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THE EFFECTS OF AGING ON MULTIPLE POSTURAL MUSCLE CONTROL AND POSTURAL SWAY BEHAVIOR

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This work is dedicated to all professionals passionate in

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improving well-being of individuals and societies.

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The Effects of Aging on Multiple Postural Muscle Control and Postural Sway Behavior

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Abstract

Episodes of instability and falls in the elderly represent a major public health concern. The lack of scientific information about the effects of age-related changes on neurophysiological mechanisms of postural control has limited the advance in the field of fall prevention and rehabilitation of balance disorders. The overall goal of this dissertation was to investigate the effects of aging on postural control. Considering the progressive non-homogeneous deterioration of aging physiological systems, a series of five experimental studies, with healthy young and healthy nonfaller older adults performing upright stance tasks, explored three main hypotheses: (1) intermuscular coherence analysis is able to detect signs of intermuscular synchronization at lower frequency bands as one of the strategies used by the Central Nervous System to control upright stance; (2) aging is associated with a reorganization of correlated neural inputs controlling postural muscles; and (3) aging is associated with changes in body sway behavior. The first three studies corroborated the use of intermuscular coherence analysis to investigate the formation of correlated neural inputs forming postural muscle synergies during upright stance. The fourth study revealed an age-related reorganization of the distribution and strength of correlated neural inputs to multiple postural muscles. Healthy nonfaller older adults presented stronger levels of synchronization, within 0–10 Hz, for three distinct muscle groups: anterior, posterior, and antagonist muscle groups. The fifth study investigated age-related changes on postural sway using traditional and novel postural indices extracted from the center of pressure coordinates. Although the functional base support is preserved in healthy nonfaller older adults, these seniors revealed a larger, faster, shakier, and more irregular pattern of body sway compared to healthy young adults. In addition, age-related changes on supraspinal mechanisms, spinal reflexes, and intrinsic mechanical properties of muscles and joints involved in postural control were observed by changes in both rambling and trembling components of the postural sway. Findings reported here provide valuable information regarding compensatory mechanisms adopted by healthy nonfaller older adults to control upright stance. Together, these findings suggest an age-related reorganization of correlated neural inputs controlling multiple postural muscles, accompanied by changes in body sway behavior.

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LIST OF KEY ABBREVIATIONS

APAs: Anticipatory Postural Adjustments

BCE: Bipedal Stance with Closed Eyes

BF: Biceps Femoris

BOE: Bipedal Stance with Open Eyes

CNS: Central Nervous System

COM: Center of Mass

COP: Center of Pressure

CPAs: Compensatory Postural Adjustments

EMG: Electromyographic Signal

ERE: Lumbar Erector Spinae

M-mode: Muscle Mode or Functional Muscle Group or Synergistic Muscle Group

PSD: Power Spectrum Density

RA: Rectus Abdominis

RF: Rectus Femoris

RMS: Root Mean Square

SOL: Soleus

ST: Semitendinosus

TA: Tibialis Anterior

UCM: Uncontrolled Manifold Analysis Method

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not presented here

significant *Age* effect ($p < 0.02$).

CHAPTER 1
INTRODUCTION

1.1. PROBLEM STATEMENT AND RESEARCH FOCUS

The overall focus of the dissertation was the effect of natural aging on human postural control of upright stance. Emphasis was given to the study of body sway behavior and correlated neural inputs sent from the Central Nervous System (CNS) to multiple postural muscles to control upright stance in aged individuals. This focus emerged from the importance of postural control in everyday life, the possible effects of aging on neuromuscular control, and the high incidence of falls in the elderly.

Postural control is an imperative ability that plays a role in most daily activities. It is intrinsically related to an individual's independence, self-care, social integration, and quality of life. The ability to maintain a certain body posture in space is far from simple as it depends on a complex system performing sensorimotor integration based on neural signals. As the individual grows older, a series of physiological changes impact human postural control to the point that postural instability and balance disorders in the aging population are considered a major health concern. Currently, falls are the leading cause of fatal and non-fatal injuries in older adults, affecting approximately one third of adults over 65 years of age (Hausdorff et al 2001, Sleet et al 2008, CDC 2014).

Specialists in areas of physical rehabilitation sciences recognize that, in order to advance the development of efficient models of prevention and intervention aiming to reduce episodes of instability and falls, a better comprehensive understanding of the neural mechanisms of postural control and the effects of aging on these mechanisms is needed. Therefore, the **general goal** of the dissertation was to identify the effects of the natural process of aging on organizing multiple postural muscles acting on several joints to control upright stance. The **overall rationale** behind the experimental studies reported here relates to the effects of the non-homogeneous functional declines observed across physiological systems on the resulting ability to control posture. It was **hypothesized** that such age-related changes may force the CNS to adapt its strategy to control multiple

muscles, resulting in further changes of body sway patterns. To test this central hypothesis and address the gap of knowledge regarding postural adaptations caused by the aging process, a series of studies was performed exploring the following research questions and hypotheses:

- **Question 1:** How does the CNS organize multiple muscles to control upright stance?

  Hypothesis 1: Correlated neural inputs sent to a group of postural muscles forming muscle synergies may be one of the strategies used by the CNS to control upright stance. If this is the case, *Intermuscular Coherence Analysis* would be able to detect signs of intermuscular synchronization at lower frequency bands among postural muscles.

- **Question 2:** What are the effects of the natural process of aging on the formation of correlated neural inputs to postural muscles in order to control upright stance?

  Hypothesis 2: The non-homogeneous age-related decline across physiological systems may lead to a reorganization of the generation and distribution of correlated neural inputs to control multiple postural muscles during upright stance.

- **Question 3:** What are the effects of aging on postural sway behavior during upright stance?

  Hypothesis 3: Age-related changes in multiple muscle control may be detected by changes in postural sway behavior. Healthy nonfaller older adults may sway more, faster, less smoothly and with a more irregular pattern during upright stance than healthy young adults.

The findings from this series of studies are expected to impact the scientific and clinical communities in the following ways. First, new knowledge will be gained on the mechanisms underlying postural sway behavior and multi-muscle control in older adults. Second, the findings produced here will aid advancements in the field of the assessment of balance control in older adults. It is important to emphasize that some of the studies

included in the dissertation were dedicated to the implementation of a variety of balance methods representing several dimensions of body sway characteristics that can be quantified via simple instrumentation. Finally, this work will assist future clinical and basic research work focusing on the development of more efficient interventions to optimize balance control in the elderly.

1.2. ORGANIZATION OF THE DISSERTATION

The dissertation is organized in eight (8) chapters.

Chapter 1 is an introduction to the dissertation, presenting the overall research focus.

Chapter 2 reviews the general literature related to human postural control and current literature on age-related changes to the postural control.

Chapters 3 to *7* present five experimental studies performed during PhD training in a periodic article format. The first three studies (*Chapters 3, 4*, and *5)* focused on the novel approach, the *Intermuscular Coherence Analysis*, to investigate possible mechanisms of multi-muscle control during upright stance.

Chapter 3 is based on the published article Danna-dos-Santos A, Boonstra TW, Degani AM, Cardoso VS, Magalhães AT, Mochizuki L, Leonard CT (2014) Multi-muscle control during bipedal stance: an EMG-EMG analysis approach. *Exp Brain Res,* 232(1): 75–87. Specifically, *Chapter 3* investigated the hypothesis that correlated neural inputs, as measured by intermuscular coherence, may be one of the mechanisms used by the CNS to coordinate the formation of postural muscle synergies. This hypothesis was investigated by analyzing the strength and distribution of correlated neural inputs to posterior postural muscles during upright stance. Nine healthy young participants (4 females and 5 males, mean age $= 29.2$ years old, $SD = 6.1$) performed the task of standing while holding a 5 kg barbell in front of their bodies for 10 s. The activity of three postural muscles was recorded by surface electrodes: soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE). Intermuscular coherence was estimated for three muscle pairs (SOL/BF, SOL/ERE, and BF/ERE). These three muscles were selected based on

previous reports describing them as components of a functional (synergistic) muscle group during bipedal stance. The experimental condition elicited significant pooled coherence for all three muscles within two distinct frequency bands: 0–5 and 5–20 Hz. The former frequency band showed stronger synchronizations for the more distal muscle pair (SOL/BF). The latter frequency band showed similar strength of its synchronization among all three postural muscles recorded and has been interpreted as a sign of a correlated circuitry underlying multi-muscle control. These findings corroborate the hypothesis of a synchronization of neural oscillations to multiple postural muscles as a strategy to reduce the number of variables to be controlled by the system.

Chapter 4 expands the use of *Intermuscular Coherence Analysis* as a means to detect signs of intermuscular synchronization at lower frequency bands by including a larger number of postural muscles as well as a different challenging task (upright stance with closed eyes). This chapter is based on the published article Danna-dos-Santos A, Degani AM, Boonstra TW, Mochizuki L, Harney AM, Schmeckpeper MM, Tabor LC, Leonard CT (2015) The influence of visual information on multi-muscle control during quiet stance: a spectral analysis approach. *Exp Brain Res,* 233: 657-669. The strength and distribution of correlated neural inputs were computed for six postural muscles (soleus, biceps femoris, lumbar erector spinae, tibialis anterior, rectus femoris, and rectus abdominis). These muscles were selected because they are the main skeletal muscles involved in the maintenance of the body's vertical positioning. Ten healthy young adults (4 females and 6 males, mean age = 26.8 years old, *SD* = 2.7) performed two experimental tasks: upright bipedal stance with opened eyes and closed eyes, for 30 s each. Intermuscular coherence was estimated using EMG signals recorded from the six muscles selected. Muscle pairs included pairs formed by solely anterior muscles, solely posterior muscles, antagonist muscles, or mixed muscles (one posterior and one anterior, non-antagonist, muscles). The synchronization patterns observed between muscle pairs were found to be concentrated within a frequency interval of 1–10 Hz when visual information was available. No significant intermuscular coherence was found for mixed muscle pairs. In addition, the lack of visual input during unperturbed stance not only decreased the correlation between muscle pairs in the frequency domain, but also the coherence was no longer significant. These findings corroborate the use of the

intermuscular coherence approach to investigate the formation of muscle synergies during human upright stance.

Chapter 5 advances the use of intermuscular coherence analysis as a means to investigate the formation of muscle synergies in older adults. In general, the presence, distribution, and strength of correlated neural inputs to the same three postural muscles studied in *Chapter 3* (SOL, BF, and ERE) were explored in thirteen healthy nonfaller older adults (8 females and 5 males, mean age = 69.0 years old, *SD* = 3.4). All participants performed upright stance holding a 5 kg barbell in front of their bodies. Intermuscular coherence analysis revealed the presence of correlated neural inputs to all three posterior postural muscles within the frequency band of 0–10 Hz, as happened for young adults. However, older adults showed significant synchronization not only for the most distal muscle pair (SOL/BF), as was observed for young adults. Older adults also revealed significant synchronization for the other two muscle pairs (SOL/ERE and BF/ERE). In addition, intermuscular coherence estimates within the frequency band of 0-10 Hz for the muscle pairs SOL/ERE and BF/ERE were significantly stronger in older adults, compared to that in young adults. These findings corroborate the use of Intermuscular Coherence Analysis as a sensitive method to detect the effects of aging on the organization and strength of neural drive to postural muscles.

Chapters 6 investigates the effects of aging on the formation, distribution, and strength of correlated neural inputs forming muscle synergies during upright stance. Ten healthy young adults (mean age = 26.8 years old, *SD* = 2.7) and ten healthy nonfaller older adults (mean age = 68.7 years old, *SD* = 3.5) performed two tasks: unperturbed bipedal stance with open eyes and unperturbed bipedal stance with closed eyes. The EMG activity of six postural muscles was recorded and intermuscular coherence analyses were performed. Older adults presented significant levels of intermuscular coherence within the frequency band of 1–10 Hz, indicating the presence of correlated neural inputs to aging skeletal muscles responsible for controlling upright stance. Signs of synchronization were reported for three distinct muscle groups: "push-forward M-mode", "push-back M-mode", and antagonist group. No significant intermuscular coherence was found for muscle pairs formed by one anterior and one posterior non-antagonist muscles, as also happened for young adults. Interestingly, coherence estimates within 1 to 10 Hz

were stronger in older adults, compared to that in young adults. Taken together, the use of intermuscular coherence analysis provided new knowledge on the mechanisms underlying the strategies adopted by the aging CNS to control multiple postural muscles. The findings suggest a reorganization of correlated neural inputs forming synergistic muscle groups responsible to control upright stance and avoid falls.

Considering that age-related changes on multiple postural muscles control may be reflected in changes in postural sway behavior, *Chapter 7* investigates the effects of aging on postural sway using postural indices from multiple domains (spatio-temporal, frequency, and structural domains). Eleven healthy young volunteers (*Control* group, mean age = 27.1 years old, *SD* = 3.8) and fourteen healthy nonfaller older volunteers (*Senior* group, mean age = 68.8 years old, *SD* = 3.2) performed three upright standing tasks on a force platform: bipedal stance with opened eyes, bipedal stance with closed eyes, and body oscillation to the limits of stability. Postural indices from multiple domains were extracted from the center of pressure (*COP*) coordinates recorded by a force platform. The results showed that the functional base support seems to still be preserved in healthy nonfaller older adults. However, these individuals tended to oscillate more and faster in both anterior-posterior and medio-lateral directions; increase their medio-lateral sway frequency; present a more irregular and random body sway pattern in both directions; and modify both rambling and trembling components of their postural sway, compared to healthy young adults. When older adults were asked to close their eyes, they increased their body sway velocity, frequency, and irregularity in the anterior-posterior direction. The effects of temporary visual removal on postural sway in older adults were also observed for the rambling component of the COP displacement in the anterior-posterior direction. In conclusion, the inclusion of postural indices from multiple domains allowed the detection of subtle changes in postural control in the first stage of aging (65 to 74 years old). Therefore, the use of multiple variables provides a more comprehensive understanding of the neural mechanisms underlying the effects of aging on postural control. These findings are crucial to direct efforts of health professionals to optimize treatment and rehabilitation of age-related postural instability.

