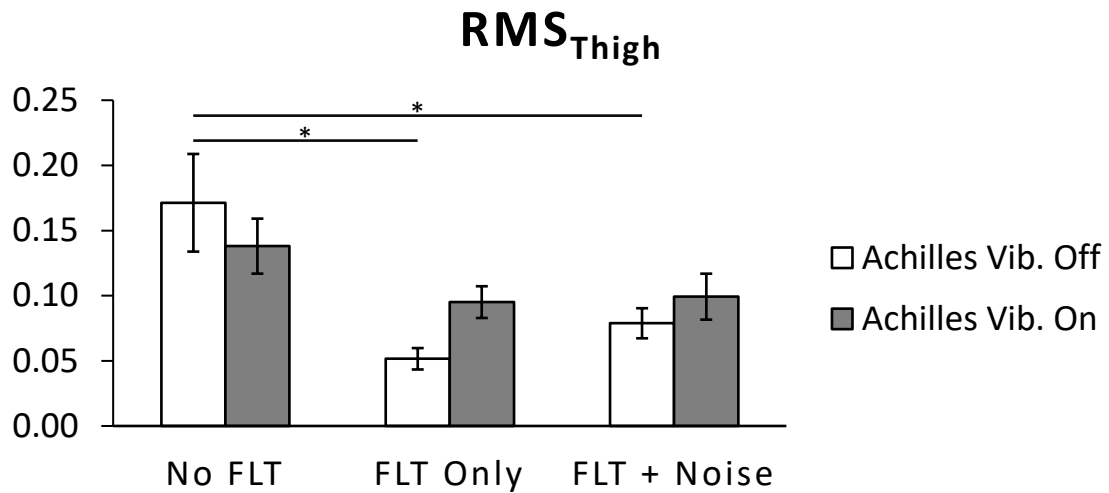


Figure 4.3.4.4 Thigh demeaned RMS for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions when Achilles vibration is not present.

4.3.5 Experiment #2 Anchoring Index Findings

AI_{Head} was not significantly different for either Achilles vibration comparisons or FLT comparisons. This indicates that the “anchoring” or angular motion of the head with respect to the torso was not significantly impacted by either Achilles vibration or FLT conditions (see Figure 4.3.5.1 below).

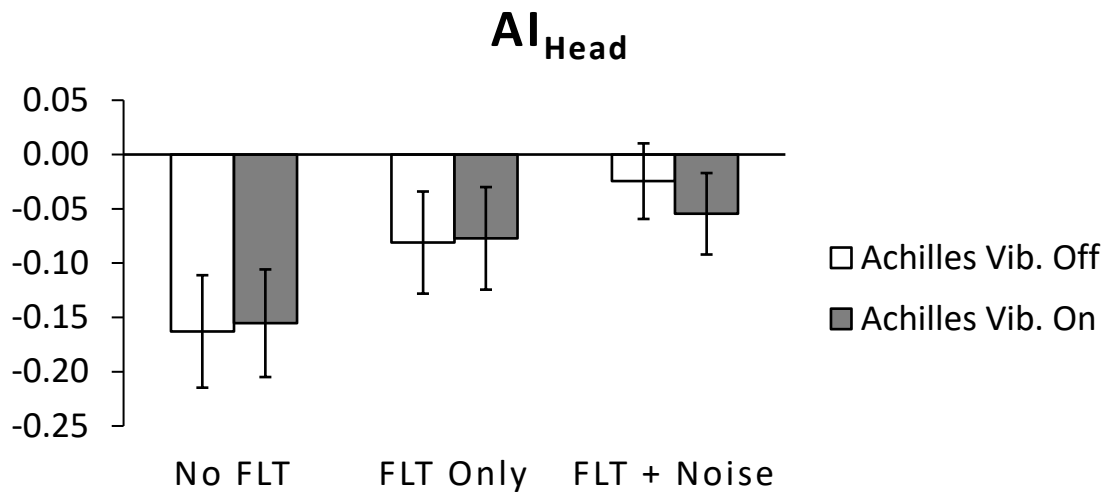


Figure 4.3.5.1 AI_{Head} for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE.

AI_{Torso} was significantly greater with Achilles vibration on compared to Achilles vibration off, regardless of FLT condition (No FLT: $p = 0.029$; FLT Only: $p = 0.036$; FLT + Noise: $p = 0.006$). AI_{Torso} was also significantly increased by FLT compared to No FLT when Achilles vibration was off (FLT Only vs. No FLT: $p = 0.008$; FLT + Noise vs. No FLT: $p = 0.004$), as well as when Achilles vibration was on (FLT Only vs. No FLT: $p = 0.011$; FLT + Noise vs. No FLT: $p = 0.001$). These results indicate that the presence of Achilles vibration caused the torso to be significantly less anchored to its inferior segment (the thigh) than in space. The presence of FLT further significantly decreased the anchoring of the torso to the thigh, regardless of Achilles vibration condition (both off and on). In fact, when FLT conditions were present along with Achilles vibration, stabilization of the trunk to the thigh was decreased such that there was more predominant torso stabilization in space than on the thigh. Adding noise to the light touch did not

significantly affect Al_{Torso} beyond the effects of FLT itself though (see Figure 4.3.5.2 below).

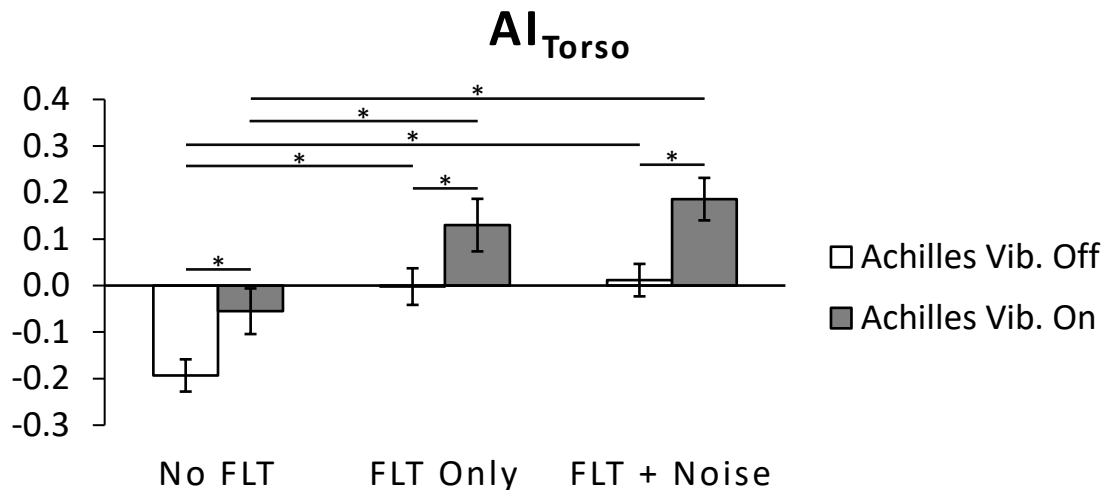


Figure 4.3.5.2 Al_{Torso} for subjects during the six conditions experienced in experiment #2 ($n = 22$). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

Only two significant pairwise comparisons for the Al_{Thigh} occurred. Al_{Thigh} was significantly greater with Achilles vibration off compared to Achilles vibration on ($p = 0.025$) during the FLT + Noise condition. Al_{Thigh} was also significantly greater with FLT + Noise compared to No FLT ($p = 0.009$) when Achilles vibration was off. These results indicate that the presence of Achilles vibration generally did not affect the anchoring of the thigh to the shank. The effect of FLT when combined with noise (FLT + Noise) however did significantly decrease anchoring of the thigh to the shank when Achilles vibration was not present, such that anchoring of the thigh to the shank was also significantly reduced with

Achilles vibration off than with Achilles vibration on during the FLT + Noise conditions (see Figure 4.3.5.3 below).

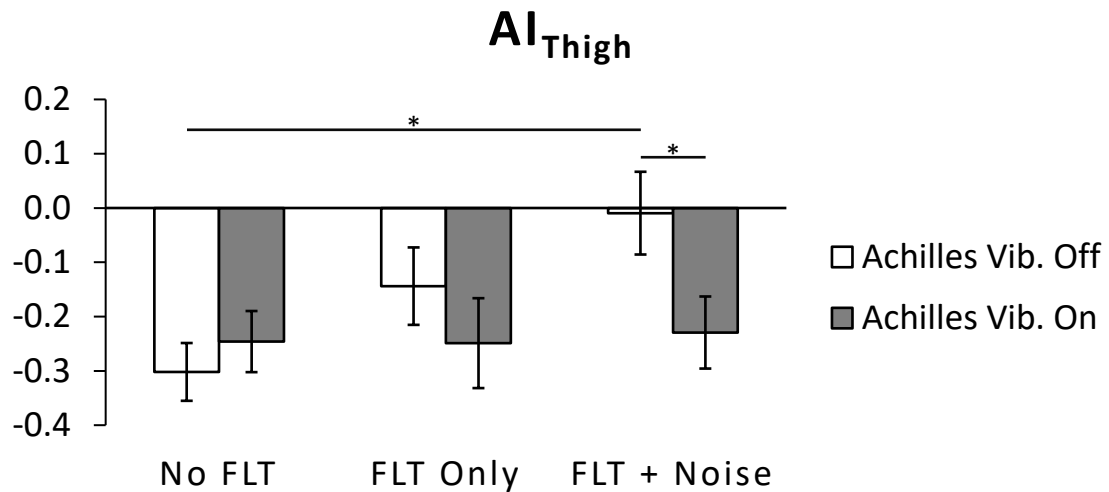


Figure 4.3.5.3 AI_{Thigh} for subjects during the six conditions experienced in experiment #2 (n = 22). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between Achilles vibration and FLT conditions.

4.4 Experiment #3

During experiment #3, the MANOVA revealed significant main effects of arm vibration [$F(44, 356) = 10.189, p < 0.001$] and FLT condition [$F(44, 356) = 3.022, p < 0.001$]. A significant interaction effect of arm vibration and FLT condition was also found [$F(88, 720) = 2.37, p < 0.001$], and several arm vibration by FLT condition pairwise comparisons did yield significant findings in the metrics.

4.4.1 Experiment #3 CoP Findings

When FLT was present (both FLT Only and FLT + Noise), POS_{CoP} was significantly more anterior with biceps vibration than with triceps vibration (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$) as well as with biceps vibration than with no arm vibration (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$). POS_{CoP} was also significantly more anterior with FLT than with No FLT when biceps vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). When triceps vibration was present, POS_{CoP} was significantly less anterior with FLT than with No FLT (FLT Only vs. No FLT: $p = 0.004$; FLT + Noise vs. No FLT: $p = 0.026$). These findings indicate that arm vibration (both biceps and triceps) did not modulate POS_{CoP} until it was combined with FLT. Biceps vibration significantly moved POS_{CoP} forward compared to no arm vibration during FLT conditions. Although Triceps vibration tended to move POS_{CoP} backward compared to no arm vibration during FLT conditions, it was only significantly more backward when compared to biceps vibration conditions (not the no arm vibration conditions). Adding noise to the light touch did not significantly affect POS_{CoP} beyond the effects of FLT itself (see Figure 4.4.1.1 below).

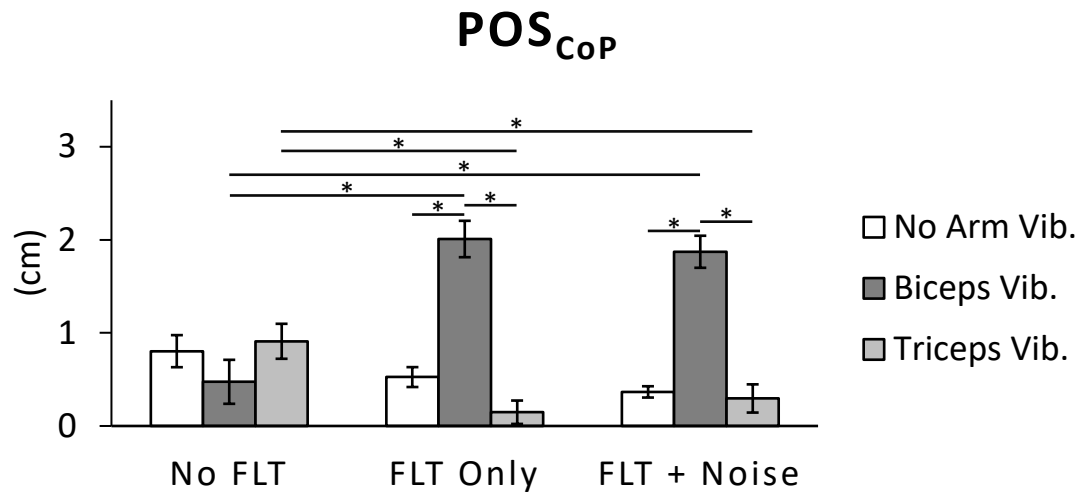


Figure 4.4.1.1 A/P CoP mean position for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in cm with error bars of ± 1 SE. Positive values depict a more anterior position. Asterisks denote significant comparisons between arm vibration and FLT conditions.

Comparisons revealed SPD_{CoP} was not significantly impacted by either arm vibration or FLT conditions (see Figure 4.4.1.2 below).

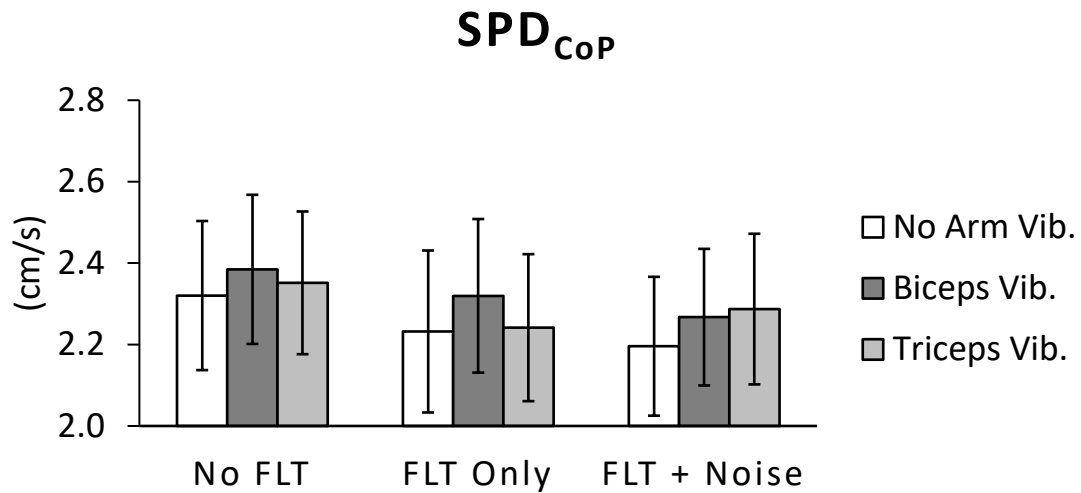


Figure 4.4.1.2 A/P CoP mean speed for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in cm/s with error bars of ± 1 SE.