Chapter 8 presents a summary of conclusions, contributions to science, and general clinical relevance of the dissertation.

CHAPTER 2
BACKGROUND AND LITERATURE REVIEW

Due to the consequences resulting from a fall, postural instability in the aging population is considered a major public health concern. Falls affect approximately one third of people over the age of 65 annually and are generally associated with disability, decreased quality of life, institutionalization, and high morbidity and mortality rates (Hausdorff et al 2001, Stevens et al 2006, Sleet et al 2008, Boyé et al 2014). In 2012, 2.4 million nonfatal falls among older adults were treated in emergency care; however, this number is considered to underrepresent the total number of fall occurrences in United States (Stevens et al 2012). In addition, the elderly population has grown dramatically in the last century (Paige 1992). It is expected that the number of people 60 years and older across the world will reach 2 billion by 2050 (World Health Organization 2002).

Despite current efforts to optimize programs and interventions aiming to reduce fall risk in the elderly, further understanding of the age-related changes to the neural mechanisms underlying human postural control is needed. This chapter presents an overview on human postural control followed by the current literature of the effects of the natural process of aging on upright postural control.

Within the dissertation, terms such as body posture, postural sway, unperturbed stance, postural control, postural perturbation, postural task, center of pressure, synergy, and correlated neural inputs will appear several times. In order to avoid possible confusions, some definitions are presented here. The term "*body posture*" will be considered as the configuration of the body in space. It is important to note that this configuration may or may not change over time. The term "*postural sway*" will be used to denote small variations in body position when individuals are asked to maintain a certain posture (e.g., upright stance). The term "*unperturbed stance*" will be considered as the maintenance of the upright posture in the absence of any other motor task or perturbation. The term "*postural control*" will be referred to as the functional ability to control the center of pressure (COP) sway by maintaining the projection of the center of mass (COM) within the manageable limits of the base of support either to stay upright or

to recover equilibrium after perturbations or challenging postures. The term "*postural task*" will be referred to as the ability to keep the body in a certain configuration. The term "*postural perturbation*" will be considered as any external force (force generated by the environment and applied to any part of the body) or internal force (mechanical force generated by contraction of muscles or movement of one segment that propagates to adjacent segments) that can induce changes in the current body configuration. The term "*Center of Pressure*" (COP) will be referred to as the point of application of the ground reaction force to the body.

Considering the broad use of the term "*synergy*" in motor control literature, "*synergy*" in the dissertation will be referred to as the task-specific organization of elemental variables stabilizing a particular performance variable and reducing their variability across repetitive trials (Latash, Scholz & Schöner, 2007, Latash 2008). In line with this definition, Gelfand and Tsetlin (1966) defined "*functional synergies*" as a fixed and reproducible interaction of the joints or groups of joints, organized and controlled by the Central Nervous System (CNS) for effective solution of a specific problem. These definitions are different from those often used in clinical practice, in which "*synergy*" refers to the set of muscles (or muscle groups) performing essentially the same action, such as the "*synergistic*" muscles biceps femoris, semitendinosus, and semimembranosus related to the knee flexion. This definition was first presented at the beginning of the last century by Sir Charles Sherrington (Sherrington 1910, Burke 2007) and it is purely based on the anatomical function of the muscles.

The term "*correlated neural inputs*" (or *common neural inputs*) refers to a pattern of distribution of neural inputs coming from a common commanding element or neural network, which is formed by the synchronization at low frequency bands of different areas within the CNS . Such correlated neural inputs diverge onto alpha-motoneuron pools, causing rhythmic discharge among specific muscles. Therefore, skeletal muscles receiving correlated neural inputs will present signs of intermuscular synchronization at lower frequency bands.

Other important terms used in the dissertation will be defined throughout the chapters.

2.1. HUMAN POSTURAL CONTROL

Along the life span, one becomes very familiar with the ability to control the body to execute a great variety of daily motor tasks. For example, reaching to shake someone else's hand demands the control of the position of several body segments in time. This control includes not only the reaching movement of the upper extremity, but also the muscle activation necessary to control the vertical position of the axial skeleton and to counter-act the effects of gravity and internal body forces. Simple and routine postural tasks are generated smoothly and without a great deal of thought in healthy individuals. One only starts realizing how difficult and complicated it is to control posture when facing neuromotor deficits, such as those resulting from neurological diseases (e.g., Parkinsonism, stroke, and peripheral neuropathies), trauma, or aging.

2.1.1. A Brief Historical Perspective on the Study of Postural Control

Systematic investigation of mechanisms involved in human postural control began with experimentation on four-legged animals by Sir Charles Sherrington (Burke 2007). In the beginning of the 20th century, Sherrington used cat specimens to study reflexes mediated by the spinal cord and midbrain. He used a technique, now known as "*decerebration*", to isolate the midbrain and spinal cord from higher areas within the Central Nervous System (e.g., cortical areas). Sherrington reported an increased activation of extensor muscles in all four legs and other anti-gravity muscles when decerebration was performed between the superior and inferior colliculi. This pattern of increased muscle activation was termed "*rigidity*", and it sometimes allowed the animal to stand unsupported. The opposite result was reported when decerebration was accompanied by "*deafferentation*", which is the disruption of the afferent information from sensory nerves. The resulting absence of muscle activity suggested the importance of afferent signals to postural control. Sherrington's findings suggested that the rigidity of extensor muscles represented the foundation of posture control. Subsequent studies corroborated Sherrington's observations. For example, Prof. Rudolf Magnus (Magnus

1926a,b) uncovered series of hierarchically organized reflexes in cats, varying from simple to more complex motor outputs depending on the site where decerebration was performed. The closer the decerebration was to the higher brain centers, the more complex the resultant motor actions were. Magnus (1926a,b) suggested that postural control could be achieved by a summation of reflexes. Further studies provided strong evidence that, despite the role of reflexes and reflex-like reactions in the maintenance of postural equilibrium, reflexes do not fully account for all postural control (Belen'kiy et al 1967, Cordo and Nashner 1982, Aruin and Latash 1996, Shiratori and Latash 2000).

In 1889, Hughling Jackson introduced the idea that muscles are controlled in groups and not independently by stating that "*the central nervous system knows nothing about muscles, it knows only movements*" (Hughling Jackson 1889). Later, Nicolai Bernstein (Bernstein 1947,1967) advanced studies in human movement coordination and postural control by questioning how the brain chooses a solution from the vast number of seemingly equal options, emphasizing the redundancy of the human system. Bernstein's idea of several degrees of freedom for the brain to select was based on his observations on the work of blacksmiths. He observed that the task of hammering presented higher variability of shoulder, elbow, and wrist configurations than those needed for the tip of the hammer to hit the chisel. He suggested that motor variability allowed the joints to place the hammer in a similar final position by using different joint configurations. Bernstein's idea of redundant degrees of freedom as a central issue in motor control has been termed the Bernstein's Problem or Problem of Motor Redundancy (Turvey 1990). Following the idea of different limb configurations (degrees of freedom) achieving a specific motor task, Bernstein suggested that the control of human movements was organized hierarchically in at least four different levels: muscle tone, muscular and articular links, space, and motor actions. Regarding muscle coordination, Bernstein suggested that the complexity of controlling human posture could be partly solved by the formation of "*postural synergies*" (Bernstein 1967). He described these synergies as the combination of signals sent to postural muscles aimed at decreasing the number of variables that the CNS needs to control body equilibrium during either unperturbed posture, in anticipation of a voluntary movement, or in response to an external perturbation.

For decades, postural control has been studied using different perspectives and approaches. Biomechanical approaches have been used to investigate the mechanical complexity of the human body. Behavioral and neurophysiological studies have been performed to understand the integration of sensory information with motor responses. Some researchers have dedicated their efforts to proposing reliable methods to quantify and test Bernstein's hypothesis of the CNS organizing muscle synergies to decrease the number of variables to be manipulated (Gelfand and Latash 1998, Scholz et al 2000, Latash et al 2005,2007, Latash 2008, Danna-dos-Santos et al 2008).

2.1.2. The Mechanical Challenge of the Human Upright Posture

Human bipedal upright posture, defined as the configuration of the body segments relative to the gravitational force and the relative position among body segments at any given time (Zatsiorsky 1998), is inherently unstable. The mechanical challenge of the human upright posture is also related to the fact that the human body is not a rigid body. The human body has two-thirds of its mass located two-thirds of the body height above the ground. In addition, the human body consists of multiple linked segments (kinematic chain) connected to each other by muscles, tendons, ligaments, skin, and other soft tissues. A kinematic chain is said to be in mechanical equilibrium when all the links of the chain are in equilibrium (Zatsiorsky 2002). This mechanical equilibrium can be characterized according to its stability: if a mechanical system returns to equilibrium after being subjected to small disturbances, it is said to be in a stable equilibrium; whereas if the system departs from an equilibrium state after the application of a mechanical disturbance, it is said to be in an unstable equilibrium. To maintain the human body in a stable equilibrium while one or more body segments move, forces and torques created by the movement itself (internal forces) must be synchronously counteracted. It is important to note the complexity of the muscular system. For example, the action of a given muscle can stabilize one joint and destabilize another, complicating the control of multiple muscles acting to control movement and posture.

Based on this brief introduction, a complexity of mechanisms is responsible for controlling the upright position of the human body. The maintenance of the body orientation and postural equilibrium requires continual integration between perception and action to create corrective torques. Principles of sensorimotor integration are discussed in the next section.

2.1.3. The Central Nervous System (CNS) and Postural Control

Human bipedal postural control relies on a continuous and complex integration of sensory and neuromuscular systems. *Figure 2.1* illustrates the systems involved in balance control.
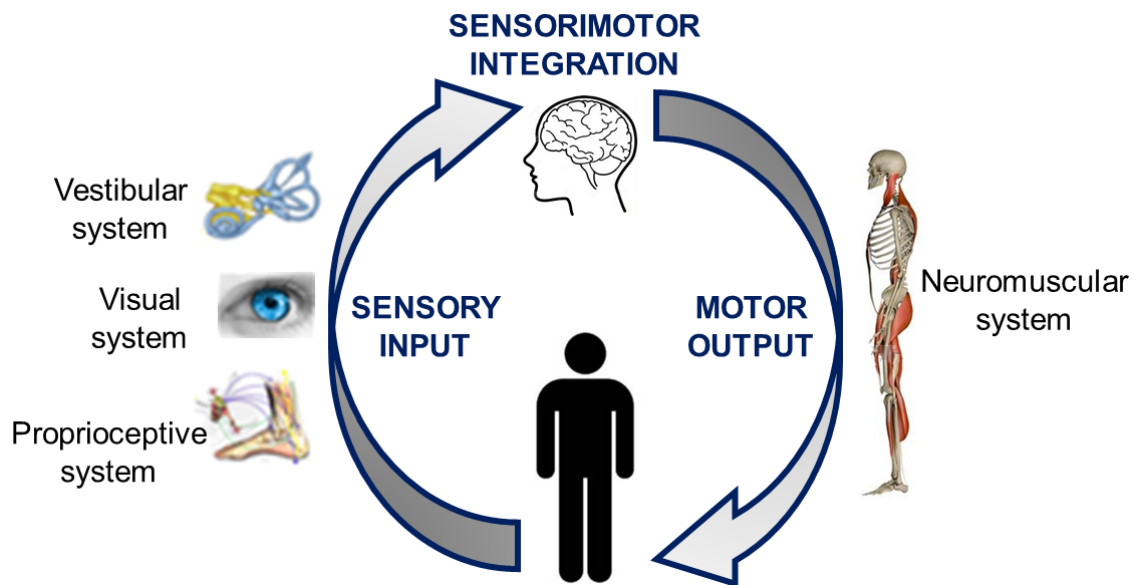
Figure 2.1. Contribution of different systems to balance control.

The variety of sensory inputs from different parts of the body provides a reliable time-to-time neural representation of the body configuration and its relation with the surrounding environment (Winter 1995, Horak and McPherson 1996, Konrad et al 1999, Peterka 2002, Mergner et al 2003). This representation is essential for corrective

adjustments of body position. The primary sensory inputs (or afferent inputs) come from the visual, vestibular, and proprioceptive systems. The roles of visual, vestibular, and proprioceptive information regarding human postural control have been extensively studied (Roll et al 1980, Dijkstra et al 1994a, Fitzpatrick and McCloskey 1994, Kuo et al 1998).