RMS_{CoP} was not significantly different between arm vibration conditions during any of the FLT conditions; however, it was significantly less during conditions with FLT than with No FLT when biceps vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). It was also significantly less during conditions with FLT than with No FLT when no arm vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). These findings indicate that like Achilles vibration (from experiment #1 and #2), arm vibration did not have a significant effect on the variability of CoP motion about the mean. When both biceps and no arm vibration conditions were present, variability of CoP motion about the mean was significantly decreased by FLT conditions (both FLT only and FLT + Noise). RMS_{CoP} also tended to be decreased in triceps vibration conditions by FLT, but these comparisons did not

reach significance. Additionally, combining noise with the light touch did not significantly affect RMS_{CoP} beyond the effect of FLT itself (see Figure 4.4.1.3 below).

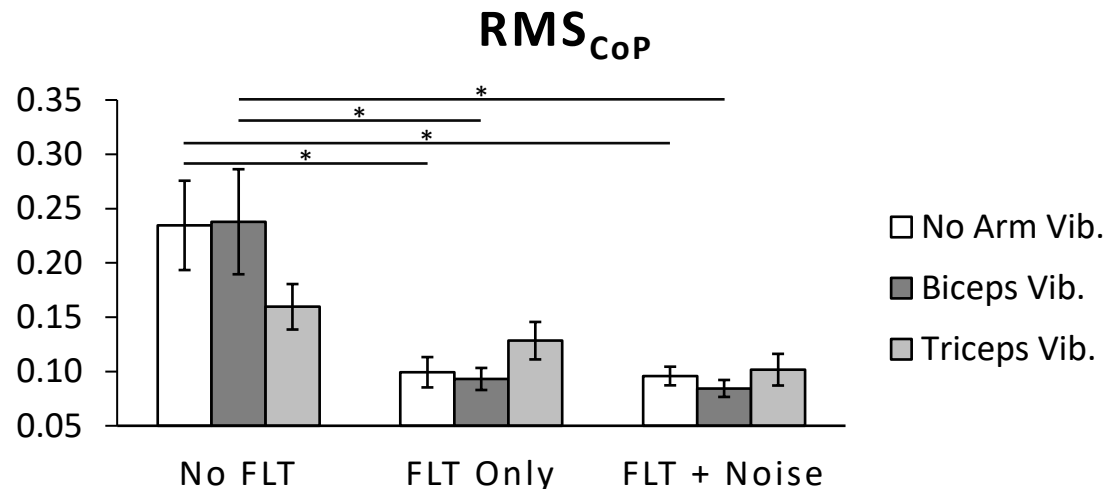


Figure 4.4.1.3 A/P CoP demeaned RMS for subjects during the nine conditions experienced in experiment #3 (n = 23). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions.

$ApEn_{CoP}$ was significantly less with biceps vibration present than no arm vibration as well as triceps vibration, during conditions with FLT present (No Arm Vibration: FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$; Triceps Vibration: FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$). When no arm vibration, as well as when triceps vibration was present, $ApEn_{CoP}$ was significantly greater during conditions with FLT than the conditions with No FLT present (No Arm Vibration: FLT Only vs. No FLT, $p < 0.001$; FLT + Noise vs. No FLT, $p < 0.001$; Triceps Vibration: FLT Only vs. No FLT, $p = 0.002$; FLT + Noise vs. No FLT, $p = 0.001$).

These findings indicate that when triceps vibration and no arm vibration conditions were present, the regularity of CoP motion was significantly decreased by FLT conditions (both FLT only and FLT + Noise). Furthermore, FLT conditions significantly decreased the regularity of CoP motion when triceps and no arm vibration conditions were present compared to biceps vibration conditions, as $ApEn_{CoP}$ during biceps vibration was generally unchanged by FLT conditions. However, the addition of noise to the light touch did not significantly affect regularity of CoP motion beyond the effects of FLT itself (see Figure 4.4.1.4 below).

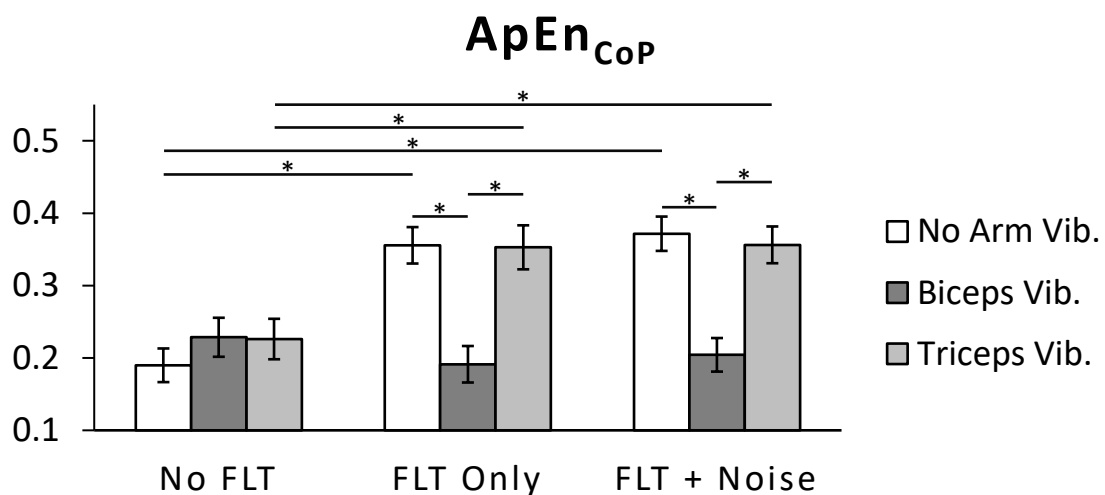


Figure 4.4.1.4 A/P CoP approximate entropy for subjects during the nine conditions experienced in experiment #3 (n = 23). Means are represented with error bars of ± 1 SE. Asterisks denote significant comparisons between arm vibration and FLT conditions.

4.4.2 Experiment #3 Mean Angular Position Findings

POS_{Head} was significantly more flexed with biceps vibration than with triceps vibration, during conditions with FLT (FLT Only: $p < 0.001$; FLT + Noise: $p = 0.004$). POS_{Head} was also significantly more flexed with biceps vibration than with no arm vibration, during the FLT Only condition ($p = 0.002$). Effects of FLT Only were also observed, with POS_{Head} being significantly more flexed with FLT Only than No FLT when biceps vibration was present ($p = 0.019$) and significantly less flexed with FLT Only than No FLT when triceps vibration was present ($p = 0.047$). These results indicate that arm vibration was able to significantly affect POS_{Head}, but only when FLT was present. With FLT, biceps vibration moved POS_{Head} more anteriorly, while anterior POS_{Head} was reduced with FLT and triceps vibration present. However, adding noise to the light touch did seem to reduce the effects of arm vibration and light touch on POS_{Head}, such that the effect of biceps vibration was no longer significantly different than the no arm vibration condition, and both biceps and triceps vibration conditions were not significantly different from their No FLT conditions, when FLT + Noise was present (see Figure 4.4.2.1 below).

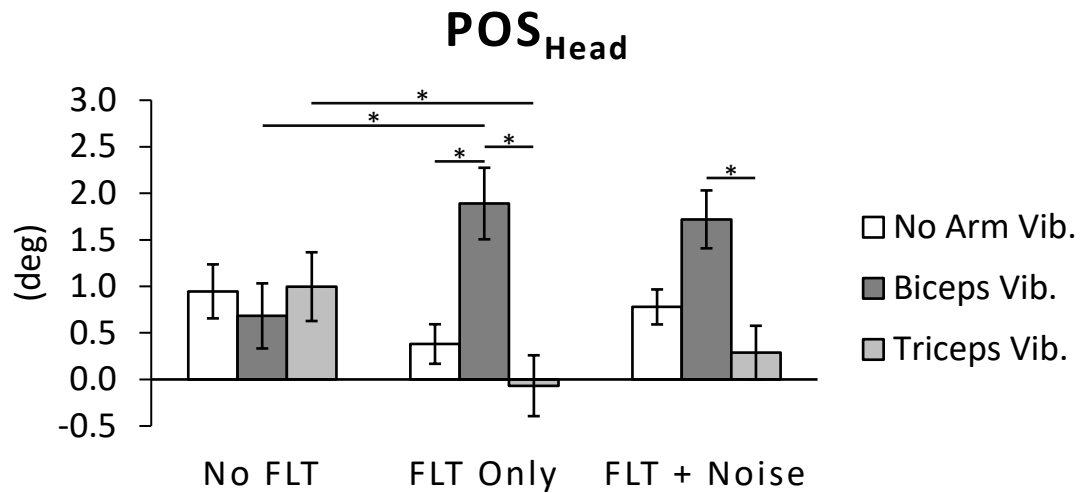


Figure 4.4.2.1 Head mean angular position for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in degrees with error bars of ± 1 SE. Positive values depict head flexion, while negative values depict head extension. Asterisks denote significant comparisons between arm vibration and FLT conditions.

POS_{Torso} was significantly more extended with biceps vibration than with triceps vibration, during conditions with No FLT ($p = 0.026$). However when FLT was present, POS_{Torso} was significantly more flexed with biceps vibration than triceps vibration (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$) and significantly more flexed with biceps vibration than no arm vibration (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$). Effects of FLT were also observed, with POS_{Torso} being significantly more flexed with FLT than No FLT when biceps vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$) and significantly extended with FLT than No FLT when triceps vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$).

These results indicate that arm vibration was able to significantly affect POS_{Torso} , but like the head, primarily when FLT was present. Biceps vibration moved POS_{Torso} significantly more anterior, while triceps vibration moved POS_{Torso} more posterior when combined with FLT conditions. However, the addition of noise to the light touch did not significantly affect POS_{Torso} beyond the effects of FLT itself (see Figure 4.4.2.2 below).

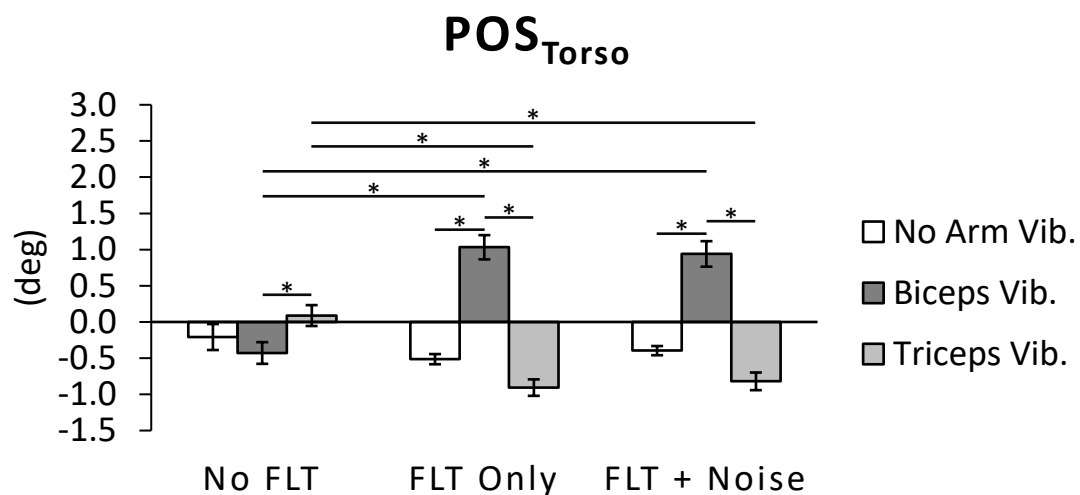


Figure 4.4.2.2 Torso mean angular position for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in degrees with error bars of ± 1 SE. Positive values depict torso flexion, while negative values depict torso extension. Asterisks denote significant comparisons between arm vibration and FLT conditions.

When FLT was present, POS_{Thigh} showed significantly more knee extension with biceps vibration than triceps vibration (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$) and significantly more knee extension with biceps vibration than no arm vibration (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT:

$p < 0.001$). Effects of FLT were also observed, with POS_{Thigh} revealing significantly more knee extension with FLT than No FLT when biceps vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). POS_{Thigh} also showed significantly less knee extension with FLT Only than with No FLT ($p = 0.016$), when triceps vibration was present. These results indicate that arm vibration was able to significantly affect POS_{Thigh}, but again, only when FLT was also present. Biceps vibration moved POS_{Thigh} significantly more anterior, while triceps vibration moved POS_{Thigh} less anterior when combined with FLT conditions. Adding noise to the light touch however did seem to generally reduce the effects of arm vibration and light touch on POS_{Thigh}, such that the triceps vibration condition with FLT + Noise was not significantly different from its No FLT condition (see Figure 4.4.2.3 below).

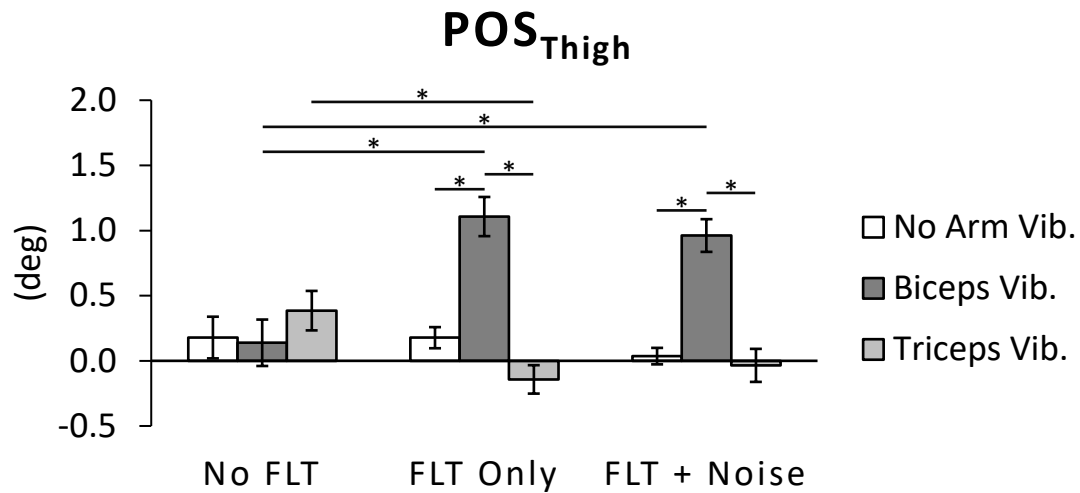


Figure 4.4.2.3 Thigh mean angular position for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in degrees with error bars of ± 1 SE. Negative values depict knee flexion, while positive values depict knee extension. Asterisks denote significant comparisons between arm vibration and FLT conditions.

When FLT was present, POS_{Shank} showed significantly more ankle dorsiflexion with biceps vibration than triceps vibration (FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$) and significantly more ankle dorsiflexion with biceps vibration than no arm vibration (FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$). Effects of FLT were also observed, with POS_{Shank} revealing significantly more ankle dorsiflexion with FLT than No FLT when biceps vibration was present (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$). These results indicate that biceps vibration significantly affected POS_{Shank}, but only when FLT was also present. Biceps vibration moved POS_{Shank} significantly more anterior when combined with FLT conditions. However, the addition of noise to the light

touch did not significantly affect POS_{Shank} beyond the effect of FLT itself (see Figure 4.4.2.4 below).

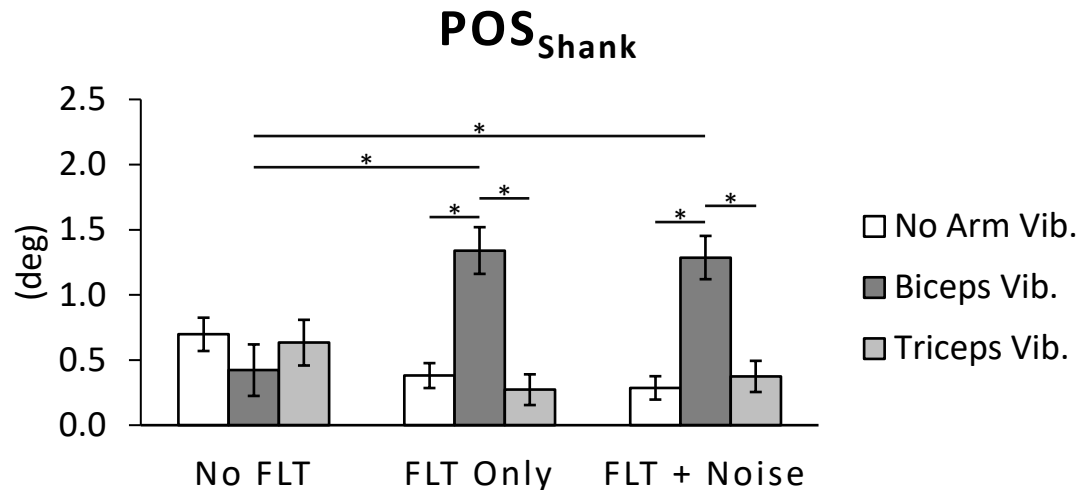


Figure 4.4.2.4 Shank mean angular position for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in degrees with error bars of ± 1 SE. Positive values depict ankle dorsiflexion. Asterisks denote significant comparisons between arm vibration and FLT conditions.

POS_{Elbow} was significantly more flexed with biceps vibration than triceps vibration and significantly more flexed with biceps vibration than no arm vibration, regardless of FLT condition (Biceps vs. Triceps Vibration: No FLT, $p < 0.001$; FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$; Biceps vs. No Vibration: No FLT, $p < 0.001$; FLT Only, $p < 0.001$; FLT + Noise, $p < 0.001$). POS_{Elbow} was also significantly more extended with triceps vibration than no arm vibration when No FLT was present ($p = 0.032$). Effects of FLT showed POS_{Elbow} being significantly less flexed with FLT than No FLT when biceps vibration was present (FLT Only

vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$) and significantly less extended with FLT than No FLT when triceps vibration was present (FLT Only vs. No FLT: $p = 0.017$; FLT + Noise vs. No FLT: $p = 0.008$). These results indicate that biceps vibration significantly flexed the elbow regardless of FLT condition, and triceps vibration significantly extended the elbow without the presence of FLT. The effects of arm vibration were significantly reduced by FLT conditions however, so much so that triceps vibration did not yield significant differences in POS_{Elbow} from no arm vibration conditions when FLT was present. The addition of noise to the light touch did not significantly affect POS_{Elbow} beyond the mitigating effects FLT had on arm vibration responses itself though (see Figure 4.4.2.5 below).

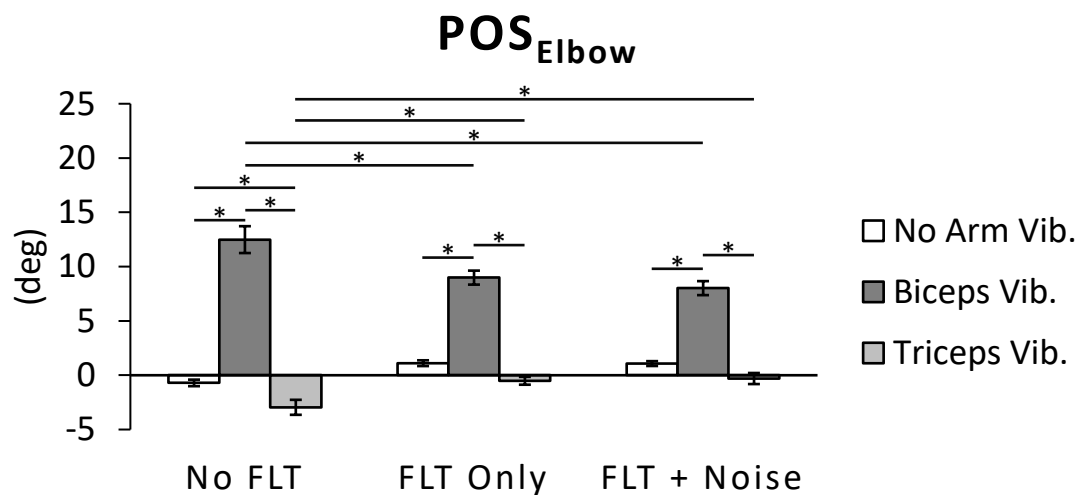


Figure 4.4.2.5 Right elbow mean angular position for subjects during the nine conditions experienced in experiment #3 ($n = 23$). Values are in degrees with error bars of ± 1 SE. Positive values depict elbow flexion, while negative values depict elbow extension. Asterisks denote significant comparisons between arm vibration and FLT conditions.

4.4.3 Experiment #3 Mean Angular Speed Findings

SPD_{Head} was not significantly impacted by arm vibration conditions.

Without arm vibration present however, SPD_{Head} was significantly slower during conditions with FLT than during No FLT (FLT Only vs. No FLT: $p = 0.045$; FLT + Noise vs. No FLT: $p = 0.011$). SPD_{Head} during arm vibration conditions were also generally slower during conditions with FLT than with No FLT, but not significantly slower. These findings showed FLT could significantly reduce SPD_{Head} when no vibration was present, much like it was able to reduce all other body segment mean angular speeds from experiment #2. Adding noise to the light touch did not significantly affect SPD_{Head} beyond the effect of FLT slowing the speed itself though (see Figure 4.4.3.1 below).

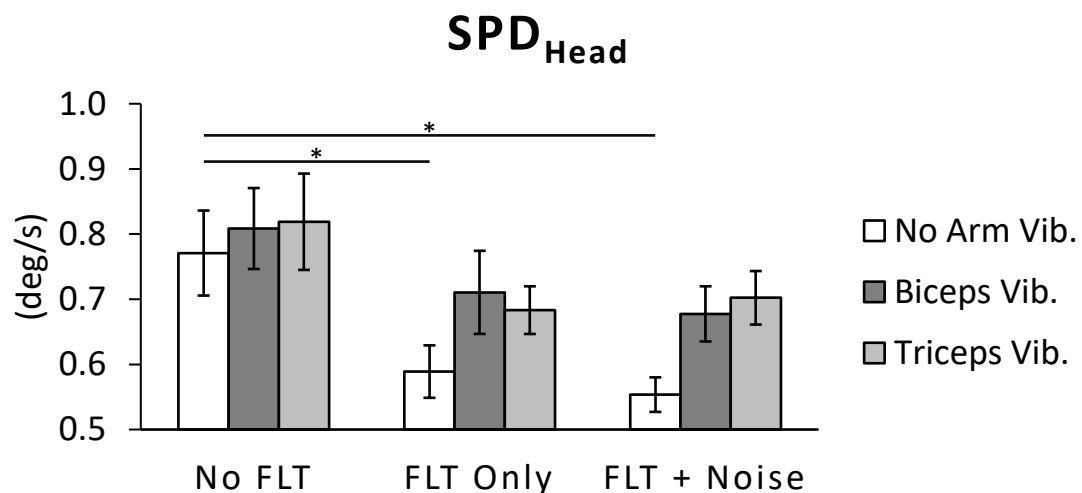


Figure 4.4.3.1 Head mean angular speed for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions when no arm vibration is present.

SPD_{Torso} was also not significantly impacted by arm vibration conditions. However, SPD_{Torso} was significantly slower with FLT than No FLT during conditions with biceps vibration (FLT Only vs. No FLT: $p = 0.008$; FLT + Noise vs. No FLT: $p = 0.002$) and no arm vibration (FLT Only vs. No FLT: $p = 0.005$; FLT + Noise vs. No FLT: $p = 0.001$). SPD_{Torso} during triceps vibration was also generally slower during FLT conditions than with No FLT, but not significantly slower. These findings showed that FLT can significantly reduce SPD_{Torso} during conditions with biceps vibration, much like the way FLT reduced SPD_{Torso} during conditions with Achilles vibration from experiment #2. The finding from experiment #2 that SPD_{Torso} was significantly reduced by FLT under conditions with no vibration was also confirmed here. Adding noise to the light touch did not significantly affect SPD_{Torso} beyond the effect of FLT slowing the speed itself though (see Figure 4.4.3.2 below).

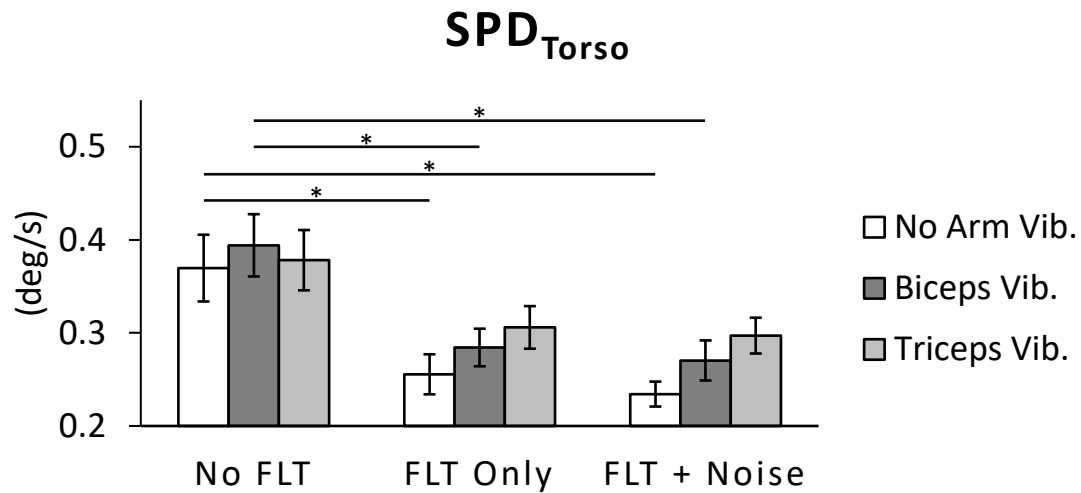


Figure 4.4.3.2 Torso mean angular speed for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions.

Like the head and torso, SPD_{Thigh} was not significantly impacted by arm vibration conditions. SPD_{Thigh} was significantly slower with FLT than No FLT during conditions with biceps vibration (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$) and no arm vibration (FLT Only vs. No FLT: $p < 0.001$; FLT + Noise vs. No FLT: $p < 0.001$) though, like the torso. SPD_{Thigh} during triceps vibration was also significantly slower during the FLT + Noise condition than with No FLT ($p = 0.006$). These findings showed that FLT can significantly reduce SPD_{Thigh} during conditions with biceps vibration, much like the way FLT reduced SPD_{Thigh} during conditions with Achilles vibration from experiment #2, and the finding from experiment #2 that SPD_{Thigh} was significantly reduced by FLT under conditions with no vibration was also confirmed here. Furthermore,

adding noise to the light touch was able to further slow the SPD_{Thigh} when triceps vibration was present, such that SPD_{Thigh} was significantly slower with FLT + Noise than with No FLT, when triceps vibration was present.

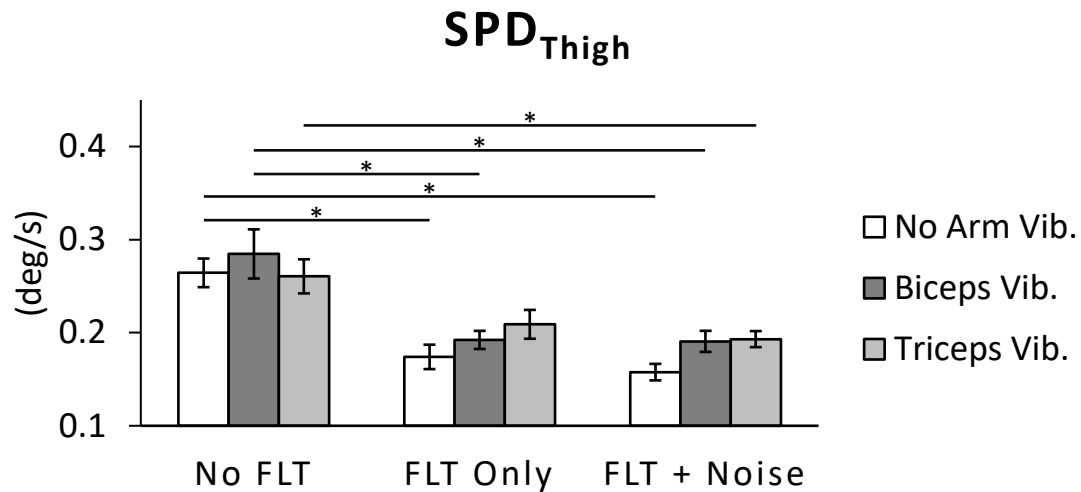


Figure 4.4.3.3 Thigh mean angular speed for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions.

SPD_{Shank} was significantly faster with biceps vibration than with no arm vibration during the FLT + Noise condition ($p = 0.042$), but no other significant effects of arm vibration were noted. Like the other three body segment angles located in the sagittal plane (head, torso, and thigh), SPD_{Shank} was significantly slower with FLT than No FLT during conditions with no arm vibration (FLT Only vs No FLT: $p = 0.003$; FLT + Noise vs. No FLT: $p = 0.001$). SPD_{Shank} during biceps vibration was also significantly slower during the FLT Only condition than

with No FLT ($p = 0.019$). These findings generally showed that FLT reduces SPD_{Shank} , especially during conditions with no arm vibration or biceps vibration.

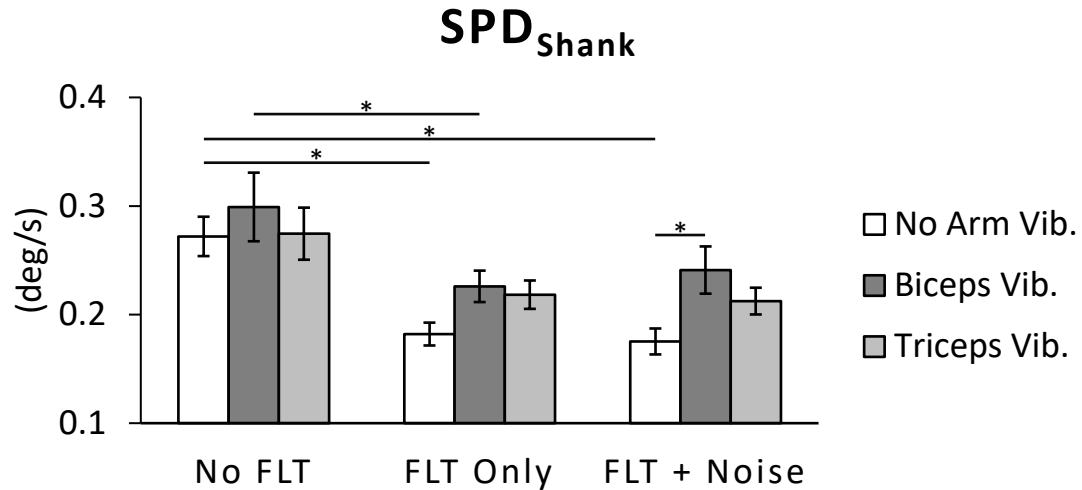


Figure 4.4.3.4 Shank mean angular speed for subjects during the nine conditions experienced in experiment #3 ($n = 23$). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between arm vibration and FLT conditions.