The **visual system** uses oculomotor movements, such as smooth pursuit and saccadic movements, together with peripheral visual perception to provide a reference for verticality, head/body orientation, and body velocity relative to the visual world. Several studies have shown changes in the features of postural sway when the eyes are closed (Allum and Pfaltz 1985, Fitzpatrick et al 1992a, Simoneau et al 1992, Schumann et al 1995). Other than the temporary removal of visual input, the manipulation of the visual environment has also been shown to impact postural control. Under the paradigm of the "*moving room*", the visual environment is manipulated by either actually moving the room or altering the display in front of the individual to mimic motion of the visual environment (Lee et al 1980, Schöner 1991, Dijkstra et al 1994a,b). For example, when the visual environment accelerates towards the individual, he or she perceives this as a forward sway of their body and reacts by an actual backward sway of the body.

The **vestibular system** employs the use of six semicircular canals to detect angular head movements, and two utricles and saccules to detect both linear head movements and the acceleration of gravity. Studies of the role of vestibular information for human postural control commonly use galvanic stimulation applied at the vestibular apparatus behind the ear. Such stimulation changes the firing rate of peripheral vestibular afferent information. Depending on the position of the individual's head and the polarity of the current, the body leans in a particular direction (Hlavacka et al 1995,1996, Coats and Stoltz 1969). For example, when an individual faces forward and a positive current is applied to the right vestibular organ, a body sway to the right is observed. When a similar current is applied but the head is initially positioned in rotation to the right, the whole body moves backward (Hlavacka and Njiokikjien 1985). An increase in the amplitude of the vestibular stimulation leads to an approximately linear increase of the body sway (Coats and Stoltz 1969). Sinusoidal stimulation results in body sway towards the positive

stimulus and away from the negative one, which leads to a sinusoidal sway pattern at low frequencies (Petersen et al 1995).

The **proprioceptive system** detects the orientation and velocity of body segments in space and relative to each other, the muscle length, and any contact with objects and surfaces via sensory receptors embedded in the muscles, tendons, joint capsules, and skin. Some examples of sensory receptors are the muscle spindles, golgi tendon receptors, and skin receptors. The muscle spindles are located among striated fibers of skeletal muscles. Muscle-tendon vibration has been used as a powerful stimulus for muscle spindles (*Ia* afferents) and a way to better understand their role in controlling upright stance. Lackner and Levine (1979) reported a linear correspondence between the muscle spindle discharge and the stimulus at frequencies below 100 Hz. Muscle vibration generates a tonic contraction (also called the tonic vibration reflex) a few seconds after the beginning of the vibration. This contraction increases gradually and, then, stays at a relatively constant level until few seconds after the stimulus is turned off. The Golgi Tendon Organ (GTO), a stretch receptor sensitive to tendon force, is located close to the junction between the tendon and extrafusal muscle fibers. GTOs generate spinal reflexes and supraspinal responses, which helps controlling muscle contraction (Stephens et al 1975, Moore 1984, Latash and Zatsiorsky 2016). The skin receptors represent another important source of sensory information. Studies have shown that a light touch by a fingertip at mechanically non-supportive force levels ($< 1N$) greatly attenuates postural sway during unperturbed stance (Holden et al 1994, Jeka and Lackner 1994). It was suggested that the higher receptor density present in the index finger was able to detect subtle changes in force level and direction. This sensory information contributed to the decrease in body sway (Holden et al 1994, Jeka and Lackner 1994). The effects of touch were also observed in blind individuals and those with vestibular loss (Jeka et al 1996, Lackner et al 1999).

Interestingly, the interaction between vision and touch was investigated by Jeka et al (2000). They manipulated both the visual field and finger touch. Their results suggested that the sensory integration of visual and cutaneous information behaved in an approximately linear fashion. In another study, the relation between cutaneous and muscle spindle information was investigated by asking participants to stand on Romberg

position (feet touching each other in parallel arrangement) with and without a light finger touch to a stable surface, while their fibularis longus and brevis tendons were vibrated (Lackner et al 2000). The results indicated that the finger touch was sufficient to suppress the destabilizing effects of the vibration.

Based on all information coming from the aforementioned sensory systems, the CNS integrates redundant but divergent inputs simultaneously to influence and alter sensory reweighting. During the dynamic process, the system scales the relative importance of each sensory input on postural control. Subsequently, the CNS sends motor outputs to postural muscles spanning across several joints of the axial skeleton (Horak and McPherson 1996, Jeka et al 2000, Teasdale and Simoneau 2001).

Considering the importance of sensorimotor integration, changes in sensory input or motor output, such as those associated with aging, are detrimental to postural control. Numerous studies over the years have examined which brain structures process information related to the control of human posture. For example, it is suggested that not only a few but a great number of supraspinal pathways are fundamentally involved in the processes of posture control; to name a few, the cerebellum, basal ganglia, thalamus, brainstem, and cortex of the hemispheres (Diener et al 1984, Bazalgette et al 1986, Viallet et al 1987, Bouisset and Zattara 1990, Horak and Diener 1994, Karnath 2000a,b, Slobounov 2006a). The supraspinal contributions to the control of human posture have been investigated in behavioral studies and clinical observations relating changes in body behavior to specific injured areas and areas usually targeted by pathologies (Diener et al 1984, Bazalgette et al 1986, Viallet et al 1987, Bouisset and Zattara 1990, Horak and Diener 1994, Nakamura et al 1997, Slobounov et al 2006b,2008).

Individuals with cerebellar disorders typically show increased anterior-posterior postural sway during unperturbed stance, and exaggerated postural responses (such as increased activation of muscles acting at the ankle joints) to unexpected perturbations (Diener et al 1984, Horak and Diener 1994). Neurophysiological abnormalities of the basal ganglia also affect postural behavior (Bazalgette et al 1986, Viallet et al 1987, Bouisset and Zattara 1990). Patients with Parkinson's disease commonly present smaller amplitude of anticipatory postural adjustments (APAs) in postural muscles, anticipatory cocontraction of antagonist muscles, and impaired ability to voluntarily modulate

preprogrammed postural reactions during postural perturbations (Bazalgette et al 1986, Viallet et al 1987, Bouisset and Zattara 1990). Postural control after a stroke may also result in a series of impairments since different areas of the brain may be affected. Although the consequences of cortical stroke vary among individuals, a general clinical finding is hemiparesis (i.e., the decrease of muscle strength and impairment on the ability to move and control joints in certain segments on the side of the body contra-lateral to the side of the brain lesion). These individuals may have a relatively good trunk balance soon after the stroke, but they may lose lateral balance and may fall toward the affected side even when sitting (Brunnstom 1970, Davies 1985, Bohannon 1986, Pedersen et al 1996, Karnath 2000a,b). Another growing body of experimental studies demonstrating the role of supraspinal structures involved in postural control comes from a series of behavioral studies involving individuals with history of mild traumatic brain injury (mTBI). These studies suggest that, even in mild cases with no detected anatomical tissue lesions, the trauma affects the intricate network of mechanisms controlling the upright posture. The effects of mTBI on postural stability are not only present shortly after the injury, but also as long-term effects of the trauma (Ingersoll and Armstrong 1992, Wober et al 1993, Slobounov et 2006b,2008). These individuals present a larger, slower, and more random body sway compared to individuals with no history of mTBI (Ingersoll and Armstrong 1992, Wober et al 1993).

Besides behavioral studies and clinical observations, several studies using other techniques have advanced knowledge in how the CNS produces movements and controls human posture. Deecke et al (1969) studied the electroencephalogram (EEG) in humans. They reported a late activation of structures within the primary motor area (the dorsal portion of the frontal lobe) compared to that within the other portions of the frontal lobe, suggesting that the primary motor area serves as a muscle activator instead of a planner. A few studies using EEG techniques to investigate postural control have suggested that postural adjustments and compensatory postural movements involve supraspinal structures, such as the cerebral cortex (Dietz et al 1985, Dimitrov et al 1996, Slobounov et al 2000, 2005, Quant et al 2004a,b, Adkin et al 2006). For example, Slobounov and colleagues described an increase in the gamma-range activity about 200 ms prior to the reversal point during body sway in the anterior direction (Slobounov et al 2005). They

also described a decrease in the EEG power in individuals with a history of mTBI when performing a postural task requiring the recognition of unstable postures. They suggested that these individuals had an impaired ability to recognize the limits of their functional boundaries of stability (Slobounov et al 2000).

In another study using event-related functional Magnetic Resonance Imaging (fMRI), this group of researchers evaluated the ability of healthy individuals to recognize unstable positions of a projected image of a virtual body. They reported that the recognition of unstable postures induced the activation of distinct areas of the brain, including bilateral parietal cortex, anterior cingulate cortex, and bilateral cerebellum (Slobounov et al 2006a). The positron emission tomography (PET), which may be used to measure metabolic processes in the body, has also been used to investigate the role of brain structures in postural control. Based on an assumption that areas of high radioactivity are associated with brain activity, PET neuroimaging indirectly measures the blood flow to different parts of the brain. Ouchi (1999) reported activation of cerebellar structures and visual cortex during different standing postures, while no activation was shown during supine posture. In general, studies using brain imaging techniques support the hypothesis that some postural adjustments are not just automatic muscle responses to perturbation mediated by the brainstem and spinal cord. Rather, they are cortically controlled intentional movements.

2.1.4. Principles of Human Upright Postural Control

The complexity of human bipedal upright posture is related to the control of the center of pressure (COP, e.g., the point of application of the ground reaction force to the body) in order to keep the projection of the center of mass (COM) within the base of support. Several models have been suggested to explain upright control and describe the COP migration during upright stance. Five of them will be addressed in the following paragraphs.

The **inverted pendulum model** has been frequently used to describe human upright posture (Fitzpatrick et al 1992b, Winter et al 1993,1998, Morasso and Schieppati

1999). According to this model, the body sways in a small magnitude and the motion occurs only around ankle joints with the feet in a fixed position (Zatsiorky and King 1998). Winter et al (1998) suggested that the continuous movement of the COP regulates the projection of the whole body COM (considered the controlled variable) to fall within the base of support. They showed that the COP-COM error signal was proportional to the horizontal acceleration of the COP in the anterior-posterior and medio-lateral directions. According to their model, muscles act as springs to move the COP in phase with the COM as the body sways about a desired equilibrium position. This model also predicts instant corrective responses due to the stiffness of the postural muscles placed around the joints. Therefore, it is suggested that there is less necessity of the CNS to intervene along the process of postural control. However, the inverted pendulum model has been strongly criticized. According to Morasso and colleagues, muscle stiffness alone is not enough to control the body upright. Sensory information from the pressure receptors in the soles of the feet and from muscle receptors is also likely to contribute to the control of upright posture (Morasso and Schieppati 1999, Morasso and Sanguinetti 2002). Other scientists discussed that the assumption of a fixed reference point for stabilization of posture may not be correct since several studies have emphasized a moving reference point (Accornero et al 1997, Zatsiorsky and Duarte 1999). Another criticism came from the assumption that only motion at the ankle is of importance (Day et al 1993, Kuo and Zajac 1993, Accornero et al 1997, Aramaki et al 2001).

The **open- and closed-loop of postural control** was proposed by Collins and De Luca (1993). They decomposed the COP signal in two stochastic processes (open- and closed-loop of postural control), also called random walk or Brownnian movement. In the random walk analysis of the COP trajectory, for each instant of time, the next position of the COP is predicted to be at fixed amplitude and random direction. According to this model, an open-loop control mechanism was called to act during short-term intervals (< 1s), whereas a closed-loop mechanism was called to act during long-term intervals. Collins and De Luca (1993) suggested that the CNS allows a certain amount of "sloppiness" in the control of upright posture. In the case of the sensory systems indicating a COP movement beyond a certain threshold value, feedback mechanisms are then used to bring the COP back into a "safety zone".

Winter et al (1998) proposed **the decomposition of the gravity line** (the vertical line passing through the center of mass of the body). In their model, the COP migration was represented as an outcome of two processes: the gravity line migration and the COP deviation from the gravity line trajectory. Winter et al (1998) found a large negative correlation between the COP deviation from the gravity line trajectory and the horizontal force at a time lag of approximately 4 ms.

Another hypothesis of postural sway control proposed was the **Shifting-Fidgeting-Drifting patterns** of prolonged upright stance. Duarte and Zatsiorsky (1999) identified three patterns of COP behavior during a prolonged unconstrained standing task: **(a)** shifting (a step), a quick change in the mean position of the COP; **(b)** fidgeting (a pulse), a fast and large change of the mean position of the COP followed by a return of the mean position of the COP to approximately the same previous mean position; and **(c)** drifting (a ramp), a continuous and slow displacement of the mean position of the COP. This analysis has been sensitive in detecting changes in postural behavior among different populations (Freitas et al 2005, Lafond et al 2009).

The same researchers also proposed **the Rambling-Trembling decomposition of the stabilogram** (Zatsiorsky and Duarte 2000). The rambling-trembling decomposition was based on the *Equilibrium Point Hypothesis* of motor control. According to this hypothesis, the transient shifting of the COP from one instant equilibrium point to another is one of the mechanisms underlying the control of upright stance (Feldman 1986, Feldman and Levin 1995). The Rambling-Trembling Hypothesis suggests a superposition of two processes of upright stance control: the rambling and the trembling processes. The rambling component of the COP trajectory represents the migration of the reference point from one instant equilibrium point to the subsequent instant equilibrium point along a smooth trajectory; whereas the trembling component represents the COP migration around this moving reference point.

In general, human balance control depends on a continuous integration of sensory input and neuromuscular output. Based on sensorimotor integration, the CNS organizes the activation of postural muscles to maintain upright posture or to recover equilibrium after external perturbations or challenging postures (Bauer et al 2008, Santos et al 2008). There are basically two main mechanisms of postural control: the feedback and the

feedforward mechanisms (Peterka 2002, Seidler et al 2004, Papegaaij et al 2014). The **feedback control** is a reactive mechanism involving an online correction of posture based on ongoing sensory inputs, sensorimotor integration, and motor outputs; whereas the **feedforward control** is a predictive mechanism involving the anticipation of potential disturbances.

In addition, there are several lines of defense against the forces threatening body stability. The main mechanisms of postural control are presented in chronological order:

- **Anticipatory Postural Adjustments (APAs):** The CNS anticipates the disturbing effects of a voluntary movement (such as a step or reaching for a glass of water) or an expected perturbation by changing the background activity of postural muscles in a feedforward manner. APAs are commonly quantified by using EMG signals, body segment kinematics, and displacements of the COP. Examples of APAs include the shifting of the center of gravity in the opposite direction of the upcoming perturbation (Bouisset and Zattara 1983,1987); different patterns of COP displacement in preparation to stepping forward, stepping down, stepping up, and stepping over an obstacle (Degani et al 2007); and anticipatory muscle activity prior to fast voluntary arm raising (Ramos and Stark 1990). Such anticipatory adjustments are necessary to counteract upcoming forces and moments large enough to shift the COM outside the base of support and have the body at risk of fall.

APAs were first observed by Belen'kiy et al (1967). They reported the activation of postural leg muscles 50–100 ms prior to raising the arm in a standing posture. Previously, it was suggested that APAs were generated only in preparation of expected movements involving large body parts. However, recent studies have shown that APAs can also be generated by either small movements (such as finger movements) or even in the absence of movements. Aruin and Latash (1995b) reported that even a very small finger movement was enough to trigger APAs. Later, Shiratori and Latash (2001) reported that when a standing individual was required to catch a load without any movement, visual information about the falling object was sufficient to trigger APAs. Interestingly, a few studies reported no APAs when the experimenter triggered a load release from the participant's hands even though the participant knew that the

experimenter would release the load (Hugon et al 1982, Dufosse et al 1985, Paulignan et al 1989, Scholz and Latash 1998). Aruin and Latash (1995b) reported that only self-initiated perturbations were accompanied by APAs. Since then, several studies have described APAs prior to voluntary movements of the arm, leg, trunk, and head while standing (Belen'kiy et al 1967, Cordo and Nashner 1982, Breniere and Do 1986, Mouchino et al 1991, Danna-dos-Santos et al 2007b); forearm loading and unloading (Hugon et al 1982, Dufosse et al 1985, Lacquantini and Maioli 1989, Paulignan et al 1989, Aruin and Latash 1995a, Bennis et al 1996); and quick loading and unloading of the upper extremities during sitting and standing postures (Lavender et al 1993, Aruin and Latash 1995b,1996, Shiratori and Latash 2000).

In general, the generation of APAs depends on several factors, such as the magnitude and direction of the perturbation, the characteristics of the voluntary movement or expected perturbation, the current postural task, and time constraints. Aruin et al (1998) reported that situations of extreme stability/instability tend to decrease the magnitude of APAs or even abolish them. Slijper (2001) suggested that under situations of time constraint, such as reaction time, the APAs may be delayed and suboptimal.

The CNS origin of APAs is still not clear. Several studies have reported cortical activity before an expected perturbation (Jacobs et al 2008, Mochizuki et al 2008, 2009, Varghese et al 2016). Cortical activity associated with APAs has been reported prior to self-initiated postural perturbation of releasing a load (Mochizuki et al 2008), externally-cued perturbations (Jacobs et al 2008, Mochizuki et al 2009), and step initiation (Varghese et al 2016). The supplementary motor area (SMA) may also be involved in APAs. Jacobs et al (2009) investigated APAs to step initiation in participants with and without Parkinson's disease. Considering the progressive dysfunction of circuits associated with the SMA in individuals with Parkinson's disease, and findings revealing impaired timing of APAs, Jacob et al (2009) suggested the contribution of the SMA to the timing of the APAs to step initiation.

- **Passive elasticity of soft tissues surrounding joints (such as muscles, tendons, ligaments, and capsules) opposing perturbing forces:** The muscle-tendon unit plays an important role in postural control (Fitzpatrick et al 1992b, Winter et al 1998,

Morasso and Schieppati 1999). Their mechanical property generates an opposing force when the muscle-tendon unit is deformed. This property, commonly referred to as "*muscle stiffness*", will be termed "*apparent elasticity*" as suggested by Latash and Zatsiorky (1993). Studies have reported a linear relation between the ankle torque and the ankle angular displacement under very small postural perturbations when neither vestibular nor visual information was available (Fitzpatrick et al 1992b, Winter et al 1998). These researchers suggested that the apparent elasticity of the muscle-tendon unit around ankle joints acted as a spring, resulting in a COP displacement in phase with the COM displacement. Therefore, they suggested that the ankle passive elasticity was enough to maintain quiet standing. However, Morasso and Schieppati (1999) used their own method of computing "*stiffness*" and reported that restoring forces provided by the passive elasticity of the muscle-tendon unit were too low to maintain the vertical posture. It is important to note that these two groups of researchers defined stiffness differently, which might have led to the difference of opinion regarding the role of "*muscle stiffness*" in posture control.

 - **Stretch reflexes at the latency of 30–50 ms:** External small perturbations to upright posture are partly counteracted by stretch reflexes, which are muscle contractions in response to the muscle spindle stretching. Postural perturbations, such as the movement of a force platform inducing ankle dorsiflexion or plantiflexion, have shown muscle activity in response to the stretching of muscles acting on the ankles (Schieppati et al 1995). Nakazawa et al (2003) also observed the presence of stretch reflexes in the tibialis anterior muscle to stabilize the ankle joint during upright stance.

 - **Compensatory Postural Adjustments (CPAs):** Compensatory adjustments are feedback-based control mechanisms. They occur during the balance restoration phase after the perturbation. These adjustments may include changes in the magnitude of muscle activity and COP displacements (Park et al 2004, Maki and McIlroy 2006, Mohapatra et al 2012).

- **Preprogrammed reactions at the latency of 50–100 ms:** While passive resistance of the surrounding joint tissue and stretch reflexes are commonly used to maintain postural balance under small displacements of the COM, as happens during unperturbed stance; preprogrammed reactions emerge approximately 50–100 ms after a larger perturbation has been applied. These preprogrammed reactions consist of spatial and temporal muscle activation specific for a given perturbation.

Preprogrammed postural reactions in humans have been studied using moving force platforms. Stereotypical postural responses in leg and trunk muscles of healthy young adults were observed depending on the direction of the perturbation. For example, a small and slow backward translation of the platform led to a forward body sway and increased the activity of posterior muscles (soleus, biceps femoris, and lumbar erector spinae) at a latency of about 80 ms. The opposite happened when a small and slow forward translation was applied. In this case, the body swayed backward and the activity of anterior muscles increased (Horak and Nashner 1986). In both cases, muscles were recruited in a distal-to-proximal order, pulling the body either backward or forward. Although simplistic, this motor response around the ankle joint aiming corrective torques while keeping the feet in the same place is referred to as the "*ankle strategy*" (Horak and McPherson 1996, Gatev et al 1999, Horak 2006). The ankle strategy is commonly observed under small and slow postural perturbations.

In the case of larger and/or faster perturbations, the recruitment order of preprogrammed reactions in postural muscles changes to proximal-to-distal, and the individual tends to flex or extend the hip to minimize the excursion of the COM. This strategy is referred to as the "*hip strategy*". The hip strategy recruits muscles acting mainly on the hips, spine, and abdominals, while keeping the feet in the same place. Depending on the perturbation, the hip strategy may be used in conjunction with the ankle strategy to maintain equilibrium. When ankle and hip strategies occur together, the ankle and hip move in opposite directions (Horak and McPherson 1996, Gatev et al 1999, Horak 2006).

- **Voluntary actions:** Additional motor strategies commonly used to control balance are voluntary actions, such as the stepping and reaching strategies (Horak and

McPherson 1996, Gatev et al 1999, Horak 2006). These strategies are based on changing the base of the body support to accommodate the projection of the COM within its limits and restore equilibrium. The stepping strategy involves voluntary forward, backward, or lateral step(s) to restore balance, whereas the reaching strategy involves the use of the hand or another part of the body for support.

2.2. THE USE OF POSTURAL INDICES AND MUSCLE ACTIVITY TO INVESTIGATE POSTURAL CONTROL

Human postural control can be investigated by analyzing postural behavior during upright standing tasks. It can be assessed by either static or dynamic posturography. In static posturography, the individual is asked to stand as still as possible (unperturbed stance), and balance control will basically rely on feedback mechanisms. In dynamic posturography, the individual is asked to stand still while either a perturbation (auto perturbation or externally triggered perturbation) is applied or a second task is performed (such as a cognitive task or a voluntary movement). During unexpected perturbed stance, feedback mechanisms are responsible for corrective postural adjustments to restore balance, whereas expected perturbations and voluntary movements trigger APAs (feedforward control) and feedback mechanisms.

Current studies apply different quantitative techniques to measure balance, such as the analysis of the center of pressure and center of gravity, the relative position among body segments, the head and body segments position in space, and the patterns (magnitude and sequencing) of muscle activation (Collins and De Luca 1995, Winter 1995, Farina et al 2004, Duarte and Freitas 2010, Murillo et al 2012, Cimadoro et al 2013). These analyses provide more information regarding mechanisms underlying postural control than a simple clinical visual analysis of the upright stance. Despite promising advantages of using these laboratory techniques to assess balance, it is important to know what measures should be considered and how to translate biomechanical outcomes into clinically meaningful information.

2.2.1. Postural Indices

During unperturbed upright stance, the human body presents small oscillations. These oscillations have been quantified using different methods. The most frequently used biomechanical tool to assess postural sway is the force platform. This equipment can record the coordinates of the **center of pressure (COP)** in time, which is the displacement of the point of application of the ground reaction force to the body in time. Data extracted from the COP has been used to make inferences about postural sway behavior. Since the COP is the point of application of the resultant ground reaction force vector in a force platform, the COP position is given by two coordinates (*x* and *y*) (Winter 1995). The center of pressure in each anterior-posterior and medio-lateral direction (*COPap* and *COPml*, respectively) is computed based on six signals measured by the force platform (*x, y*, and *z* forces, and moments around *x, y* and *z* axis) and the height (*h*) of the base of support above the force platform:

$$COPap = (- h *Fx - My) / Fz \qquad\qquad \textbf{(2.1)}$$

$$COPml = (- h *Fy - Mx) / Fz \qquad\qquad \textbf{(2.2)}$$

In general, the COP migrates approximately 0.4 cm in the *anterior-posterior* direction and 0.18 cm in the *medio-lateral* direction during unperturbed bipedal stance. Interestingly, the magnitude of the COM displacement is smaller than that for the COP (Winter et al 1996,1998). The differences in migration of COP and COM in the *anterior-posterior* direction have been associated with the generation of torques at the ankle joints, whereas displacements in the *medio-lateral* direction have been associated with activity of hip muscles (Winter et al 1996).

Different variables can be extracted from the COP signals to measure different aspects of the body sway. Considering that each variable measures only part of the body's sway characteristics, it is necessary to use COP measures from different domains, such as spatio-temporal, frequency, and structural domains. In general, the variables in the **spatio-temporal domain** include, but are not limited to, the area of the dispersion of

the COP displacement, the total length of the COP trajectory, the amplitude of the COP displacement in the anterior-posterior and medio-lateral directions, the mean velocity of the COP displacement in each direction, and the variability of the COP displacement around its mean value (Collins and De Luca 1995, Pyykko 2000, Demura et al 2001, Kitabayashi et al 2002, Duarte and Freitas 2010).

Quantities in the **frequency domain** define important characteristics of a signal, such as the main frequencies composing it. The complexity (or lack of complexity) of a signal has not only been commonly linked to principles of posture control, but also provide the recognition of signal patterns associated with specific disease states (see Duarte and Freitas 2010 for review). These quantities are computed using the Fourier or spectral analysis, which decomposes the COP signal as a sum of the sine and cosine function with different amplitudes, frequencies, and phases (Collins and De Luca 1995, Pyykko 2000, Demura et al 2001, Kitabayashi et al 2002, Duarte and Freitas 2010). The resulting power spectral density profile (PSD) describes the levels of energy contained in each of the single frequencies composing the resulting COP. Some of the variables extracted from the COP signal in the frequency domain include the mean power frequency for each direction, the peak frequency for each direction, and the frequency band with either 50% or 80% of the spectral power for each direction. *Figure 2.2* shows the COP excursion of a healthy young adult standing for 60 seconds on a force platform (panel *A*), the decomposition of the COP trajectory into COPap and COPml (panel *B*), the power spectrum density (PSD) of the COPap (panel *C*), and the decomposition of the COPap into rambling (panel *D*) and trembling (panel *E*) components. The computation of these latter components of the COP (rambling and trembling) is explained later in the text.