SPD_{Elbow} was significantly faster with biceps vibration and triceps vibration conditions than no arm vibration conditions, regardless of FLT condition (Biceps vs. No Vibration: No FLT, $p < 0.001$; FLT Only, $p = 0.009$; FLT + Noise, $p = 0.003$; Triceps vs. No Vibration: No FLT, $p < 0.001$; FLT Only, $p = 0.04$; FLT + Noise, $p = 0.01$). For effects of FLT, SPD_{Elbow} was significantly slower with FLT than with No FLT, during conditions when triceps vibration was present (FLT Only vs. No FLT: $p = 0.008$; FLT + Noise vs. No FLT: $p = 0.022$). SPD_{Elbow} was also significantly slower with FLT Only than with No FLT during conditions when biceps vibration was present ($p = 0.034$). These results indicate that arm

vibration significantly increased SPD_{Elbow} regardless of FLT condition, but adding FLT to conditions with arm vibration somewhat reduced these increased speeds (see Figure 4.4.3.5 below).

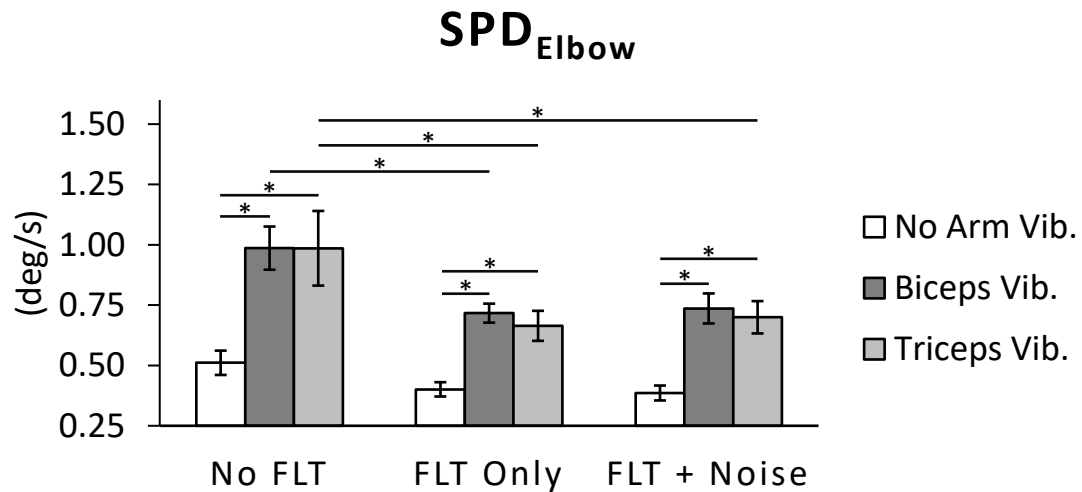


Figure 4.4.3.5 Right elbow mean angular speed for subjects during the nine conditions experienced in experiment #3 (n = 23). Values are in deg/s with error bars of ± 1 SE. Asterisks denote significant comparisons between arm vibration and FLT conditions.

4.4.4 Experiment #3 Angular Demeaned RMS Findings

Arm vibration by FLT condition pairwise comparisons of RMS_{Head} and RMS_{Shank} yielded no significant comparisons. This indicates that angular variability about the mean angular positions for the head and shank were not significantly impacted by either arm vibration or FLT conditions (see Figures 4.4.4.1 and 4.4.4.2 below).

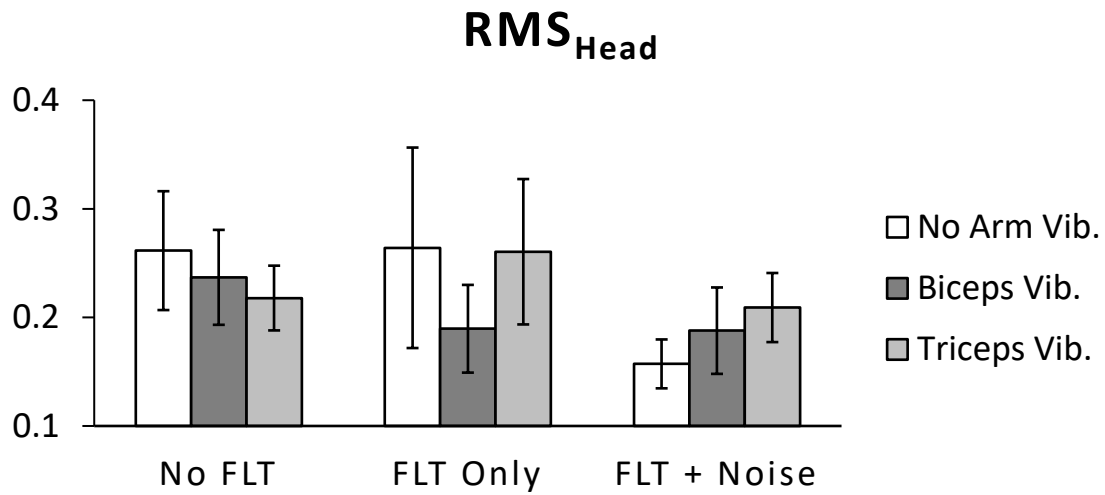


Figure 4.4.4.1 Head demeaned RMS for subjects during the nine conditions experienced in experiment #3 (n = 23). Mean values are depicted with error bars of ± 1 SE.

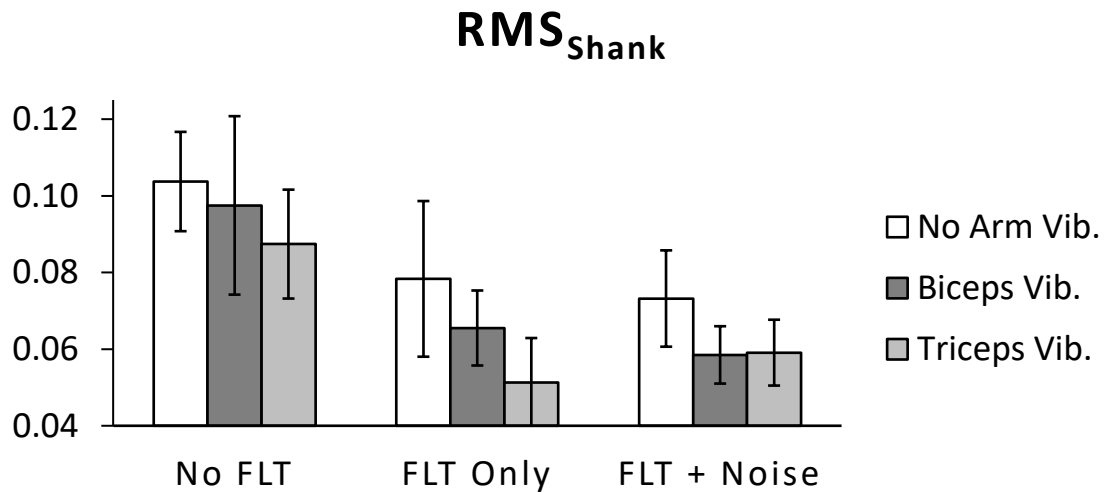


Figure 4.4.4.2 Shank demeaned RMS for subjects during the nine conditions experienced in experiment #3 (n = 23). Mean values are depicted with error bars of ± 1 SE.

RMS_{Torso} was significantly less with biceps vibration than with no arm vibration, during conditions with No FLT ($p = 0.02$), much like the way Achilles vibration reduced RMS_{Torso} in experiment #2. The effect from experiment #2 of FLT significantly reducing RMS_{Torso} under conditions with no vibration was also confirmed here, with RMS_{Torso} for FLT conditions (FLT Only: $p < 0.001$; FLT + Noise: $p < 0.001$) being significantly less than the No FLT condition when no arm vibration was present. Additionally, RMS_{Torso} was found to be significantly reduced with FLT Only compared to No FLT during conditions when triceps vibration was present ($p = 0.026$). These results indicate that under conditions with No FLT, angular variability about the torso mean angular position was generally less when arm vibration was present, especially for biceps vibration compared to no arm vibration. However, the presence of FLT seemed to reduce angular variability about the torso mean angular position even more so than arm vibration, especially during conditions with no arm vibration (see Figure 4.4.4.3 below).

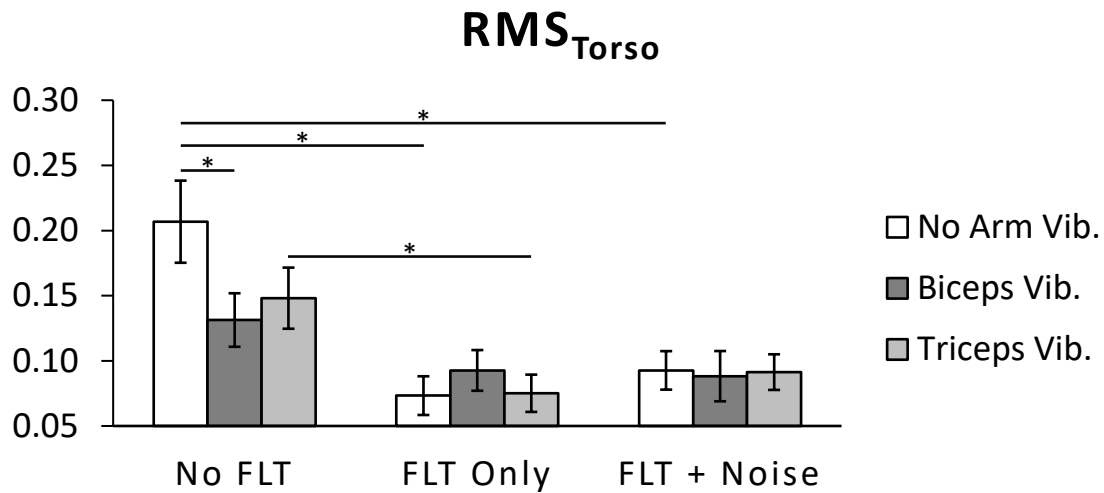


Figure 4.4.4.3 Torso demeaned RMS for subjects during the nine conditions experienced in experiment #3 (n = 23). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between arm vibration and FLT conditions.

RMS_{Thigh} was significantly reduced by arm vibration conditions, with both biceps vibration ($p = 0.029$) and triceps vibration ($p = 0.003$) having significantly less RMS_{Thigh} than the no arm vibration condition when No FLT was present. RMS_{Thigh} was also significantly less during FLT conditions than during No FLT conditions when biceps vibration (No FLT vs. FLT Only: $p = 0.008$; No FLT vs. FLT + Noise: $p = 0.005$) was present as well as when no vibration (No FLT vs. FLT Only: $p < 0.001$; No FLT vs. FLT + Noise: $p < 0.001$) was present. Similar to findings for the torso, these results indicate that angular variability about the thigh mean position was significantly reduced by arm vibration conditions when not contacting a stable surface. The result from experiment #2 that angular variability about the thigh mean angular position was significantly reduced by FLT

without vibration present was also confirmed here, and thigh angular variability can likewise be reduced by FLT when biceps vibration is present. Adding noise to the light touch did not significantly affect RMS_{Thigh} beyond the effects of FLT itself though (see Figure 4.4.4.4 below).

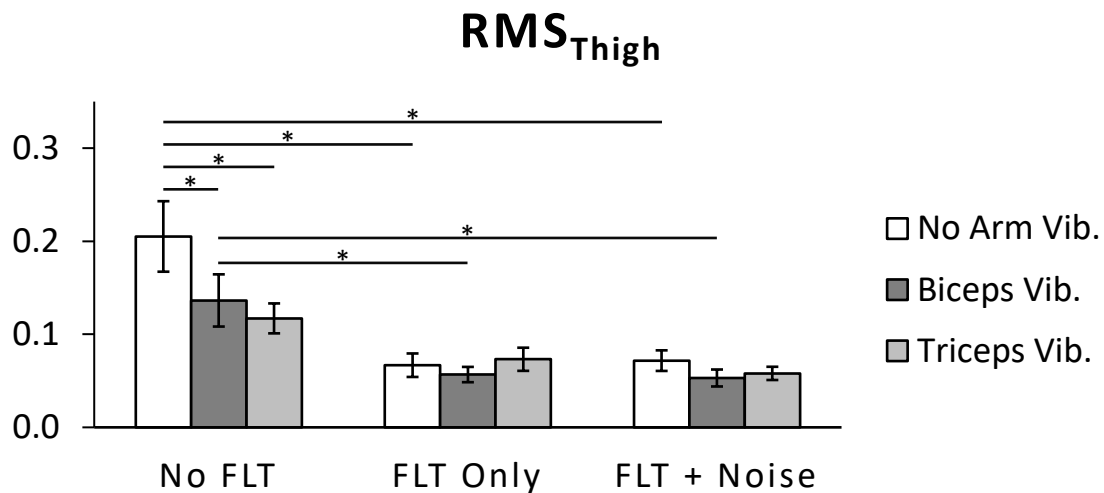


Figure 4.4.4.4 Thigh demeaned RMS for subjects during the nine conditions experienced in experiment #3 (n = 23). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between arm vibration and FLT conditions.

RMS_{Elbow} was significantly increased with biceps vibration compared to no arm vibration, during both the No FLT ($p = 0.001$) and FLT Only ($p = 0.026$) conditions. RMS_{Elbow} was also significantly increased with biceps vibration compared to triceps vibration, during the No FLT ($p < 0.001$) condition. The sole significant effect of FLT came when FLT Only significantly increased RMS_{Elbow} compared to No FLT when triceps vibration was present ($p = 0.043$). These results indicate that the angular variability about the right elbow mean position

was significantly increased by biceps vibration even when FLT Only was present, but not by triceps vibration. However, the addition of noise to the FLT reduced elbow angular variability during biceps vibration such that the biceps vibration condition was no longer significantly different from the no arm vibration condition during FLT + Noise conditions. Angular variability about the right elbow mean position was also significantly increased when triceps vibration was engaged with FLT Only compared to the triceps vibration condition with No FLT. However, the addition of fingertip noise also mitigated this effect, such that the triceps vibration condition with FLT + Noise was not significantly different from the triceps vibration condition with No FLT (see Figure 4.4.4.5 below).