(A) COP displacement
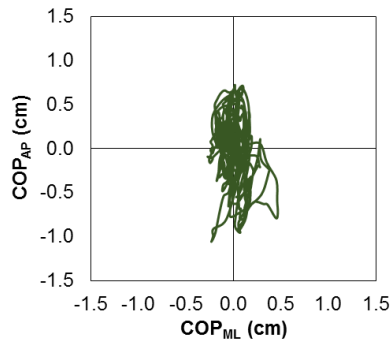
(B) COP displacement
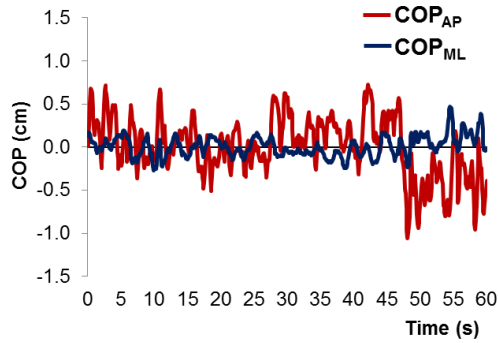
(C) PSD (*COP*$_{AP}$)
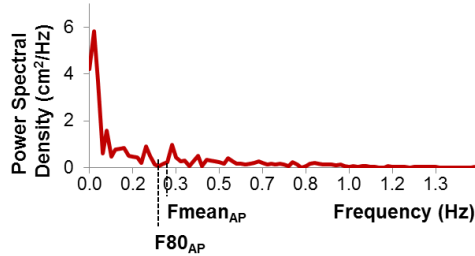
(D) COP$_{AP}$ and Rambling$_{AP}$
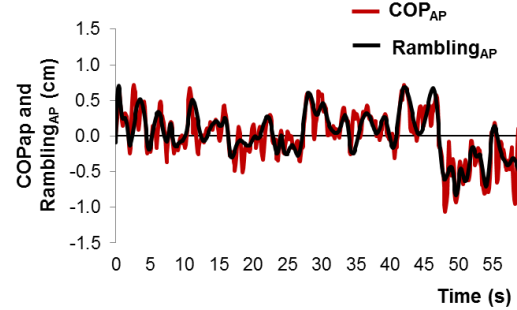
(E) Trembling$_{AP}$
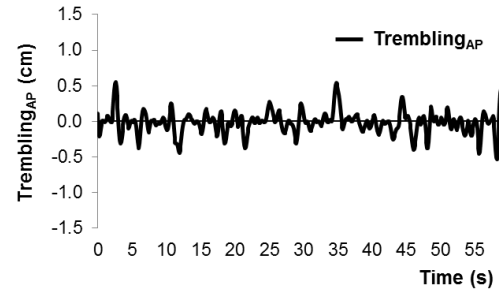
Figure 2.2. (A) The body's center of pressure (*COP*) excursion of a healthy young adult standing for 60 seconds. **(B)** The anterior-posterior and medio-lateral component of the COP displacement in time (*COPap* and *COPml*, respectively). **(C)** The power spectrum density (PSD) of the COPap. **(D)** The COPap and its respective rambling component (*Ramb$_{AP}$*). **(E)** Trembling component of the COPap displacement (*Tremb$_{AP}$*).

A few **structural analyses** of the COP have been proposed to analyze the COP signal (Duarte and Zatsiorsky 1999, Richman and Moorman 2000, Zatsiorsky and Duarte 2000). The dynamics of the human postural sway have been investigated using entropy analysis of the COP signal (Ramdani et al 2009, Borg and Laxaback 2010, Mei et al

2013, Rigoldi et al 2013, Clark et al 2014, Perez et al 2014, Fino et al 2015). The use of entropy analysis in biological signals was introduced by Pincus (1991). He proposed the *Approximate Entropy* method to investigate the complexity of the heart rate and beat-to-beat blood pressure. The innovative use of entropy analysis to quantify the unpredictability of the COP fluctuation in time was later proposed by Richman and Moorman (2000).

In general, there are two types of entropy: the state and the sequence entropies. The state entropy, such as the *Shannon Entropy* and the *Rényi Entropy*, quantifies the amount of information contained within the COP signal. These computational methods measure the probability of the signal occupying discrete states by examining the frequency that a COP position is visited throughout the signal without regarding the path to or from that position (Shannon 2001). In contrast, sequence entropies (such as the approximate entropy and its derivatives sample, multiscale, and composite multiscale entropies) examine the repetition of patterns within a signal. Sequence entropies evaluate the probability that particular values occur within a signal given that the sequence preceding that value is similar to a template sequence (Pincus 1991, Richman and Moorman 2000, Costa et al 2002).

Considering that the interpretation of entropy may vary between these methods, the sequence entropies may better assess the dynamical structure and regularity of the COP behavior. More specifically, the sample entropy of the COP signals has been successfully used to address the irregularity and randomness of body sway in healthy individuals (Ramdani et al 2009), older adults (Borg and Laxaback 2010, Fino et al 2015), and other populations (Mei et al 2013, Rigoldi et al 2013, Clark et al 2014, Perez et al 2014). Smaller sample entropy estimates indicate a more predictable and regular pattern of COP displacement in time, whereas higher estimates indicate a more irregular and random postural sway.

Another dynamic approach to measure postural sway is the Rambling-Trembling decomposition of the stabilogram proposed by Zatsiorsky and Duarte (2000). As a reminder, the rambling component represents the migration of the reference point from one instant equilibrium point to the subsequent one, while the trembling component represents oscillation around this moving reference point. This approach was introduced

previously in this chapter. *Figure 2.2* (panels *D* and *E*) shows the rambling and trembling components of the COPap in a representative healthy young adult standing for 60 seconds. Zatsiorsky and Duarte (2000) suggested that the rambling mechanism was related to supraspinal processes. The supraspinal mechanism of postural sway may be viewed as a sequence of drift-and-act episodes, where sensory inputs inform the CNS about changes in the body deviation from the vertical, and motor outputs are elicited for corrective actions (Milton et al 2009a,b). Zatsiorsky and Duarte (2000) also suggested that the trembling mechanism is related to spinal reflexes and changes in the intrinsic mechanical properties of muscles and joints.

This idea of two distinct processes controlling postural sway has been supported by other studies investigating the effects of visual feedback, different bases of support, external load, and peripheral impairments on the behavior of rambling and trembling trajectories (Mochizuki et al 2006, Danna-dos-Santos et al 2008, Shin et al 2011, Tahayori et al 2012). Danna-dos-Santos et al (2008) asked healthy adults to maintain their upright stance while a visual feedback of the COP displacement was provided. In order to maintain their COP inside different targets on the screen, participants decreased their rambling sway area and increased their trembling area as the target difficulty increased. In another study, an external load was applied in healthy adults performing upright stance (Tahayori et al 2012). Interestingly, there were changes in the trembling trajectory, independent of the area of postural sway and the rambling trajectory. The manipulation of the base of support also imposed changes in the behavior of rambling and trembling trajectories, reinforcing the idea of the rambling component of the COP representing a search strategy for postural stability (Mochizuki et al 2006). Their findings support the hypothesis that the rambling mechanism is activated by central processes, whereas the trembling mechanism is activated by peripheral processes and reflexes.

As observed, many COP variables are available to measure postural sway. It is important to understand the clinical meaning of each one when selecting the best approach to assess human upright control. While the variables are undoubtedly important when assessing balance, the protocol used to record COP data is equally fundamental in ensuring the quality of the measurements. Consideration should be given to the task(s) to be performed, the characteristics and age of the individuals, the duration of the task, and

the sampling frequency. The task, for example, may differ among studies by using different bases of support (such as feet apart, feet together, tandem position, semi-tandem position, and unipedal stance), manipulating sensorial input (such as occluding or distorting visual input and/or ankle joint somatosensory senses), or adding a secondary task. Gender, health status, activity level, anthropometric characteristics, and age also need to be considered. Another reason for discrepancies among studies is the duration of the task. The recommended trial duration to assess upright postural sway is between 60 s and 120 s (Lafond et al 2004, Kooij et al 2011). Despite the fact that COP signals with less than 60 s in upright posture may not provide enough data, several studies analyze only 30 s due to the task complexity or to avoid long duration of testing or fatigue. Therefore, the choice of trial duration depends on the clinical context. In addition, the sampling frequency should be carefully selected. Frequencies of 10 Hz, 20 Hz, or higher (100 Hz) for data acquisition have been used to assess human upright postural sway since the frequency bandwidth during human upright sway is below 10 Hz (Winter 1995). In addition, data with higher acquisition frequencies should be down sampled or filtered to eliminate signal noises. Therefore, the lack of standardized testing protocols to measure body sway using COP signals also explains discrepancies in the literature.

2.2.2. Multiple Postural Muscle Activation

As mentioned earlier, the control of upright stance depends on the ability of the CNS to generate adequate patterns of postural muscle activation. These patterns must ensure counter-action of all forces applied to the axial skeleton, implying an accurate distribution of forces across joints and temporal accuracy in their delivery. To understand how the CNS achieves this goal, signals extracted from electromyography (*EMG*) have been used as a general measure of motor unit activity. Recordings of EMG signals of postural muscles have been largely used to investigate muscle activation under different standing tasks. Traditionally, primary stabilizers of major axial joints (ankles, knees, hips, and intervertebral) are the focus of these studies.

Conventional methods for EMG signal processing involve analyses in the time and frequency domains. In the **time domain**, muscle activity can be assessed by quantifying the amplitude or magnitude of muscle activation, which is related to the recruitment and discharge rates of active motor units. In general, EMG magnitude can be a measure of estimation of muscle force using surface or intramuscular EMG recording electrodes. The magnitude of the EMG signal can be linked to activation level provided by the spinal cord. This is a general idea, since the signal extracted from EMG electrodes is influenced by several factors, such as electrode placement, thickness of the subcutaneous tissues, motor unit conduction velocities, positive and negative phased of motor unit action potentials leading to amplitude cancellation, and equipment used to record EMG signals (Farina et al 2004). It is also important to note that data extracted from surface EMG signals are global measures of the activity of motor units rather than firing rates of a single motor unit (Farina et al 2004). The measurement of the amplitude of muscle activation can be performed by two different approaches. In the first method, the filtered and full-wave rectified EMG signal is integrated over the desired time window. In the second method, the root mean square (*RMS*) of the EMG signal (EMG signals from several time frames from a recorded trial) is normalized by the maximal voluntary isometric contraction (*MVC*, which measures the maximal EMG activity) of this muscle (Murillo et al 2012, Cimadoro et al 2013). Other than the magnitude of muscle activation, analyses in the time domain may also include the timing and sequence of multiple muscle activation.

Conversely, signal processing in the **frequency domain** can be performed by spectral analysis of the EMG signal. The transformation of the EMG signal from the time domain to the frequency domain can be done by Fast Fourier Transform (*FFT*), as discussed previously in this chapter. As a reminder, some of the measures derived from the power spectrum of an EMG signal include the mean frequency (frequency at which the average power is reached), the peak frequency (frequency at which the maximum power occurs), and the median frequency (frequency at which 50% of the total power is reached).

Many previous studies have proposed reliable methods to investigate muscle synergies and test Bernstein's hypothesis of muscle synergies as a strategy to reduce the

number of degrees of freedom of the system (Gelfand and Latash 1998, Scholz et al 2000, Latash et al 2005,2007, Latash 2008). Scholz and Schöner (1999) developed a method to explore the synergistic features of the motor system, termed the *Uncontrolled Manifold Hypothesis* (*UCM Hypothesis*). According to their hypothesis, the controller (CNS) acts in the space of elemental variables (such as joint angles and digit forces) creating synergies to stabilize (decrease variability across repetitive trials) the performance variable (such as the end position or the total force produced). In other words, they investigated synergies using kinematics as the performance variables. More specifically, Scholz and Schöner (1999) applied the UCM method using the variability of the joint configuration in time during sit-to-stand tasks. The variability of these elemental variables was decomposed into components that did and did not affect the performance variable (the body's center of mass displacement). The component parallel to the uncontrolled manifold, termed the V_{UCM} component, represents the variability that does not exert any changes in the performance variable (also known as the "good variability"), whereas the component orthogonal to the UCM component, termed the V_{ORT} component, represents any variation of the magnitude of the muscle mode that influences the performance variable (also known as the "bad variability"). Scholz and Schöner (1999) reported that the position of the center of mass in the sagittal plane was very well stabilized by co-variations of the joint configurations. Therefore, the controller stabilizes a particular performance by allowing a variability of multiple elemental variables. It is achieved by selecting a subspace (termed uncontrolled manifold) within the space of the elements' actions related to the desirable value of the performance variable.