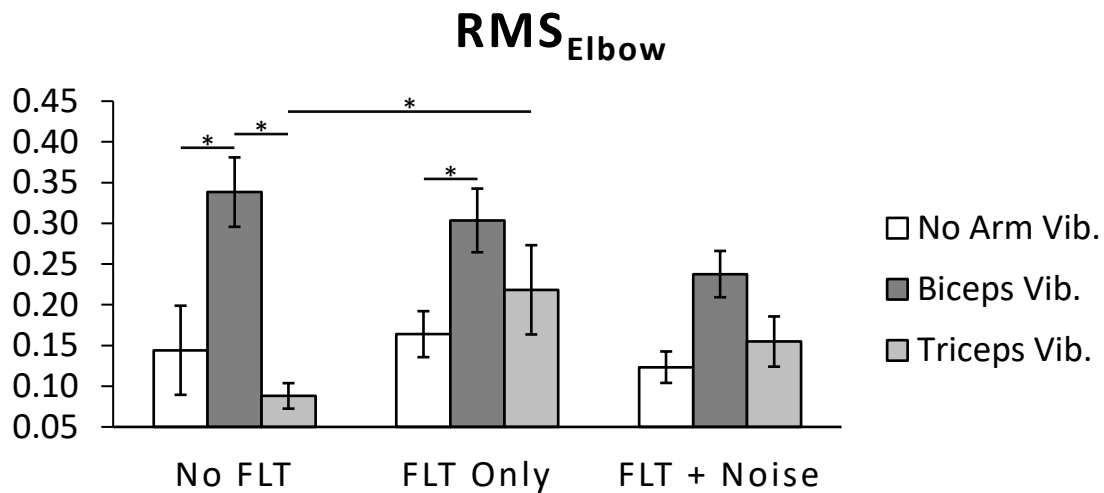


Figure 4.4.4.5 Right elbow demeaned RMS for subjects during the nine conditions experienced in experiment #3 (n = 23). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between arm vibration and FLT conditions.

4.4.5 Experiment #3 Anchoring Index Findings

AI_{Head} was only found to be significantly different between arm vibration conditions (biceps vs triceps vibration) during the FLT Only condition, with biceps vibration having a significantly more negative AI_{Head} than the triceps vibration condition ($p = 0.004$). This indicates that when FLT Only was present, the head was significantly more anchored to the torso with biceps vibration present, than it was with triceps vibration present. However, when noise was added to the light touch, this effect of biceps vibration was reduced such that it was no longer significantly different (see Figure 4.4.5.1 below).

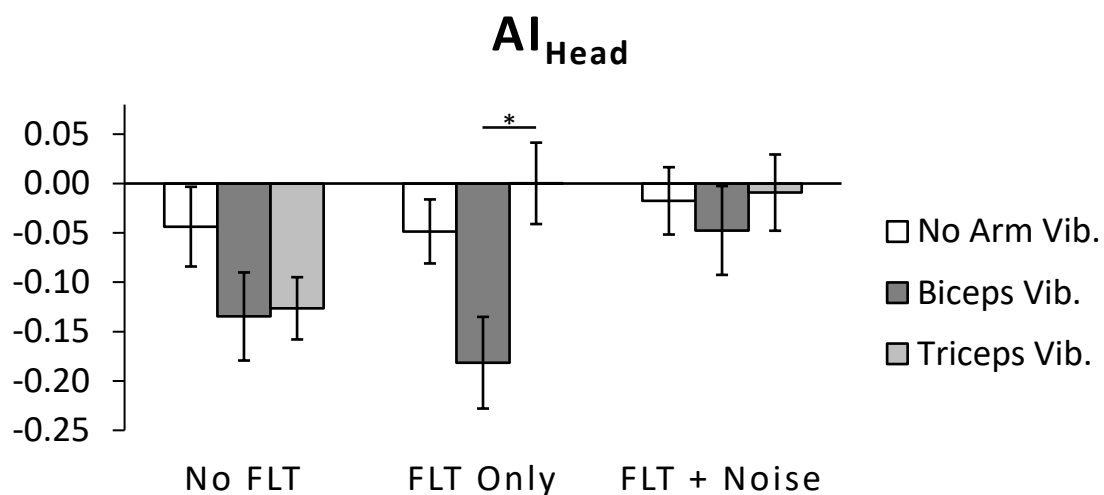


Figure 4.4.5.1 AI_{Head} for subjects during the nine conditions experienced in experiment #3 ($n = 23$). Mean values are depicted with error bars of ± 1 SE. Asterisk denotes significant comparison between biceps and triceps vibration conditions when only FLT is present.

AI_{Torso} was significantly less negative during FLT conditions when no arm vibration was present (No FLT vs. FLT Only: $p < 0.001$; No FLT vs. FLT + Noise:

$p < 0.001$). These results confirmed the finding from experiment #2 that FLT significantly decreased the anchoring of the torso to the thigh when no vibration was present. Unlike Achilles vibration however, arm vibration conditions did not significantly impact the anchoring of the torso to the thigh, regardless of FLT condition. Furthermore, adding noise to the light touch conditions did not significantly affect Al_{Torso} beyond the effect of FLT itself (see Figure 4.4.5.2 below).

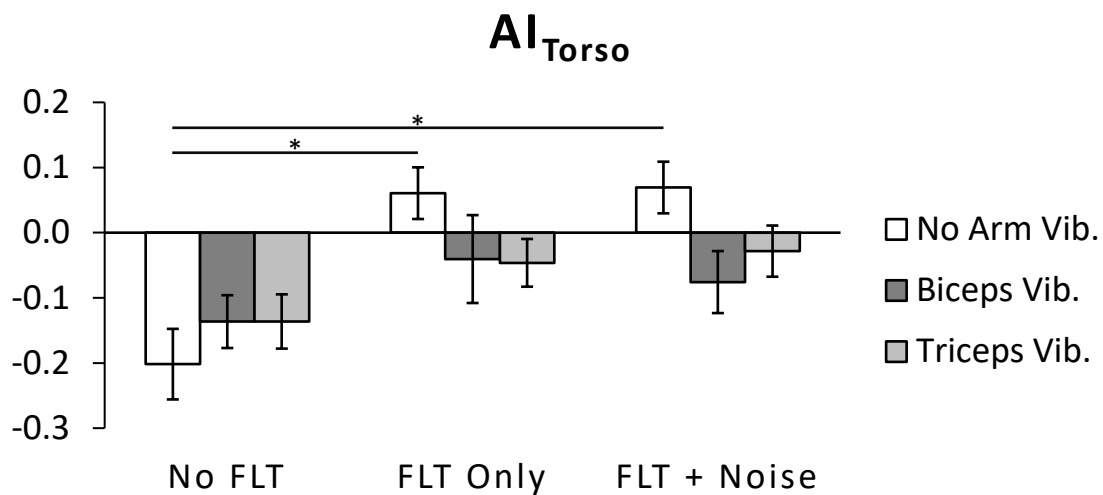


Figure 4.4.5.2 Al_{Torso} for subjects during the nine conditions experienced in experiment #3 ($n = 23$). Mean values are depicted with error bars of ± 1 SE. Asterisks denote significant comparisons between FLT conditions when no arm vibration is present.

Al_{Thigh} was not significantly different for either arm vibration or FLT comparisons. This indicates that the anchoring of the thigh to the shank was not significantly impacted by either arm vibration or FLT conditions (see Figure 4.4.5.3 below).

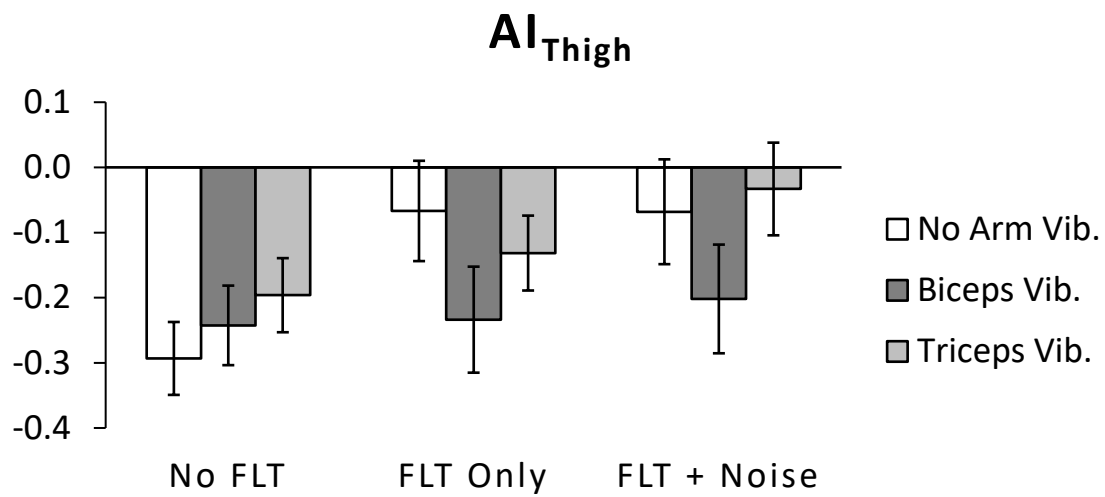


Figure 4.4.5.3 AI_{Thigh} for subjects during the nine conditions experienced in experiment #3 ($n = 23$). Mean values are depicted with error bars of ± 1 SE.

V. DISCUSSION

The goal of these experiments was to further examine the concept of sensory reweighting, which is often explained in a holistic sense as occurring among the three sensory systems responsible for postural control (visual, vestibular, and somatosensory), based on the reliability of their input (Dettmer, 2014; Haran & Keshner, 2009; R J Peterka, 2002). However, such an explanation often fails to note the various sensory receptor inputs available in each of the three sensory modalities, and the potential for reweighting to occur within a single modality itself, i.e. between different receptors within that modality. Specifically, within the same modality, different receptors may be providing competing stimuli. We tested effects of combinations of three different stimuli within the same modality (tendon vibration, FLT, and mechanical noise) on postural control. Tendon vibration preferentially activated muscle spindle type Ia afferent fibers in the calf in experiments 1-2 as well as in the biceps and triceps in experiment #3. Mechanical noise was utilized in an attempt to evoke SR and enhance cutaneous receptor inputs at the bottom of the foot in experiment #1 and at the fingertip in experiments #2 and #3. FLT was also utilized in experiments #2 and #3 to determine if cutaneous inputs from the fingertip modified the postural responses from tendon vibration.

We proposed several directional hypotheses across the three experiments, which supported the notion of reweighting occurring within the somatosensory system. If sensory reweighting was occurring within the somatosensory system, we assumed that a combination of two or more stimuli

would result in significant changes to postural control compared to conditions with just one stimulus. Conversely, if sensory reweighting did not occur within the somatosensory system itself and the entire somatosensory system was up-weighted or down-weighted as a whole compared to the visual or vestibular modalities that were unaffected, we expected that the strongest stimuli would dominate postural control effects and the addition of other somatosensory stimuli would essentially be ignored or unable to significantly impact posture. It is important to note that our hypotheses for these experiments were directional, and in some cases the direction of postural metrics' responses to certain stimuli were significantly different but in the opposite direction than was hypothesized. Thus, although some of our hypotheses were not supported, we interpret significant findings directionally opposite to our hypothesized direction of change as providing support for the notion of reweighting occurring within the somatosensory system.

5.1 Main Effects of Stimuli

The first section of this discussion will focus on the effects that each type of stimulus utilized by itself had on postural control. Although all three types of stimuli utilized were able to significantly impact postural control by themselves, there were clearly some stimuli that had more pronounced effects than others.

5.1.1 Tendon Vibration Effects

The typical response of Achilles vibration shifting CoP and angular body motion backwards (Eklund 1972; Ivanenko et al. 1999; Ceyte et al. 2007; Caudron et al. 2010a; Caudron et al. 2010b; Duclos et al. 2014; Kabbaligere et

al. 2017) was confirmed in this study. Achilles vibration caused POS to shift significantly in a posterior direction for CoP and all body segments (head, torso, thigh, and shank) tested in both experiments #1 and #2, compared to the baseline conditions without any stimuli. These shifts in POS caused by contraction of the musculature attached to the tendons being vibrated were also evident in experiment #3, where POS_{Elbow} was significantly flexed for biceps vibration and significantly extended for triceps vibration, compared to the baseline. POS_{Elbow} shifts caused by vibration did not occur equally however, as the effect of biceps vibration was much larger than the effect of triceps vibration. Some studies have also noted unequal effects of tendon vibration when applied to different sides of a joint (Roll and Vedel 1982; Caudron et al. 2010a), which may be due to other biomechanical constraints. It is notable that without FLT to a stable surface being present to link upper body finger and arm inputs to postural sway for the rest of the body, POS for the CoP, head, torso, thigh, and shank were not significantly affected by arm vibration alone.

SPD was also significantly increased by Achilles vibration for the CoP and at all body segments tested (head, torso, thigh, and shank) in both experiments #1 and #2, compared to baseline. This indicated greater body motion and sway with Achilles vibration as others have reported (Hatzitaki et al. 2004; Caudron et al. 2010b). During experiment #3 SPD_{Elbow} was also significantly increased by arm vibration (both biceps and triceps) compared to baseline; however, SPD for the CoP and the body segments utilized in postural control (head, torso, thigh,

and shank) were unaffected by the presence of arm vibration, as long as the arm was not linked to a stable surface through FLT.

Achilles vibration generally did not impact variability of motion about the mean positions (the demeaned RMS) for any metrics in the first two experiments other than RMS_{TorsO} in experiment #2. RMS_{TorsO} actually revealed a decrease in variability caused by Achilles tendon vibration in experiment #2. The RMS of CoP motion has been previously shown to be significantly increased by vibrating musculature about the ankles (Kabbaligere et al., 2017; Temple et al., 2014); however, it is possible that prior significant changes in RMS were driven primarily by the large directional shift of the CoP motion (POS_{CoP} shifting posteriorly for Achilles vibration or anteriorly for tibialis vibration). To account for this possibility, RMS of the data in the current study was demeaned by subtracting the absolute value of the POS. Therefore, we do not believe the findings of no effect of Achilles vibration on demeaned RMS in the current study are inconsistent with our previous research calculating RMS (Temple et al., 2014). Similar to Achilles vibration in experiment #2, RMS_{TorsO} was also significantly reduced with biceps vibration in experiment #3, and both arm vibration conditions (biceps and triceps) had significantly less RMS_{Knee} compared to baseline conditions. Despite some postural sway variability generally decreasing with arm vibration, biceps vibration actually significantly increased variability (RMS_{Elbow}) compared to the baseline and triceps vibration conditions.

ApEn of the CoP was generally unaffected by tendon vibration itself; however, Achilles tendon vibration tended to have lower ApEn values than the

baseline conditions in experiments 1-2, suggesting greater regularity and less postural complexity with Achilles tendon vibration, as has been noticed in both children and elderly individuals (Newell & Molenaar, 1998). Conversely, the arm vibration conditions in experiment #3 (both biceps and triceps vibration) tended to have higher ApEn values than baseline, suggesting more irregularity and greater complexity with arm tendon vibration.