The first studies using the UCM approach employed mechanical variables (such as joint angles and finger forces) as elemental variables (Scholz and Schöner 1999, Scholz et al 2000, 2002). Follow-up studies used EMG signals as elemental variables to identify muscle synergies associated with postural tasks (Krishnamoorthy et al 2003a,b, Wang et al 2005,2006, Danna-dos-Santos et al 2007a,2008). Krishnamoorthy and colleagues (Krishnamoorthy 2003a,b) improved the UCM approach by implementing the following steps: (**1**) principal component analysis (*PCA*) on the indices of integrated EMG signals across repeated trials to identify the elemental variables, referred to as *muscle modes* or *M-modes*; (**2**) selection of a performance variable, such as the COP

displacement, shear forces, or moments around the vertical axis; **(3)** multiple linear regression to compute the Jacobian of the system to verify changes in the relationship between magnitudes of M-modes and COP shifts; and **(4)** UCM computation and decomposition of the variance in magnitude of the muscle modes into V_{UCM} and V_{ORT}. According to their findings, three M-modes were identified during upright stance: the "*push-back M-mode*" (formed by the gastrocnemius lateralis, gastrocnemius medialis, soleus, biceps femoris, semi-tendinosus, and erector spinae), the "'*push-forward M-mode*" (formed by the vastus lateralis, vastus medialis, rectus femoris, and tibialis anterior), and the "*mixed M-mode*" (formed by the tibialis anterior, rectus abdominis, vastus lateralis, and gastrocnemius lateralis). Wang et al (2005, 2006) reported a significant decrease in the index of multi-M-mode synergies stabilizing COP shifts during either a fast step or a fast voluntary upright body sway. They suggested that the structure of M-modes and/or their effects on COP sway depend on the velocity of the COP displacement. Moreover, Danna-dos-Santos and colleagues investigated multi-M-mode synergies under different challenging whole-body tasks. They analyzed cyclic voluntary anterior-posterior body sway at different frequencies and reported a dependency of multi-M-mode synergies on the speed of the voluntary COP shift (Danna-dos-Santos et al 2007a). These researches also observed flexible muscle modes during voluntary whole-body cyclical sway tasks at 0.5 Hz (bipedal or unipedal sway with eyes either open or closed) (Danna-dos-Santos et al 2008). Their findings revealed an increase in the number of M-modes (three to five) in function of the increase in the task complexity. They suggested that the controller manipulates a larger number of elemental variables under more challenging tasks.

In line with the organization of muscle synergies controlling human posture, relatively recent studies have proposed that a synchronization at lower frequency bands of neural oscillations from different sources within the CNS (also referred to as correlated or common neural inputs) may be the mechanism used by the CNS to coordinate a large-scale integration among its cortical and subcortical components (Farmer 1998, De Luca and Erim 2002, Santello and Fuglevand 2004, Semmler et al 2004, Johnston et al 2005, Winges et al 2008, Boonstra et al 2009, Poston et al 2010, Danna-dos-Santos et al 2010). In other words, the synchronization of neural oscillations seems to be related to the

activation of multiple muscles forming postural synergies. Considering that traces of such synchronizations within the CNS at lower frequency bands are likely to be embedded in EMG signals of target postural muscles (Farmer 1998, De Luca and Erim 2002), synchronization features, such as coherence between pairs of EMG signals, can be used to investigate the formation of multi-muscle synergies. *Figure 2.3* illustrates a schematic representation of the hypothesized synchronization of neural oscillations within the CNS diverging onto multiple postural muscles and forming postural synergies.
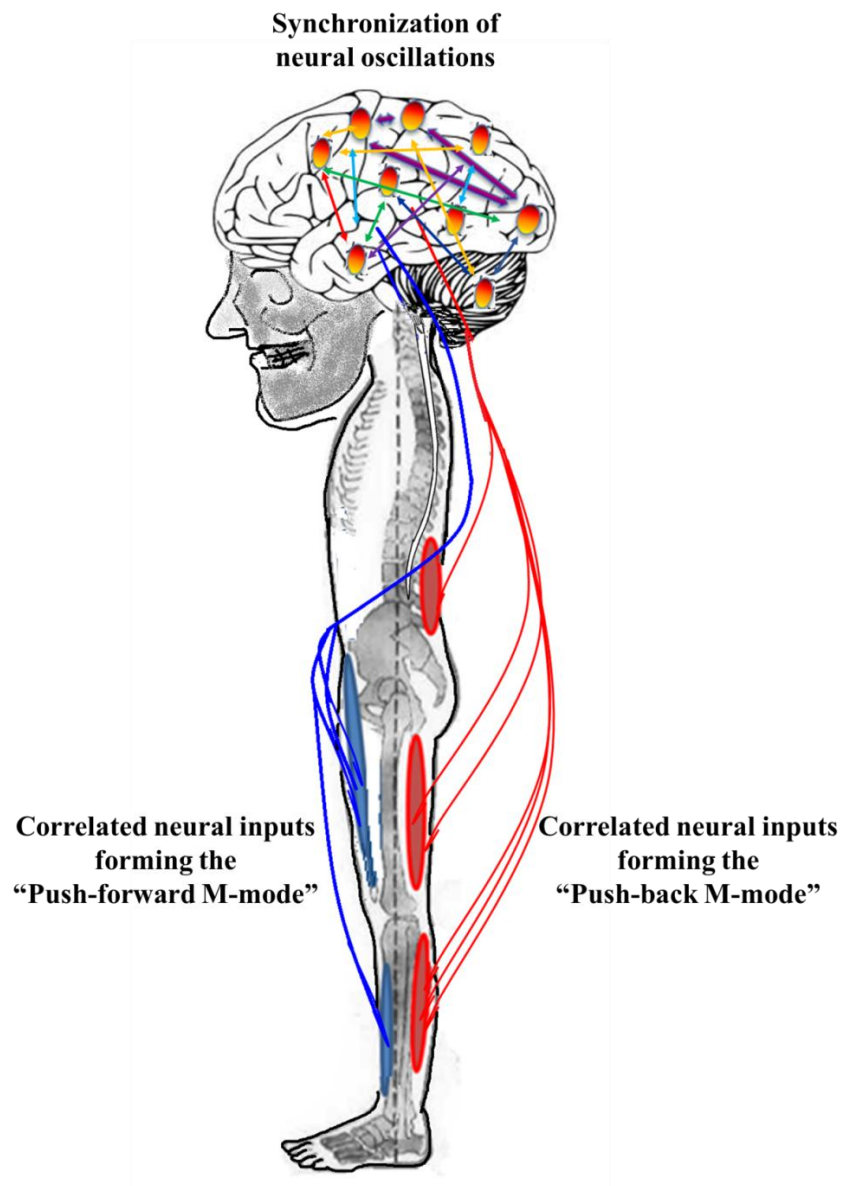
Figure 2.3. Schematic illustration of the hypothesized synchronization of neural oscillations in the Central Nervous System forming two distinct postural muscle synergies ("*push-back M-mode*" and "*push-forward M-mode*") during upright stance.

The *Intermuscular Coherence Approach* has recently been used to determine the formation of correlated neural inputs to skeletal muscles. In general, intermuscular coherence quantifies the correlations of muscle activation in the frequency domain, e.g., the synchronization between two EMG signals. The coherence between two signals is estimated using the cross-spectrum of two EMG signals (f_{xy}) squared and normalized by the product of the autospectrum of each signal (f_{xx} and f_{yy}) at each frequency (λ), as follows:

$$\left| R_{xy}(\lambda) \right|^2 = \frac{\left| f_{xy}(\lambda) \right|^2}{\left| f_{xx}(\lambda) f_{yy}(\lambda) \right|} \tag{2.3}$$

Coherence is considered statistically significant if it exceeds the significance limit of the null distribution, computed as proposed by Rosenberg et al (1989). The significance limit for zero coherence at $\alpha = 0.05$ and for the number of disjoint segments (L) is determined by the following equation:

$$Sig(\alpha) = 1 - (1-\alpha)^{\frac{1}{L-1}} \tag{2.4}$$

Figure 2.4 shows the EMG signal of both right tibialis anterior and right soleus muscles and the intermuscular coherence estimate profile of a representative individual during upright bipedal stance. Considering that significant coherence is observed when coherence estimates are above the significant level, there was a significant synchronization of the right tibialis anterior and right soleus within the frequency band of 0–12 Hz. This frequency interval represents the frequency distribution of the coherence between these two muscles during upright bipedal stance. In this case, the strength of such synchronization can be computed by the integral of the coherence profile over the frequency band of interest (0–12 Hz).
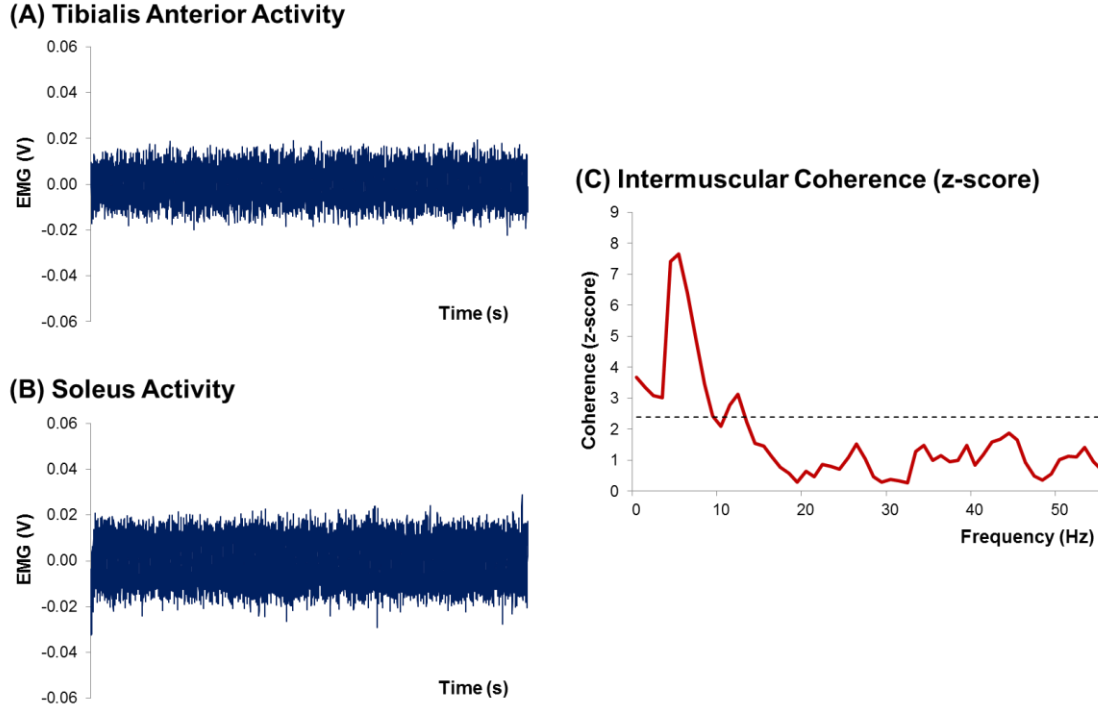
Figure 2.4. Filtered EMG signal of the tibialis anterior (panel *A*) and soleus (panels *B*) muscles during upright stance of a representative individual, along with the z-scored intermuscular coherence estimate profile for this muscle pair and respective significance level (panel *C*, *red line* and *black dashed line*, respectively).

In addition to the intermuscular coherence for pairs of muscles, estimates of coherence obtained for each muscle pair can be combined to calculate pooled coherence estimates across muscle pairs, as proposed by Amjad et al (1997) as follows:

$$\frac{\left|\sum_{i=1}^{k} f_{xy}(\lambda)L_i\right|^2}{\left(\left(\sum_{i=1}^{k} f_{xx}(\lambda)L_i\right)\left(\sum_{i=1}^{k} f_{yy}(\lambda)L_i\right)\right)} \tag{2.5}$$

Recently, a few studies have used intermuscular coherence analysis during object grasping tasks (Boonstra et al 2009, Danna-dos-Santos et al 2010, Poston et al 2010) and during the execution of whole-body tasks (Boonstra et al 2008,2009). Poston et al (2010) reported not only the presence of correlated neural inputs to the hand muscles, but also a

singular distribution of these inputs among intrinsic and extrinsic muscles. Danna-dos-Santos et al (2010) have demonstrated the effects of hand muscle fatigue on the distributions of correlated neural inputs previously reported by Poston et al (2010). In addition, Boonstra et al (2009) reported that pairs of homologous muscles from both lower limbs showed signs of synchronization during the execution of voluntary whole-body anterior-posterior oscillations. These studies suggest that correlated neural inputs to skeletal muscles have specific spatial distributions, strength, and periodicity.

2.3. THE EFFECTS OF PHYSIOLOGICAL AGE-RELATED CHANGES ON BALANCE

The natural process of aging is associated with degenerative processes of many physiological systems (such as cardiovascular, musculoskeletal, cognitive, sensory, and nervous systems), affecting balance control and increasing susceptibility to falls. *Figure 2.5* illustrates structural and functional age-related changes in different systems contributing to balance deficits and consequent increased risk of fall.
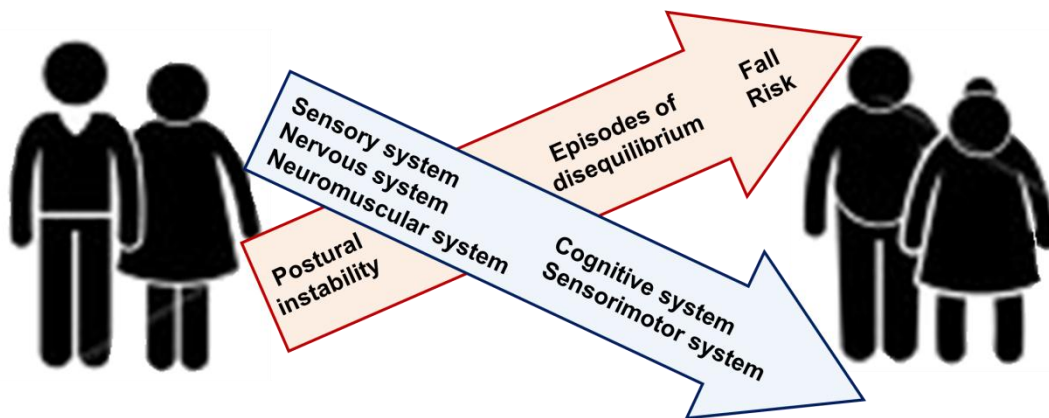
Figure 2.5. Structural and functional age-related changes in different systems contributing to balance deficits and consequent increased risk of fall.