AI_{Thigh} for experiment #1 and AI_{Torso} for experiment #2 were both significantly less negative with Achilles vibration compared to their baseline. These findings suggest a decrease in the “anchoring” of the thigh and torso to their inferior segments, the shank and thigh respectively, during conditions with Achilles vibration. Much of the previous research has primarily utilized AI as a means for investigating head stabilization on the trunk during postural control (Amblard et al., 2001; Assaiante & Amblard, 1993; R. Mills, Levac, & Sveistrup, 2018; R. S. Mills & Sveistrup, 2018; Sveistrup et al., 2008); however, AI for the A/P motion of various body segments (head, trunk, and thigh) have been shown to be significantly impacted by cutaneous torso vibration (B.-C. Lee, Martin, Ho, et al., 2013). To our knowledge, the current study is the first to note significant changes in AI_{Thigh} and AI_{Torso} as a result of Achilles vibration. Ultimately, these changes in thigh coordination on the shank (AI_{Thigh}) and torso coordination on the thigh (AI_{Torso}) induced by Achilles vibration does support the notion that the typical ankle strategy associated with postural control for normal quiet stance is somewhat altered by vibration of ankle joint musculature (Temple et al., 2014). Thus, the inverted pendulum model often used to characterize body sway during

postural control (Maurer et al., 2006; Wall & Kentala, 2010; Winter et al., 1993), may not be appropriate for scenarios in which vibration is introduced to ankle musculature, as there appears to be some breaking in the pendulum from the torso to the shank. The varying angular POS values across body segments (shank, thigh, torso, and head) during Achilles vibration in this study also support the notion of sway less akin to an inverted pendulum, as the POS values for an inverted pendulum's segments would generally be the same. These slight modifications to the postural control strategy seen in our subjects are similar to ones previously found with Achilles vibration (Thompson et al., 2007). Such strategy modification utilizes more musculature than an inverted pendulum implementing a strict ankle strategy, thus Achilles vibration may induce less efficient means of postural control as is often observed in the elderly and in those with fear of falling (Adkin, Frank, Carpenter, & Peysar, 2000; Fay B Horak, 2006; Maki, Edmondstone, & McIlroy, 2000). Arm vibration itself from experiment #3 however, did not significantly modify AI, thus the inverted pendulum model appeared to be more intact during arm vibration conditions than it was with Achilles vibration.

Overall, our results confirmed that tendon vibration is a strong proprioceptive stimulus capable of modifying body sway when applied to the Achilles tendon. Arm vibration also provides a strong proprioceptive stimulus affecting elbow joint motion; however, when the arm is not linked to a stable surface through FLT, the effects of arm vibration on postural control are minimal. Regarding sensory reweighting, the main effects of tendon vibration suggest that

a weighting of muscle spindle inputs impacted by the tendon vibration is applied to postural control when the muscle spindle inputs are relevant to postural control. For example, calf muscle spindle afferents are used for postural control during Achilles vibration, and the disrupted afferents from the calves cause observable modifications to balance (more posterior POS values, increased SPD, less anchoring of superior segments to inferior segments, etc.).

Conversely, if the vibration is applied to musculature not providing relevant information for postural control, such as for arm vibration without FLT, the erroneous muscle spindle afferents in the arm may affect elbow motion; however, these inputs do not generally affect postural control. Without FLT, the arm muscle spindle afferents are not salient for postural control, and it appears that they are largely down-weighted or ignored for balance purposes.

5.1.2 Light Touch Effects

FLT by itself did not significantly change POS for CoP or any of the body segments measured in experiments #2 and #3. However, variability about the mean positions was reduced as can be noted by the reduced POS standard error bars and significantly less RMS_{CoP} , RMS_{Torso} , and RMS_{Thigh} observed with FLT compared to the baseline without touch in experiments #2 and #3 (refer to Figures in the Results for the CoP: 4.3.1.3 and 4.4.1.3, Torso: 4.3.4.3 and 4.4.4.3, Thigh: 4.3.4.4 and 4.4.4.4). Likewise, SPD was also significantly decreased by FLT in all body segments (head, torso, thigh, and shank). These findings all support the common observation of FLT alone without providing

biomechanical support, being able to significantly reduce postural sway (Baldan et al., 2014; John J. Jeka, 2016).

Novel for this study were the findings of the effect of FLT on ApEn and AI, suggesting that FLT is capable of modifying postural control strategy.

Somatosensory input from FLT appeared to enable the use of additional joint degrees of freedom, allowing for a more complex postural control strategy (Newell & Molenaar, 1998). This notion is suggested by ApEn results that indicated greater CoP irregularity and complexity brought on by FLT. ApEn was significantly greater with FLT than without in experiments #2 and #3, suggesting that the addition of posture relevant somatosensory input to the fingertip itself may be enough to change inverted pendulum behavior by altering the typical ankle strategy associated with A/P postural control during quiet stance (F B Horak & Nashner, 1986; L. M. Nashner, Shupert, Horak, & Black, 1989; Temple et al., 2014; Winter et al., 1993). AI_{Torso} results from experiments #2 and #3 further supported this notion that FLT alone can modify postural control strategy, as the torso became significantly less anchored to the thigh as an effect of FLT. Our findings that FLT seems to have increased the degrees of freedom, leading to greater system complexity and modified postural strategies, is consistent with previous FLT research suggesting additional postural control musculature may be triggered by touch contact and contribute to the reduced sway observed under FLT conditions (John J. Jeka, 2016).

In summary, the main effects of FLT from experiments #2 and #3 further confirm findings of decreased postural sway by the simple addition of

somatosensory inputs provided throughout the arm and fingertip of the limb touching the stable surface (Baldan et al., 2014; John J. Jeka, 2016). These inputs, which were irrelevant for postural control without FLT, suddenly become relevant while touching the stable surface. Thus, the reduced sway seen with FLT suggests that an up-weighting of somatosensory inputs from the limb utilizing FLT was responsible for the modified postural control. In addition, our $ApEn$ and AI_{Torso} findings also support the notion that this up-weighting of new inputs from the FLT can trigger additional postural control musculature in sections such as the trunk to counteract sway, as has been previously suggested (John J. Jeka, 2016). Afferents from the feet as well as ankle musculature, which are primarily thought responsible for somatosensory cues for balance maintenance during quiet stance, may actually be down-weighted in the presence of FLT, as inputs higher up at arm level (refer to Figure 2.3 from the introduction) and finer cutaneous acuity at the fingertip may provide more sensitive postural sway information (John J. Jeka, 2016).

5.1.3 Mechanical Noise Effects

In this study, mechanical noise was the stimuli with the least pronounced effect on postural control in all three experiments. In fact, by itself, mechanical noise only had a significant main effect on posture for the SPD_{CoP} metric in experiment #1. Contrary to the reduced SPD that was expected if SR were occurring, foot noise significantly increased SPD_{CoP} compared to the baseline condition without noise. This was unexpected, as foot noise likely inducing SR has generally been shown to reduce CoP speeds (Hijmans et al., 2008; Lipsitz et

al., 2015; Zhou et al., 2016) and path length (Dettmer et al., 2015). Often there is a significant effect of foot noise reducing these CoP metrics in those having compromised somatosensation, with statistically insignificant effects in young healthy populations. Essentially, a ceiling effect may occur where compromised somatosensory systems have greater ability to reduce CoP motion through SR, but fully intact systems from young, healthy individuals cannot be improved upon. Our finding of foot noise significantly *increasing* SPD_{CoP} is contradictory to previous findings though (Aboutorabi et al., 2017; Dettmer et al., 2015; Hijmans et al., 2008; Lipsitz et al., 2015; Zhou et al., 2016). It is possible that foot noise utilized in this experiment was at amplitudes not ideal for inducing SR in many of the healthy, young subjects. If noise amplitudes were too high, they could have impeded the ability of the mechanoreceptors at the bottom of the foot to detect pressure input utilized in balance maintenance (Harry et al., 2005), possibly explaining the increases seen in SPD_{CoP}. Thus, it is possible that the noise perturbed cutaneous foot mechanoreceptor inputs much like the way Achilles vibration perturbed calf muscle spindle afferents. However, body segment angular speeds did not show significant effects of foot noise, nor did any other metrics besides the SPD_{CoP} to suggest that foot noise may have been compromising balance performance.

When compared to what the addition of FLT produced itself, in experiments #2 and #3 the addition of fingertip noise to the FLT did not produce additional significant effects in any postural control metrics. Thus, the value of adding mechanical noise to improve postural control through SR observed by

some studies (Kimura et al., 2012; Magalhães & Kohn, 2011b; A. a Priplata et al., 2003; A. Priplata et al., 2002), was not apparent in the current research as a main effect. Our findings from all three experiments suggest that overall as a main effect of noise, cutaneous mechanoreceptors in the skin (both at the bottom of the feet and in the fingertip) were not up-weighted to induce SR and improve postural control.

Although the noise introduced largely left postural control unaffected, there is evidence that the noise input was processed by the postural control system. First, there was a significant effect of foot noise increasing SPD_{CoP} as previously stated in experiment #1, suggesting the postural control system was impacted by the foot noise. Secondly, in experiments #2 and #3 the vast majority of the subjects (19 of 22 in experiment #2 and 22 of 23 in experiment #3) correctly reported being able to feel the slightly noisy stimulus in their fingertip during FLT + Noise conditions without tendon vibration. This self-reporting by the subjects indicated that the noise was entering the system at a level above the perceptual threshold for most subjects, which was due to the fact that our subjects were young, healthy, and most of them could detect with their fingertip the lowest level of stimulus the tactors were capable of producing when assessed during the threshold testing. Finally, there were several interaction effects with mechanical noise where insignificant main effects of either tendon vibration (Achilles, biceps, or triceps) or FLT suddenly became significant, or where significant effects of either tendon vibration or FLT suddenly became insignificant, when the presence of mechanical noise (either at the foot or fingertip) was added. These interaction

effects with noise present suggest that the noise played some role in combination with either the tendon vibration or FLT conditions to produce different postural control than was observed solely as an effect of the tendon vibration or FLT. This possibility is further discussed in section 5.2.1 below.

5.2 Interaction Effects

5.2.1 Tendon Vibration × Noise

5.2.1.1 Achilles Vibration × Foot Noise (Experiment #1)

Our hypothesis for experiment #1 that we would see less posterior POS values, decreases in speeds, and reductions in variability (RMS) when insole noisy tactor vibration was present with Achilles tendon vibration than with Achilles tendon vibration alone was not confirmed. POS, SPD, and RMS were all unaffected during Achilles vibration and foot noise conditions compared to conditions with Achilles vibration alone, except for SPD_{CoP}, which was significantly *increased* when adding the noise (refer to Figure 4.2.1.2 in the Results). Foot noise thus led to an overall increase in postural motion as indicated by the increased SPD_{CoP}.

However, foot noise also could have promoted greater complexity in the postural control system. ApEn was highest with the presence of foot noise, consistent with other SR research indicating greater CoP complexity during foot noise conditions (Costa et al., 2007; Zhou et al., 2016), but that value was significantly decreased when Achilles vibration was added to the foot noise in this study (refer to Figure 4.2.1.4 in the Results). Essentially, foot noise caused CoP motion to become significantly more irregular when no Achilles vibration was

present compared to when the vibration was present. This increased CoP irregularity suggests that when foot noise was present, there may have been greater complexity in the postural control system (Borg & Laxåback, 2010; Haran & Keshner, 2009), with less system constraint (Cavanaugh et al., 2007) and more degrees of freedom being utilized (Newell & Molenaar, 1998) when the Achilles vibration was off compared to when the Achilles vibration was on. Generally, higher ApEn values suggesting greater complexity are often thought to indicate a benefit to postural control, as a system acting with less constraint at the base of support (Cavanaugh et al., 2006; Cavanaugh, Guskiewicz, Giuliani, et al., 2005; Cavanaugh, Guskiewicz, & Stergiou, 2005; Haran & Keshner, 2009; Newell & Molenaar, 1998; Stergiou, Harbourne, & Cavanaugh, 2006). There is not a consensus regarding how the effects of foot noise on CoP motion regularity are to be interpreted if SR is occurring. One study utilizing multiscale entropy analysis found foot noise to significantly increase multiscale complexity in healthy elderly and suggested it was a result of SR occurring (Costa et al., 2007). Another study utilizing ApEn to assess complexity when applying foot noise did not find any significant effects of the noise in either healthy young or elderly individuals (Dettmer et al., 2015). Our significant effect of Achilles vibration found during conditions with foot noise is somewhat in agreement with Costa et al. 2007, in that ApEn was highest during foot noise conditions without Achilles vibration, but when Achilles vibration was also applied with the foot noise, ApEn significantly declined. There was no significant effect of foot noise itself increasing ApEn in the current study however, and thus whether or not SR

actually occurred in experiment #1 is still debatable. More research on how SR impacts postural regularity is warranted, if measures such as ApEn are to be validated as a means of inferring SR occurrence in the future.

Overall, foot noise did not curb the perturbing postural response of Achilles vibration in the vast majority of the metrics, and in our young healthy sample SR did not appear to contribute to better postural control under Achilles vibration conditions. To our knowledge, the effects of noise applied to the soles on balance in healthy individuals concurrently being perturbed by the strong proprioceptive stimulus of Achilles vibration, has not been previously studied. It is often the case that stronger SR responses are seen in individuals with degraded postural control systems and do not occur in young, healthy individuals with fully intact postural control systems (Dettmer et al., 2015; A. A. Priplata et al., 2006; A. a Priplata et al., 2003; A. Priplata et al., 2002). The subjects in our study were all young and healthy, and this may explain our general lack of results suggesting the presence of SR occurring during foot noise conditions. If our subjects' fully intact somatosensory systems were responsible for SR not occurring however, then it can also be inferred that although Achilles vibration is disruptive to postural control in young healthy individuals, it is not disruptive in a manner that allows foot SR to be more effective. Thus, utilizing Achilles vibration as an analog to imitate individuals with compromised somatosensory systems who tend to see greater effects of foot SR improving postural control (e.g. elderly, stroke patients, neuropathy patients, etc.), would not be warranted.

In regard to sensory reweighting, we hypothesized that foot noise would allow SR to increase reliability of plantar foot mechanoreceptor input such that the erroneous calf muscle spindle input provided by Achilles vibration would be down-weighted in favor of the more accurate foot mechanoreceptor input when both stimuli were present. Metrics composed from the individual angular body segment data did not support this notion of reweighting within somatosensation. Furthermore, there was little evidence to support the notion that the foot noise utilized in experiment #1 induced SR in the healthy subjects tested. However, some CoP metrics, which tend to characterize the postural control system as a whole (as opposed to analyzing individual segments), did suggest potential reweighting, even if that reweighting was not a result of SR occurring.

Both the effect of Achilles vibration and foot noise by themselves generated significantly faster SPD_{CoP} , and furthermore, the magnitude of their increases were about the same (~ 1 cm/s). Without somatosensory reweighting occurring, it could be speculated that the effect of both stimuli on SPD_{CoP} would be roughly equal to the sum of their individual effects. However, this was not the case in experiment #1. The increase in SPD_{CoP} caused by foot noise occurred to a lesser extent when Achilles vibration was concurrently present, such that there was no longer a significant effect of Achilles vibration under conditions when foot noise was present. This reduced effect under conditions when both stimuli were present suggests that foot mechanoreceptor inputs, calf muscle spindle inputs, or both were down-weighted in order to constrain SPD_{CoP} . To further determine

which direction reweighting may have occurred, the ApEn results may provide some clarity.

ApEn for the CoP motion was another example where the magnitude of the effects (although these effects were not significant) were roughly the same for Achilles vibration and foot noise (~ 0.05); however, these effects differed in direction. Achilles vibration tended to increase regularity, while foot noise tended to decrease regularity. When both stimuli were combined, the trend of Achilles vibration generally dominated the ApEn response, such that Achilles vibration produced significantly greater regularity in CoP motion under conditions when foot noise was present. This effect of Achilles vibration during conditions with foot noise suggests foot mechanoreceptor inputs impacted by the noise may have been down-weighted while calf muscle spindle afferents impacted by the Achilles vibration were up-weighted. Although reweighting may not have occurred in the direction that was initially hypothesized for experiment #1, the findings for SPD_{CoP} and ApEn suggest that a down-weighting of foot mechanoreceptors and an up-weighting of calf muscle spindles may have occurred during conditions with both stimuli present compared to conditions with just foot noise present.