Even though this idea seems simple, the aging process occurs at different rates across different tissues, and with accelerated declines with advanced age (Nair 2005, Shaffer and Harrison 2007). The non-homogenous rates of tissue degeneration affect the

functionality of the anatomic structures. Physiological changes seem to evoke a reorganization of cortical and spinal control of movements, which may lead to a suboptimal but functional performance or failure, such as falling. The knowledge concerning the contribution of the sensorimotor system on postural control and the effects of aging on physiological systems is essential to understanding why some individuals can overcome this challenge better than others (ex. fallers *vs* nonfallers).

Structural and functional changes in **sensory systems** due to the aging process include reduced visual acuity and accommodation, contour and depth perception, contrast sensitivity, peripheral vision, pupil size and agility, fluid balance in the semicircular canals of the inner ear, vestibular function, kinesthetic sensitivity, joint position sense at the ankle, and cutaneous sensation (Kelly 1993, Lewis 2002, Wiesmeier et al 2015). These changes reduce the redundancy of sensory information, increasing the upright postural instability in older adults. In addition, older adults appear to have difficulties in sensory reweighting, i.e., in ranking the importance of each sensory information to maintain balance (Horak et al 1989, Teasdale and Simoneau 2001, Eikema et al 2012,2014).

Age-related structural and functional changes in the **neuromuscular system** also play an important role in postural control. Some of the structural changes in skeletal muscles include the progressive decrease in muscle protein synthesis (myosin heavy chains and mitochondrial proteins) (Vandervoort 2002, Nair 2005), the increase in intramuscular fat (Kent-Braun et al 2000), the degeneration of muscle spindles (Skinner 1984), the decrease in tensile stiffness of tendons indicating weaker tendons (Reeves 2006), and the decrease in the size and number of muscle fibers (Lexell 1995). Some of the age-related structural changes in the nervous system include the progressive degeneration of the mass of the gray and white matter, leading to an atrophy of the brain tissue and thinning of the cortical tissue (Good et al 2001); the degeneration of alpha motoneurons (Campbell et al 1973, Doherty et al 1993b); the decrease in synaptic density (Haug and Eggers 1991); and the increase in motor unit size indicating an increased number of muscle fibers per motor unit (De Koning 1988).

In addition, functional changes in the neuromuscular system include the progressive decrease in the number of viable motor units (Campbell et al 1973, Doherty

et al 1993b); the decrease in descending commands for voluntary muscle activation (Yue et al 1999); the decrease in axonal conduction velocity (Doherty et al 1993a, Wang 1999); the decrease in the excitation-contraction process leading to reduced response amplitude (Delbono 2000, Rivner et al 2001); sarcopenia (age-related loss of skeletal muscle mass and strength) (Bemben et al 1991); and the reorganization of muscle response to balance adjustments (Papegaaij et al 2014). Regarding progressive sarcopenia, studies report a decrease of 20–40% in skeletal muscle mass at the age of 80 (Rice et al 1989, Porter et al 1995) and a 20–50% decrease in skeletal muscle strength (Shumway-Cook and Woollacott 2001, Doherty 2003). In addition, Tsai et al (2014) reported a decrease in postural muscle power in older adults. The rate of mass and strength loss, however, differs among skeletal muscles (Doherty 2003). The lower extremity muscles appear to lose more strength than upper extremity muscles (Aniansson et al 1996, Frontera et al 2000). In addition, some studies reported a larger loss in distal muscles compared to proximal ones (Nakao et al 1989), whereas other studies did not report the same findings (Viitasalo et al 1985).

Regarding the pattern of muscular activation, older adults seem to reorganize the relative contribution of cortical and spinal inputs during voluntary contraction. There is an increase in cortical activation (cortical mechanism) and a decrease in the modulation of presynaptic inhibition (spinal mechanism) (Papegaaij et al 2014). These changes seem to be related to the muscular co-activation pattern observed in older adults. In other words, a stiffening of joints by agonist-antagonist cocontraction seems to be a common strategy of postural control used by older adults to improve instability (Woollacott et al 1988, Manchester et al 1989, Melzer et al 2001, Laughton et al 2003, Benjuya et al 2004, Klass et al 2007, Tucker et al 2008, Baudry et al 2010, Nagai et al 2011, Papegaaij et al 2014, Lee et al 2015, Craig et al 2016). In general, this co-activation of agonist and antagonist muscles during upright stance associated with joint stiffness, particularly in distal joints (ankles), **(a)** increases postural rigidity and potentially restrict dynamic responses to postural control due to the reduced number of degrees of freedom (Ge 1998, Tucker et al 2008); **(b)** reduces the performance of agonist muscles (Pereira and Gonçalves 2011); **(c)** increases the energetic cost of transport (Mian et al 2006); and **(d)** increases chances of fatigue (Hortobagyi et al 2009). Therefore, unlike a flexible system

using optimal spatio-temporal patterns of reciprocal muscle activation to control posture, older adults tend to stiffen their lower extremity joints by antagonist cocontraction in order to compensate for the lack of their ability to fine-tune postural adjustments during upright stance.

The increase in co-activation of antagonist postural muscles have been reported not only during static standing (Melzer et al 2001, Laughton et al 2003, Benjuya et al 2004, Nagai et al 2011, 2013), but also during single-joint movements of the arms (Klein et al 2001), gait (Mian et al 2006, Schmitz et al 2008), stepping down (Hortogagyi and DeVita 2000), and, mostly, as a response to postural perturbation (Woollacott et al 1988, Manchester et al 1989, Lin and Woollacott 2002, Maki and McIlroy 2006, Lee et al 2015). In general, age-related changes in compensatory postural adjustments (CPAs) may include increased magnitude of muscle activity and larger COP displacements (Maki and McIlroy 2006). Other CPAs observed in older adults include longer latency of responses in muscles acting in the ankle, increased cocontraction of antagonist muscles, changes in the temporal sequencing of the distal and proximal activation, and increased variability in the magnitude of activation of proximal and distal muscles in response to perturbation (Woollacott et al 1986,1988, Lin and Woollacott 2002, Tsai et al 2014, Lee et al 2015, Craig et al 2016). Tsai et al (2014) reported an asymmetrical increase in the EMG activity of postural muscles in response to perturbation in older adults. This asymmetric pattern might be a contributing factor of unstable postural responses as the individual ages. In another study, Lee et al (2015) reported age-related changes on CPAs when individuals were asked to push an object (a pendulum attached to the ceiling) using both hands. Older adults used co-activation of postural muscles (tibialis anterior, medial gastrocnemius, rectus femoris, biceps femoris, rectus abdominis, and lumbar erector spinae), whereas young adults used reciprocal activation of muscles. As observed, older adults co-activate agonist and antagonist postural muscles as their compensatory mechanism to overcome aging-related deficits in postural control, a typical motor strategy observed when function is not optimal.

In summary, the decline in postural control due to the natural process of aging leads to a reorganization of sensorimotor integration and motor output to maintain

balance and avoid falls. Consequently, failure in feedback and feedforward mechanisms results in suboptimal corrective torque and possible loss of balance.

Current efforts of health professionals to reduce episodes of fall include fall risk screenings and functional performance-based tests to assess balance. Fall risk screenings are usually based on key fall risk factors, such as age, previous falls, decreased level of daily physical activity, medications, vision health, risk taking behaviors, home safety issues, fear of falling, depression, use of assistive devices, and leg muscle strength. Some of the tests available to be performed in clinical settings include the following: the "*Performance Oriented Mobility Assessment*" scale (POMA) to assess balance and mobility; the "*Timed Up and Go test*" (TUG) to assess mobility; the "*Berg Functional Balance Scale*" (BBS) to assess balance; the "*30-Seconds Chair Stand Test*" to assess leg strength and endurance; the "*Functional Reach Test*" to measure postural control; the "*Four Square Step Test*" (FSST) to assess fall risk; the "*Balance Evaluation Systems Test*" (BESTest) and its mini version (mini-BESTest) to assess balance problems; the "*Short Physical Performance Battery*" (SPPB) to assess balance; the "*4-Stage Balance Test*" to assess balance; the "*Clinical Test of Sensory Integration for Balance*" (CTSIB) to evaluate the contribution of sensory systems to control balance; the "*Activities-Specific Balance Confidence (ABC) Scale*" to assess balance; the "*Dynamic Gait Index*" (DGI) to assess gait, balance, and fall risk; and the "*Gait Assessment Rating Score*" (GARS) to assess gait (Shumway-Cook and Horak 1986, Tinetti 1986, Duncan and Weiner 1990, Wolfson et al 1990, Podsiadlo and Richardson 1991, Berg et al 1992, Powell and Myers 1995, Shumway-Cook and Woollacott 1995, Dite and Temple 2002, Herman et al 2009, Franshignoni et al 2010, Freire et al 2012). See Konrad et al (1999), Ambrose et al (2013), and Noohu et al (2013) for general information about the aforementioned tests.

Another recognized tool to assess fall risk is the STEADI (Stopping Elderly Accidents, Deaths & Injuries) toolkit created by the Centers for Disease Control and Prevention's Injury Center (CDC). This tool kit includes educational brochures about fall prevention for both health professionals and older adults, and it can be found online at http://www.cdc.gov/steadi/materials.html. In addition, the "Falls Free Initiative" program from the National Council on Aging, a nonprofit service and advocacy organization representing older adults and the community organizations that serve them, promotes

some evidence-based programs to prevent falls (such as "A Matter of Balance", "Stepping On", "Tai-Chi: moving for better balance", and "The Otago exercise program"). The Falls Free Initiative also sponsors the annual "Falls Prevention Awareness Day" on the first day of Fall, a national fall risk screening and fall prevention education event.

All these clinical tests and fall risk screenings are valid tools to address the necessity of preventive and/or rehabilitative interventions to improve balance. However, they do not identify the underlying causes of balance deficits. Quantitative assessment of postural control using measures extracted from COP and EMG signals in laboratory settings provides advanced knowledge regarding age-related changes impacting balance mechanisms. Current literature on the effects of aging on muscle activation and postural sway behavior is presented in the next section.

2.4. THE EFFECTS OF AGING ON MUSCLE ACTIVATION AND POSTURAL SWAY

Electromyography and force platform systems have been used to investigate postural control in different populations. Traditional measures extracted from EMG and COP signals include **magnitude** and **sequencing** of muscle activation, sway **area**, sway **amplitude**, sway **velocity,** and sway **frequency**.

Regarding the control of muscle activation during unperturbed stance, studies have shown age-related changes in both duration and amplitude of motor unit action potentials (Howard 1988, Doherty et al 1993a, Roos 1999). Increased EMG activity of muscles acting on the ankles (tibialis anterior and soleus) and hips (rectus femoris and semitendinosus) during unperturbed upright stance has been reported in older adults, compared to young adults (Amiridis et al 2003, Benjuya et al 2004, Nagai et al 2011). Studies have primarily shown a higher EMG activity of the tibialis anterior (pulling the body forward) and semitendinosus (keeping the hip extended) in older adults, compared to that in young adults, under different static stance tasks, such as wide base of support, narrow base of support, tandem Romberg stance, and unipedal stance (Amiridis et al

2003, Benjuya et al 2004). A higher activation of the muscles acting on the ankle (tibialis anterior and soleus) in older adults, compared to that in young adults, was also reported not only during unperturbed static stance, but also during dynamic stance tasks, such as the functional stability boundary task (Nagai et al 2011). Interestingly, Benjuya et al (2004) reported no significant effects of vision (eyes open or eyes closed) and the size of the base of support (wide or narrow) on the magnitude of activation of the muscles around the ankle or knee/hip in young adults during unperturbed bipedal stance. However, older adults increased the EMG activity of the tibialis anterior when they stood up with eyes closed and with a narrower base of support (Benjuya et al 2004).

The use of EMG analysis has also confirmed the use of the **cocontraction strategy** by older adults to control balance and avoid falls (Laughton et al 2003, Benjuya et al 2004, Mian et al 2006, Schmitz et al 2008, Nagai et al 2011,2013, Lee et al 2015). Nagai and colleagues investigated the cocontraction index for the muscles around the ankle (tibialis anterior and soleus) in young and older adults performing static (unperturbed bipedal stance) and dynamic (functional reach, functional stability boundary, and gait) tasks. They revealed a significantly higher co-activation of these two muscles in older adults compared to that in young adults (Nagai et al 2011,2013). Moreover, older adults with less physical function presented higher muscle co-activation of ankle muscles compared to older adults with better physical function. Benjuya et al (2004) also found an increased cocontraction around the ankle (tibialis anterior and soleus) in older adults under different static postures (narrow and wide base of support with eyes either open or closed). In another study, Cattagni et al (2016) showed an association between ankle muscles weakness and increased body sway during upright stance. They suggested that postural stability impairment in older adults is highly related to ankle muscles weakness.