Overall for experiment #1, although the direction that we hypothesized foot noise would modulate the effect of Achilles tendon vibration was not supported, our results did support the notion that normal postural sway exhibited by Achilles tendon vibration can be modified by the addition of tactor noise to the foot. The metrics of SPD_{CoP} , POS_{Head} , and ApEn all showed that the typical effect of

Achilles vibration on postural control could be modified by the presence of foot noise. However, most of our metrics indicated that the foot noise stimulus generally did not have much effect of reducing the disruptive response of Achilles vibration on postural control. The noise stimulus did not compellingly improve (or impede) postural control during conditions either with or without Achilles vibration present.

5.2.1.2 Tendon Vibration × FLT + Noise (Experiments #2 and #3)

Hypothesis 2.2 of experiment #2, in which we expected to see less posterior POS values, decreased SPD, and reduced RMS during Achilles vibration conditions with FLT + Noise conditions compared to Achilles vibration conditions with just FLT, was not confirmed. It is notable that all the POS metrics (CoP, head, trunk, thigh, and shank) of experiment #2 during Achilles vibration conditions did shift less posteriorly during FLT + Noise compared to just FLT; however, none of these shifts reached a statistical significance.

In experiments #2 and #3, as no significant comparisons were revealed between FLT and FLT + Noise conditions, the effects involving fingertip noise generally fit one of four categories: 1) instances where significant differences between tendon vibration conditions (vertical comparisons in graphs from results) with FLT were no longer significant during FLT + Noise conditions, 2) instances where insignificant differences between tendon vibration conditions (vertical comparisons) with FLT became significant during FLT + Noise conditions, 3) instances where significant differences between No FLT and FLT conditions (horizontal comparisons) were no longer significant when comparing No FLT and

FLT + Noise conditions, and 4) instances where insignificant differences between No FLT and FLT conditions (horizontal comparisons) suddenly became significant when comparing No FLT and FLT + Noise conditions. The following four paragraphs discuss findings involving fingertip noise from experiments #2 and #3 that fit the corresponding four categories.

In experiment #3, the metrics of POS_{Head} , RMS_{Elbow} , and AI_{Head} all fit category one, in that significant differences between arm tendon vibration conditions with FLT were no longer significant during FLT + Noise conditions. During conditions with FLT, biceps vibration caused POS_{Head} to become significantly more anterior compared to the no arm vibration condition, but under conditions with FLT + Noise, the effect of biceps vibration was reduced enough to no longer be significantly different than the no arm vibration condition. RMS_{Elbow} was significantly more variable with biceps vibration than with no arm vibration during FLT conditions, yet that effect was also reduced to nonsignificant levels during FLT + Noise conditions. AI_{Head} indicated significantly more anchoring of the head to the torso during FLT conditions with biceps vibration compared to triceps vibration. Under FLT+ Noise conditions though the anchoring to the torso was mitigated such that none of the arm vibration conditions differed significantly from each other. In general, the less anterior position (POS_{Head}) as well as reductions in variability (RMS_{Elbow}) and anchoring of the head to the torso (AI_{Head}) can all be seen as examples where the presence of FLT + Noise helped to mitigate the effects of arm vibration on postural control during the FLT Only conditions.

Category two saw effects for AI_{Thigh} during experiment #2 and SPD_{Ankle} during experiment #3. In experiment #2, AI_{Thigh} was not significantly impacted by Achilles vibration under conditions with FLT; however, when FLT and fingertip noise were both present (FLT + Noise), the condition without Achilles vibration increased such that there was significantly less anchoring of the thigh to the shank without Achilles vibration than there was with vibration. Thus, the insignificant effect of FLT alone, which generally appeared to provide benefit to postural control in experiment #2, was seemingly enhanced by the addition of noise, such that a significant comparison between Achilles vibration conditions (Off vs. On) occurred when noise was present. SPD_{Ankle} in experiment #3 was not significantly impacted by arm vibration during FLT alone; however, during FLT + Noise conditions biceps vibration caused a slight increase in SPD_{Ankle} , such that it became significantly faster than the condition without arm vibration. This increased speed with biceps vibration is an example where fingertip noise seemingly enhanced the disruptive biceps vibration effect.

In category three there were six metrics from experiment #3: POS_{Head} , POS_{Thigh} , SPD_{Ankle} , SPD_{Elbow} , RMS_{Elbow} , and RMS_{Torso} . POS_{Head} as previously mentioned in category one saw a detrimental effect of biceps vibration with FLT be reduced such that the biceps vibration and no arm vibration conditions were no longer significantly different under FLT + Noise conditions. Similarly, the detrimental effects of both biceps (causing more anterior POS_{Head}) and triceps (causing more posterior POS_{Head}) vibration during FLT Only conditions compared to their No FLT conditions was mitigated under FLT + Noise conditions such that

they were no longer significantly different from their No FLT conditions. Likewise, POS_{Thigh} during triceps vibration was significantly shifted posteriorly under the FLT Only condition compared to the No FLT condition; however, that detrimental effect was no longer significant with triceps vibration under the FLT + Noise condition compared to the No FLT condition. SPD_{Ankle} and SPD_{Elbow} both saw significant decreases with biceps vibration during FLT Only conditions compared to the No FLT conditions; however, that effect was no longer significant for biceps vibration conditions during the FLT + Noise compared to the No FLT condition. RMS_{Elbow} saw the benefit of a significant increase from No FLT to FLT Only conditions no longer reaching significance from No FLT to FLT + Noise conditions, for conditions with triceps vibration. However, during the same triceps vibration conditions, RMS_{Torso} significantly decreased with FLT Only compared to the No FLT condition, but that decrease was no longer significant when comparing FLT + Noise to the No FLT condition.

Finally, category four saw two metrics where interactions involving fingertip noise potentially benefitted postural control. For conditions without Achilles vibration in experiment #2, the benefit of adding FLT Only compared to No FLT did not significantly increase AI_{Thigh} by itself, but during the FLT + Noise condition, there was significantly less anchoring of the thigh to the shank. In conditions with triceps vibration for experiment #3, the benefit of FLT reducing SPD_{Thigh} did not significantly occur when comparing the No FLT to the FLT Only condition, but SPD_{Thigh} was significantly reduced with FLT + Noise compared to the No FLT condition.

Overall, as previously stated, there were no main effects of fingertip noise significantly affecting any of the metrics compared to the conditions with FLT alone (FLT Only vs. FLT + Noise). Thus, our fingertip noise stimuli did not significantly add further value or detriment to postural control than the light touch itself, in either conditions with or without tendon vibration present. We believe that these insignificant effects of adding fingertip noise to the light touch mean that weighting of somatosensory afferents coming from the arm during FLT + Noise conditions were left largely unchanged compared to the FLT Only conditions. However, the four categories that were discussed in the paragraphs above do reveal comparisons where changes in significance occurred in tendon vibration and FLT condition interactions that involved fingertip noise. Thus, fingertip noise itself did not have a large enough effect to significantly affect postural control beyond the effect of FLT itself, but it was associated with some interactions between tendon vibration and FLT conditions by either mitigating significant effects or promoting significant effects that were insignificant when tested without fingertip noise. Generally, most of these smaller effects involving fingertip noise tended to affect postural control in a manner that would be considered beneficial. For example, there were instances of reduced angular variability (RMS_{Elbow}), speeds (SPD_{Thigh}), and deviations from baseline mean positions (POS_{Head} , POS_{Thigh}) that all occurred during the presence of fingertip noise, compared to FLT Only conditions. It is possible that a slight effect of SR could have occurred, with fingertip noise enhancing somatosensory afferents

coming from the touch arm and potentially causing a minor up-weighting of these afferents to account for the postural improvements.

5.2.2 Achilles Vibration × FLT (Experiment #2)

In experiment #2 our hypotheses (2.1) that FLT would cause less posterior POS values and decreased speeds, when light fingertip touch was present along with Achilles vibration, compared to conditions with just Achilles vibration, was confirmed. FLT significantly reduced the posterior shift caused by Achilles vibration in the CoP, torso, and thigh POS metrics. Increased postural speeds due to Achilles vibration were also significantly reduced by the presence of FLT in all angular segments assessed in experiment #2 (SPD_{Head} , SPD_{Torso} , SPD_{Thigh} , and SPD_{Shank}), as we hypothesized. These findings were consistent with previous research indicating increased postural sway velocities from Achilles vibration can be reduced by FLT (Houser 2007; Caudron et al. 2010b). FLT alone was responsible for significantly reducing angular segment speeds, and all these reductions remained significant under FLT + Noise conditions. It should be noted that FLT also significantly reduced SPD for the torso, thigh, and shank under conditions with no Achilles vibration, which is consistent with previous findings of light touch reducing sway velocity during quiet stance (Baccini et al., 2007; Balcan et al., 2014; Dickstein et al., 2001).

Variability of postural motion seemed to be largely unaffected by Achilles vibration, as assessed by demeaned RMS in experiment #2. Only RMS_{Torso} revealed a significant effect of Achilles vibration, and it indicated that there was significantly less variability with Achilles vibration than without, during conditions

without touch (refer Figure 4.3.4.3 in the Results). Thus, our hypothesis (2.1) that FLT would reduce increased variability brought on by Achilles vibration was unconfirmed, as Achilles vibration did not increase demeaned RMS. During conditions without Achilles vibration, demeaned RMS was significantly reduced by FLT for the CoP, torso, and thigh (refer to Figures in the Results for the CoP: 4.3.1.3, Torso: 4.3.4.3, and Thigh: 4.3.4.4).

As was discussed under the main effects for light touch and tendon vibration, a breaking in the inverted pendulum model for postural control appeared to occur in interaction effects of Achilles vibration and FLT as well. AI_{Torso} results suggest that FLT caused further unlinking of the inverted pendulum that occurred as a result of Achilles vibration. Furthermore, it appeared to be unlinked in a manner from where the source of additional somatosensory input originated (fingertip), outward, when Achilles vibration was also present. For example, body segment angular POS values indicated that when FLT was present, the greatest posterior shift brought on by Achilles vibration occurred at the shank ($POS_{\text{Shank}} \sim 2^\circ$), which was also the furthest segment from the fingertip (the source of additional somatosensory input). Thigh (POS_{Thigh}) posterior shifts brought on by Achilles vibration during FLT conditions, were slightly smaller than those seen in the shank, and this segment was also closer to the fingertip. The torso (POS_{Torso}) segment, closest to the fingertip, actually saw a reversal of sway direction, such that Achilles vibration conditions with FLT actually had significantly more anterior positions than the FLT conditions without vibration. POS_{Head} was not affected during the Achilles vibration with FLT though.

Essentially, the largest deviation from the expected effect of Achilles vibration on angular POS during conditions with FLT happened to occur at the torso, which was also the segment most closely linked to the source of additional somatosensory input (the fingertip). The fact that AI_{Torso} showed significantly more stabilization of the torso to external space than to the shank, during FLT conditions with Achilles vibration compared to those without, further supports the notion that the presence of both stimuli caused a breaking of the inverted pendulum, particularly around the segments closest to the FLT.

Overall, we observed less sagittal plane motion in terms of less posterior POS shifts and decreased speeds with FLT added to Achilles vibration compared to conditions with just Achilles vibration, as was predicted with Hypothesis 2.1. These results suggested that the more appropriate afferent inputs from the arm and fingertip maintaining contact with a stable surface were up-weighted while the erroneous proprioceptive stimuli from the calf muscles caused by the Achilles vibration was down-weighted, resulting in improved postural control with FLT. It is important to remember that the force exerted by the fingertip was not enough to provide mechanical stabilization of the body. Experiment #2 additionally showed light touch had the value of improving postural control during stance without Achilles vibration, as several other studies have reported (Baldan et al., 2014; Clapp & Wing, 1999; Dickstein, 2005; Holden et al., 1994; John J. Jeka, 2016). These results and the finding that postural control strategies may have been adjusted even during light touch conditions without Achilles vibration, suggests that inputs from the arm and finger conducting the light touch may have

been up-weighted over other typical lower body somatosensory afferents utilized during quiet stance (such as foot mechanoreceptors, muscle spindles of ankle musculature, etc.) to minimize sway. If weighting of these lower body afferents were maintained as under conditions without FLT, we believe changes to postural control strategy and the inverted pendulum would not have occurred with FLT. Such a notion is supported by the fact that two-point discrimination studies show the fingertip is capable of detecting points as close as two millimeters, while the bottom of the foot can only detect distances around eight millimeters apart (John J. Jeka, 2016). Furthermore, the higher point of contact at the fingertip for this study (refer to Figure 2.3 of the literature review for a depiction), should allow for detection of postural sway earlier than receptors at the feet or ankles could detect (John J. Jeka, 2016). A more precise motion detection capability at the fingertip and the ability for it to detect sway earlier would seemingly both contribute to an up-weighting of inputs originating from the touch arm, corresponding with a down-weighting of lower body afferents, such as foot mechanoreceptors and muscle spindles within ankle musculature, and thus explain our findings of improved postural control and strategy changes observed with FLT.

5.2.3 Arm Vibration × FLT (Experiment #3)

In experiment #3, our hypothesis (3.1) that POS values would be less anterior with just biceps vibration than with FLT (both FLT & FLT + Noise) and biceps vibration conditions both present was generally confirmed (refer to Figures in the Results for the CoP: 4.4.1.1, Head: 4.4.2.1, Torso: 4.4.2.2, Thigh:

4.4.2.3, and Shank: 4.4.2.4). Biceps vibration produced significant right elbow flexion (refer to POS_{Elbow} Figure 4.4.2.5 in the Results), which when combined with FLT conditions, caused POS to move significantly anterior. Likewise, our hypothesis (3.2) that POS values would be less posterior with just triceps vibration than with FLT and triceps vibration conditions both present also seemed to be confirmed by results for the POS_{CoP}, POS_{Head}, POS_{Torso}, and POS_{Thigh}. Triceps vibration produced significant elbow extension, although not to the extent that biceps vibration produced elbow flexion, and when combined with FLT conditions, the triceps vibration caused POS to move significantly posterior. Both elbow extension caused by triceps vibration and elbow flexion caused by biceps vibration were significantly reduced during FLT conditions though, when compared to their No FLT conditions.

Postural speed was not significantly less under conditions with just arm vibration compared to conditions with both arm vibration and FLT as was hypothesized (hypotheses 3.1 and 3.2) though. Instead, during biceps vibration alone, angular speeds were actually significantly higher than during conditions with biceps vibration and FLT for the torso, thigh, shank, and elbow (refer to Figures in the Results for the Torso: 4.4.3.2, Thigh: 4.4.3.3, Shank: 4.4.3.4, and Elbow: 4.4.3.5). SPD likewise tended to decrease with triceps vibration and FLT compared to just triceps vibration alone; however, the majority of metrics did not reach statistical significance. Only the thigh and the elbow saw a significant effect of adding FLT to the triceps vibration conditions that reduced speed.

Our hypotheses (3.1 and 3.2) that variability would be significantly less under conditions with arm vibration alone compared to conditions with arm vibration and FLT were generally not supported. Variability during biceps vibration conditions significantly decreased in the CoP and thigh (refer to RMS Figures in the Results for the CoP: 4.4.1.3 and Thigh: 4.4.4.4) with the addition of FLT compared to the biceps vibration alone. Likewise, variability during triceps vibration conditions significantly decreased in the torso (refer to RMS_{Torso} Figure 4.4.4.3 in the Results) with the addition of FLT compared to the triceps vibration alone. Similar to findings from the first two experiments, no significant comparisons for RMS were yielded at the shank. Thus, it seems changes in angular variability associated with tendon vibration (Achilles, biceps, or triceps) or the presence of FLT occurred only at segments more superior in the body.

Similar to experiment #2, in experiment #3 ApEn was significantly greater with FLT than No FLT, during conditions without tendon vibration. Results for triceps vibration conditions seemed to closely resemble conditions without tendon vibration in that ApEn was also significantly greater with FLT than without. The biceps vibration conditions saw no significant changes in ApEn due to FLT though. These results indicate that there was potentially greater complexity in postural control brought on by FLT, during conditions with biceps vibration and no arm vibration. This notion is further supported by the AI_{Torso} data, which like in experiment #2, indicated significantly less anchoring of the trunk to the thigh brought on by FLT during conditions without tendon vibration. Unlike experiment #2 however, in experiment #3 neither biceps nor triceps vibration conditions saw

significant changes to AI_{Torso} as a result of adding FLT. Thus, arm vibration did not produce less Anchoring of the torso to the thigh as Achilles vibration did in experiment #2.

Overall, arm vibration conditions did cause significant directional postural shifts in equilibrium position when they were combined with FLT, as was hypothesized. These directional shifts under the combined conditions did not occur along with increases in speed and variability though. In fact, the stabilizing effects of FLT reducing SPD and RMS generally remained, even under conditions with arm vibration added to the FLT. These results suggest that an upweighting of arm and fingertip afferents did occur during FLT conditions. When muscle spindles from musculature controlling the elbow joint provided erroneous length input during arm vibration conditions, the upweighted afferents caused shifts in POS values (either posterior for triceps vibration or anterior for biceps vibration). It is noteworthy that although the effect of biceps vibration on the right elbow (greater flexion) was still significant under FLT conditions compared to conditions without arm vibration, the flexion did significantly decrease during biceps vibration with FLT conditions compared to no FLT. Likewise, the effect of triceps vibration on the elbow (greater extension) was significantly decreased with FLT such that triceps vibration conditions were no longer significantly different from conditions without arm vibration during FLT. These deteriorating effects of arm vibration on POS_{Elbow} during FLT indicate that when arm afferents were salient for postural control, the erroneous muscle spindle afferents from arm musculature interacted with more accurate

somatosensory afferents (e.g. cutaneous receptors at the feet or fingertip, muscle spindles within ankle musculature, etc.), and they were likely down-weighted to reduce the impact of their erroneous inputs on postural control. Further supporting the notion of down-weighted arm muscle spindles is the fact that their erroneous inputs did not cause body speed and variability to be increased during FLT conditions with arm vibration. FLT was still able to generally reduce speed and variability, even during arm vibration conditions. Although both forms of arm vibration were disruptive to POS metrics, they were not equally so. The biceps vibration appeared to be a much stronger stimulus in that it induced greater POS changes than triceps vibration. Postural complexity (ApEn) also remained unchanged during FLT combined with biceps vibration conditions, as opposed to the increases observed with FLT during the triceps and no arm vibration conditions, suggesting biceps vibration was a stronger stimulus that may have led to more freezing of degrees of freedom when combined with FLT. This notion is further supported by AI results which revealed biceps vibration conditions were generally the most negative of all conditions during FLT, indicating greater anchoring of superior segments to their inferior segments than to space.

VI. SUMMARY, LIMITATIONS, AND FUTURE DIRECTIONS

6.1 Summary

This study provided a means by which to test if sensory reweighting was capable of occurring within the somatosensory system itself, by combining different types of somatosensory stimuli that stimulate different receptor types at different points in the body and observing their impacts on postural control metrics. Combinations of somatosensory stimuli produced differing postural results than the effects of the individual stimuli. These findings were interpreted to suggest these interactions were evidence of reweighting occurring within the somatosensory system itself.

Experiment #1 tested effects of Achilles vibration and foot noise meant to induce SR on postural control. Achilles vibration proved to be a strong proprioceptive stimulus that moved posture posteriorly and increased speed, as has been established in previous research (Duclos et al., 2014). Foot noise had limited impact on postural control, and it did not significantly mitigate the effects of Achilles vibration. Thus, it is unclear if SR occurred. Any potential effects of foot noise on postural control (SPD_{CoP} and $ApEn$) however, did appear to be somewhat mitigated by Achilles vibration, suggesting that calf muscle spindles afferents may have been up-weighted over the cutaneous mechanoreceptor inputs from the bottom of the feet.

In experiment #2 the effects of Achilles vibration, FLT, and FLT + Noise on postural control were assessed. Like experiment #1, there were not many effects to suggest that the noise added to the fingertip was inducing SR or impacting

balance beyond the effects of FLT itself. Achilles vibration proved to be the same type of strong proprioceptive stimulus that it was in experiment #1 by again increasing speeds and shifting posture in a posterior direction. However, when FLT conditions were added, the effects of vibration were largely reduced. Absent of Achilles vibration, reduced postural speeds and variability seen with FLT compared to without indicated it provided a strong stabilizing effect of reducing postural sway, consistent with previous research (Baldan et al., 2014; John J. Jeka, 2016). The postural improvements noted with FLT suggest that afferents from the arm are up-weighted when their inputs become relevant for postural control through light touch with a stationary surface.

Experiment #3 further tested the effects of light touch (and FLT + Noise) and its interactions with arm tendon vibration effects on postural control. FLT + Noise again did not appear to impact posture much beyond FLT itself. Likewise, arm vibration conditions (both biceps and triceps) did not affect body posture much until combined with FLT. When combined with FLT, biceps vibration induced a more anterior body position. Triceps vibration induced a more posterior body position when combined with FLT, but the effect was not as strong as biceps vibration. Although arm vibration conditions were able to affect body position when combined with FLT, postural speeds and variability were still generally decreased by FLT compared to conditions without FLT. Elbow flexion (biceps vibration) and extension (triceps vibration) that was generated by arm vibration conditions without FLT, was also reduced some during FLT conditions. Thus, the erroneous muscle spindle afferents from the arm musculature

controlling elbow motion were likely down-weighted relative to more accurate somatosensory afferents, such as cutaneous receptors in the fingertip and feet as well as other lower body proprioceptive afferents, during FLT to reduce the elbow motion as well as postural speed and variability which could have occurred from arm vibration.

Novel in these studies were the interactions that tendon vibration and FLT had on postural complexity (ApEn) and anchoring of superior body segments to their inferior ones (AI). Such metrics revealed Achilles vibration caused less anchoring of the thigh to the shank (experiment #1) as well as less anchoring of the torso to the thigh (experiment #2), suggesting some breaking from the body's postural control behaving as an inverted pendulum. FLT seemed to likewise cause some breaking of the inverted pendulum, as the torso became less anchored to the thigh during these conditions. When FLT and Achilles vibration were combined, the unanchoring of the torso to the thigh was further enhanced, suggesting further breaking of the inverted pendulum. Such breaking during FLT conditions may have also led to greater complexity in the postural control system, as assessed by ApEn. However, when Achilles vibration was also present, the increased complexity from FLT was somewhat suppressed. Finally, arm vibration conditions combined with FLT did not cause less anchoring of the torso to the thigh in the way that FLT and Achilles vibration did. Biceps vibration seemed to be a more disruptive stimuli for postural control when combined with FLT than was triceps vibration, as FLT could not increase postural complexity when biceps vibration was also present.

6.2 Limitations

These experiments were not performed without limitations. First, it is important to note that the findings in this study may not be generalizable to the entire population. We specifically recruited young, healthy individuals to assess how reweighting may be occurring within a fully intact somatosensory system. Therefore, it is not suggested that these findings would hold true in individuals experiencing somatosensory declines, such as those seen in the elderly or patients with peripheral neuropathy. However, understanding how reweighting may be occurring within an intact somatosensory system was a reasonable starting point for increasing our understanding of sensory reweighting within a single sensory system. Also, prior to this study it was unknown if noise meant to induce SR in healthy individuals might be more effective if those individuals were also experiencing a disruptive proprioceptive stimulus, such as tendon vibration. Testing solely the young, healthy population allowed us to infer that noise did not appear to generate any greater SR effect when their proprioception was perturbed by tendon vibration than it did without disruption.

Our general lack of SR findings could be considered another limitation. We cannot definitively conclude that SR was not able to impact reweighting within the somatosensory system, because we cannot confirm that our subjects were likely exhibiting SR type behavior for postural control. Our study was not the only one to experience a lack of SR type behavior in young, healthy subjects though (Dettmer, 2014; Hijmans et al., 2008). There are multiple explanations as to why an SR effect may not have occurred. As mentioned, the notion that

subjects' somatosensory systems may have been fully intact and unable to receive much SR type benefit from the noise is one explanation. Another reason could be that the noise given was not optimal for inducing SR in our subjects; however, we did try to utilize noise in a manner that was effective in previous studies (Hijmans et al., 2007; Kimura et al., 2012; Magalhães & Kohn, 2011b). Ultimately, future studies similar to this one could be conducted with samples of subjects that might be more susceptible to exhibiting SR effects, to further assess its ability to impact reweighting within somatosensation.

The metric of ApEn in this study has a potential limitation too. The values for CoP ApEn in our study were somewhat smaller than those calculated in previous studies (Cavanaugh et al., 2006, 2007; Cavanaugh, Guskiewicz, Giuliani, et al., 2005; Dettmer et al., 2015), because we did not utilize a lag of ten as they did. In fact, calculation of ApEn can vary considerably depending on the parameters set (embedding dimension, tolerance, and lag). Future studies however, may consider utilizing other nonlinear means such as state space, phase, or control analysis in attempts to better assess complexity of the postural control system and predict behavior.

Finally, there were some issues with normality and homogeneity of variance within our data which generally is of concern when utilizing parametric statistical analysis as we did. These issues may have arisen due to the relatively small sample size that was collected. Despite potentially violating these assumptions for parametric tests, as we explained in section 3.6, utilizing nonparametric Kruskal-Wallis tests for statistical analysis did not negate our

significant findings. Furthermore, the nonparametric Kruskal-Wallis tests appeared to not be stringent enough in yielding some significant comparisons between overlapping means. Thus, we believe the more conservative parametric tests utilized for our analyses were more appropriate in order to reduce the risk of type I error.

6.3 Future Directions

Future research might be able to further distinguish how somatosensory afferents are weighted when receiving conflicting information. For example, a nerve block on the lower body somatosensory afferents could be utilized to see if arm vibration continues to reveal reduced effects at the elbow and if sway variability and speeds are still decreased, under conditions when both arm vibration and FLT are present. If these results persisted with the nerve block, then it could be inferred that reductions in body sway variability and speed, as well as elbow motion, were likely the results of upper body somatosensory afferents, such as from the fingertip, providing accurate information to counteract the erroneous muscle spindle afferents from arm vibration. Such a finding would also support the notion that upper body afferents may be up-weighted over lower body afferents when FLT provides salient information about postural (John J. Jeka, 2016).

Another direction future research might consider would be to determine how competing afferents from within other sensory modalities impact postural control. For example, how might visual information affect balance if one eye were receiving competing visual flow information from the other, or what if speed

of visual flow within the central vision did not match the speed of visual flow within the peripheral vision? Studying these examples might be able to help distinguish how visual afferents are weighted within that modality.

Overall though, more research into reweighting should be done that engages the mindset that weighting of sensory afferents does not have to occur holistically between the three sensory modalities. If sensory receptors within a single modality are receiving competing information relevant to postural control, as this study demonstrated, it can be inferred that reweighting likely still occurs within that modality. Perhaps it is not so much the modality where erroneous inputs come from that affects reweighting, but rather it may be the relevance that specific inputs provide about postural control that determines weight.

VII. REFERENCES

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APPENDIX A: RECRUITMENT FLYER

UNIVERSITY of **HOUSTON**

HEALTH & HUMAN PERFORMANCE

CNBR CENTER for NEUROMOTOR and
BIOMECHANICS RESEARCH

HOW IS SENSORY INFORMATION USED FOR BALANCE?

Healthy adults are needed for a study investigating sensory contributions to postural control.

Participate in a study that will improve our understanding of how we balance, and help those at risk of balance deficits, such as astronauts and the elderly.

- You should be healthy and between 18-35 years old.
- You should be able to stand upright for prolonged periods of time (up to 30 minutes).

The test session will be performed in the Health and Biomedical Sciences Building 2 (HBS2) at the UH main campus.

The test session can be completed in approximately 2.5 hours.

For more information on participating in this study,
please contact David Temple at:
(713) 743-7677 or drtemple@uh.edu.

This project has been reviewed by the University of Houston IRB Committee 1,
responsible for the protection of human subjects (713) 743-9204.

APPENDIX B: MODIFIED PAR-Q

Physical Activity Readiness Questionnaire (PAR-Q)

This PAR-Q has been designed to identify adults who are eligible to be subjects in this study. Please read the following questions carefully and mark yes or no as it applies to you. If you should have any concerns or questions regarding any of the items, please ask the test administrator.

Yes <input type="checkbox"/>	No <input type="checkbox"/>	1) Has a physician ever said you have a condition where you should only do physical activities recommended by a physician?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	2) Has a physician ever said you have a condition where you should not perform certain types of activities requiring balance (e.g. standing for prolonged periods of time with your eyes closed, etc.)?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	3) Do you have difficulty standing for prolonged periods of time (e.g. ~30 minutes without a break)?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	4) Has a physician ever indicated that you may have a neurological issue that could affect your balance?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	5) Has a physician ever indicated that you may have a musculoskeletal issue that could affect your balance or cause pain when standing for prolonged periods of time?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	6) Do you have insulin dependent diabetes, diabetic neuropathy, or any related conditions?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	7) Do you have any eye or vision problems? If so, what? _____
		For vision acuity issues, is it corrected with contacts or glasses well enough to not impede balance and/or locomotion? _____
Yes <input type="checkbox"/>	No <input type="checkbox"/>	8) Have you had any surgery on the ears, head, or neck regions that would impact your current balance and/or locomotion capabilities?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	9) Have you had any other medical problems or surgeries that may impact your vision, vestibular, or somatosensory systems?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	10) Do you have any ataxia (loss of control of body movements) or known gait problems?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	11) Do you ever lose consciousness, or do you lose your balance because of dizziness?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	12) Are you pregnant?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	13) Do you suffer from Epilepsy or ever experience seizures?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	14) Do you have any known musculoskeletal or neurological injuries or deficits that may affect your sensory system, posture, or locomotion?
Yes <input type="checkbox"/>	No <input type="checkbox"/>	15) Do you know of any other reason why you should not participate in any of the activities involved in this study (e.g. standing for prolonged periods of time with your eyes closed while experiencing tendon vibration)?