The effects of aging on postural control can also be detected by analyzing postural sway behavior. Subtle changes in body sway can be measured by changes in COP signals. Despite discrepancies in the literature due to variation in experimental protocols used to measure body sway, there is general agreement that older adults present a larger and faster body oscillation during bipedal upright stance when compared to young adults (Maki et al 1990, Prieto et al 1996, Amiridis et al 2003, Choy et al 2003, Benjuya et al

2004, Freitas et al 2005, Maurer and Peterka 2005, Demura et al 2008, Seigle et al 2009, Vieira et al 2009, Silva et al 2013, Wiesmeier et al 2015). The larger postural sway in older adults has been verified by a larger area of COP sway (Prieto et al 1996, Benjuya et al 2004, Demura et al 2008, Silva et al 2013), an increased total length of the COP path (Benjuya et al 2004), and an increased peak-to-peak amplitude of the COP sway (Amiridis et al 2003, Benjuya et al 2004, Demura et al 2008, Wiesmeier et al 2015). These studies compared young and older adults performing between 5 to 60 seconds of upright stance under different conditions, such as eyes open or closed, narrow or wide base of support, and bipedal or unipedal stance. Interestingly, Seigle et al (2009) found a significantly increased length of the COP path during bipedal stance with eyes closed in older adults, but not during bipedal stance with eyes open. Moreover, Seigle et al (2009) and Vieira et al (2009) did not find significant difference in the elliptical area of the COP sway of older adults, compared to that for young adults, during bipedal stance with eyes either open or closed. The variability of the COP displacement, computed by either standard deviation or root mean square (*RMS*) of the COP displacement, also seems to increase with age. Amiridis et al (2003) reported higher COP standard deviation in older adults compared to young adults, in both directions, under either bipedal, Romberg (feet together) or unipedal stance for 5 seconds; Silva et al (2013) reported higher RMS of the COP displacement in older adults during 30 seconds of unipedal stance; and Wiesmeier et al (2015) reported higher RMS during 60 seconds of bipedal stance. Controversially, Freitas et al (2005) did not find significant difference in the RMS of the COP displacement among young and older adults performing bipedal stance for 60 seconds.

Regarding the mean velocity of the COP displacement, several studies reported a significant higher velocity during upright stance in older adults compared to young adults (Maki et al 1990, Baloh et al 1994, Prieto et al 1996, Choy et al 2003, Benjuya et al 2004, Freitas et al 2005, Maurer and Peterka 2005, Demura et al 2008, Silva et al 2013, Wiesmeier et al 2015). These studies used different experimental designs to measure COP velocity, such as trial duration (20 to 60 seconds) and postural tasks (bipedal, bipedal with feet together, unipedal, eyes open, and eyes closed). Conversely, Seigle et al (2009) found significant difference in the mean velocity of the COP displacement only

when the eyes were closed, and Vieira et al (2009) found significant difference in the mean velocity only in the medio-lateral direction for eyes either open or closed.

The spectral power of the COP displacement, usually overlooked, is also useful to detect changes in postural behavior. Traditional variables extracted from the power spectrum density (PSD) of the COP signal are the mean frequency and the frequency containing 80% of the COP spectral power. In general, older adults present higher mean frequency of the COP oscillation compared to young adults during unperturbed bipedal stance (Maki et al 1990, Wiesmeier et al 2015). In particular, for both young and older adults, Wiesmeier et al (2015) reported a higher mean frequency of the COP oscillation in the medio-lateral direction compared to the anterior-posterior direction, and also a higher mean frequency with eyes closed, compared to eyes open. Silva et al (2013) also reported an increased mean frequency of the COP oscillation in older adults during unipedal stance for 30 seconds, compared to young adults. Regarding the frequency band with 80% of the spectral power, Freitas et al (2005) reported significant higher frequency in the anterior-posterior direction for older adults compared to that measured in young adults, but no significant difference in the medio-lateral direction. Vieira et al (2009) also reported an increase in the frequency containing 80% of the COP spectral power in older adults performing bipedal stance with eyes open. However, they did not find significant differences when the eyes were closed.

Recent attempts to investigate the effects of aging on the dynamical structure of postural sway are promising. The use of entropy analysis to measure the randomness and irregularity of the COP behavior is yet to be explored in both healthy individuals and those with balance disorders. Duarte and Sternard (2008) and Borg and Laxaback (2010) measured the multiscale entropy and the sample entropy, respectively, to compare the level of randomness of the COP displacement between young and older adults. Both studies found a significantly higher irregularity of the anterior-posterior COP displacement in older adults compared to that in young adults, and no significant difference in the medio-lateral direction. Borg and Laxaback (2010) and Fino et al (2015) used sample entropy analysis and reported a more irregular pattern of COP displacement when older adults closed their eyes, compared to their bipedal stance with eyes open. Moreover, Borg and Laxaback (2010) reported higher COP irregularity of fallers

compared to nonfallers among older adults performing bipedal stance with eyes closed. Conversely, Seigle et al (2009) measured the Shannon entropy and reported no significant difference between young and older adults performing bipedal stance with eyes open and a more repetitive pattern of COP displacement for older adults with closed eyes compared to young adults. Despite all these findings, the differences among studies lie in the difference in the form of entropy and its computations.

Another method used to investigate the structure of the postural sway is the decomposition of the COP signal into rambling and trembling components (Zatsiorsky and Duarte 2000). Recently, Sarabon et al (2013) compared the mean velocity, mean frequency, and variability (root mean square) of each component of the postural sway in both directions between young and older adults performing bipedal stance with eyes either open or closed. Their results revealed stronger effects of age in the rambling component compared to that in the trembling component. However, it was an exploratory study and more research is necessary to better understand the age-related changes in the rambling and trembling components of the COP signal.

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CHAPTER 3

Multi-muscle control during bipedal stance:
an intermuscular coherence analysis approach

3.1.   INTRODUCTION

The control of the human body's vertical posture requires harmonic modulation of multiple muscles with temporal, spatial, and magnitude precision. The large and redundant number of muscles, the presence of multi-articular muscles, and the inherent vertical arrangement of major body segments during bipedal stance are some of the challenges to be overcome during the generation and distribution of the neural drive to skeletal muscles. One of the mainstays of development in motor control has been the hypothesis that the central nervous system (CNS) unites motor components to functional groups in order to overcome some of these challenges (Bernstein 1967). As a result, an extensive line of scientific work exploring this hypothesis has developed several methodological approaches to identify the existence of such functional groups as well as their composition. Among others, correlation techniques and methods of matrix factorization have been successfully used to identify small sets of variables that described the dynamic behavior of muscles across a variety of actions (Maier and Hepp-Reymond 1995, D'Avella et al 2003, Ivanenko et al 2004,2006, Weiss and Flanders 2004, Tresch et al 2006). In the last decades, novel methodologies have not only allowed the identification of functional muscle groups and their components, but also linked their interaction to the control of important physical variables (Scholz and Schöner 1999, Scholz et al 2000,2001, Krishnamoorthy et al 2003a, b,2004, Latash et al 2002,2003,2010, Danna-dos-Santos et al 2007,2008,2009, Latash 2008). One such method is the Uncontrolled Manifold Analysis (*UCM*) applied during the execution of whole-body tasks in bipedal stance (Krishnamoorthy et al 2003a). These researchers reported three major functional muscle groups (termed muscle modes or M-modes) emerging and co-varying their magnitude to provide a stable trajectory of the position of

the body's center of pressure (*COP*). This phenomenon has been replicated for other task settings (Danna-dos-Santos et al 2007,2008,2009, Robert et al 2008).

Collectively, these studies strongly suggest the involvement of muscle synergies controlling human posture. However, the neurophysiological mechanisms underlying the formation of these synergistic muscle groups remain largely unknown. Recently, it has been proposed that correlated neural inputs may be the mechanism used by the CNS to coordinate the activation of muscles forming a synergistic muscle group (Farmer 1998, De Luca and Erim 2002, Santello and Fuglevand 2004, Semmler et al 2004, Johnston et al 2005, Winges et al 2008, Boonstra et al 2009, Danna-dos-Santos et al 2010, Poston et al 2010). This proposition is based on the principle that the synchronization of neural oscillations at lower frequency bands may be the mechanism used by the CNS to achieve large-scale integration among its cortical and subcortical components, including those involved in the generation and control of movements. As a result, traces of such synchronizations at lower frequency bands within the CNS are likely to be embedded in electromyographic signals (*EMG*) of targeted skeletal muscles (Farmer 1998, De Luca and Erim 2002). Therefore, it has been proposed that synchronization features, such as the coherence between pairs of EMG signals, can be used to investigate the formation of multi-muscle synergies.

The intermuscular coherence approach has been successfully used to determine the distribution of correlated neural inputs to skeletal muscles during object grasping tasks (Boonstra et al 2009, Danna-dos-Santos et al 2010, Poston et al 2010) and during the execution of whole-body tasks (Boonstra et al 2008,2009). In general, these studies indicated that correlated neural inputs to skeletal muscles have specific spatial distributions, strength, and periodicity. However, to date, no investigations have focused on the hypothesis of correlated neural inputs among postural skeletal muscles .

Therefore, to further advance knowledge on the mechanisms underlying the organization of multi-muscle synergies, the present study was designed to investigate the possible existence, distribution, and strength of correlated neural inputs to three postural muscles previously recognized by Krishnamoorthy et al (2003a) as components of a synergistic muscle group. If these postural muscles are, in theory, part of the same neural network formed by synchronized neural oscillations within the CNS, they would be

synchronized at lower frequency bands. Therefore, the present study was based on the rationale that these synergistic muscles would exhibit signs of correlated neural inputs in the form of significant intermuscular coherence within a distinct lower frequency band.

3.2.   METHODS

Participants. Nine healthy participants (4 females and 5 males, mean age 29.2 years old and *SD* = 6.1, mean height 1.71 m and *SD* = 0.74, mean weight 78.8 kg and *SD* = 20.5) participated in this study. All participants were healthy and had no history of neurological or muscular disorder. All participants were right-handed based on their preferential hand usage during writing and eating. Prior to their participation, all participants voluntarily gave informed consent based on the procedures approved by the Institutional Review Board at The University of Montana and conformed to The Declaration of Helsinki.

Apparatus. A force platform (AMTI BP400600, AMTI Inc.) was used to acquire the vertical and horizontal components of the ground reaction force as well as the moments of force around the frontal and sagittal axes. These signals were transmitted to a dedicated system (Vycon MX Ultranet and Vycon Nexus version 1.6.1, Vycon®) for the computation of the body's center of pressure coordinates in anterior–posterior and medio–lateral directions (*COPap* and *COPml*, respectively). Features of the COP were recorded due to previous reports relating low-frequency COP modulation to low-frequency EMG modulation (Mochizuki et al 2006). Consistent with previous studies, COPap and COPml were defined by

$$COPap = (-h * F_\mathrm{x} - M_\mathrm{y}) / F_\mathrm{z} \tag{3.1}$$
$$COPml = (-h * F_\mathrm{y} - M_\mathrm{x}) / F_\mathrm{z}, \tag{3.2}$$

where h = the height of the base of support above the force plate; $F\mathrm{x}$ = horizontal anterior–posterior component of the resultant force applied to the force plate; $F\mathrm{y}$ = horizontal medio–lateral component of the resultant force applied to the force plate; $F\mathrm{z}$ =

vertical component of the resultant force applied to the force plate; Mx = moment of force around the sagittal axis; and My = moment of force around the frontal axis.

Surface electrodes were used to record the activity of the following muscles: soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE). The electrodes were placed on the right side of the participant's body and over the muscle bellies based on recommendations reported in the literature (Basmajian, 1980): SOL electrode laterally below the heads of the gastrocnemius bellies, BF electrode at half way between the ischial tuberosity and the lateral epicondyle of the tibia, and ERE electrode 3 cm lateral to L1. *Figure 3.1* shows the placement of the electrodes. A reference electrode was placed over the lateral aspect of the fibular malleolus. Signals from the electrodes were amplified (1,000×) and band-pass filtered (6–500 Hz). All signals were sampled at 1,200 Hz with a 12-bit resolution.

Experimental Procedures. All participants performed one control task, followed by ten trials of an experimental task.

The **control task** consisted of an unperturbed upright bipedal stance (*UStance*). Participants were instructed to stand barefoot on the force platform with their feet parallel and 15 cm apart, their upper limbs crossed against their chest, and their vision focused to a physical static point placed at eye's height and at a distance of 2 m. Once this initial position was adopted, participants were asked to remain steady for 35 s keeping their body as vertical as possible, and distribute their body weight evenly between the two feet (*Figure 3.2A*). Data collection started after 5 s and lasted for 30 s.

The **experimental task** consisted of upright stance under the continuous perturbation of holding a load (*LOAD*). The participants remained on the force platform in the same position previously described. In addition, they were asked to hold a barbell (5 kg) in front of their body with their shoulders flexed 90$^\text{o}$ and their elbows fully extended (*Figure 3.2B*). The participants held the weight by pressing on its two circular ends, and a total of ten valid trials were performed. A trial was considered valid only when participants kept their body vertically steady during the entire length of the trial while keeping a vertical orientation of the major body segments (lower limbs, trunk and head). To avoid recording any transient effects during the initial moments of a trial, data

recordings started only after 5 s from the moment the participant took position. Each trial was then recorded for 10 s. When a trial was considered invalid, recordings were discarded and the trial was repeated at the end of the experiment. To avoid fatigue, a time interval of 60 s was provided between trials, when participants could be at ease but had to keep their feet at the same position on the force platform.

The average duration of the entire experimental session was 30 min, and none of the participants reported fatigue.
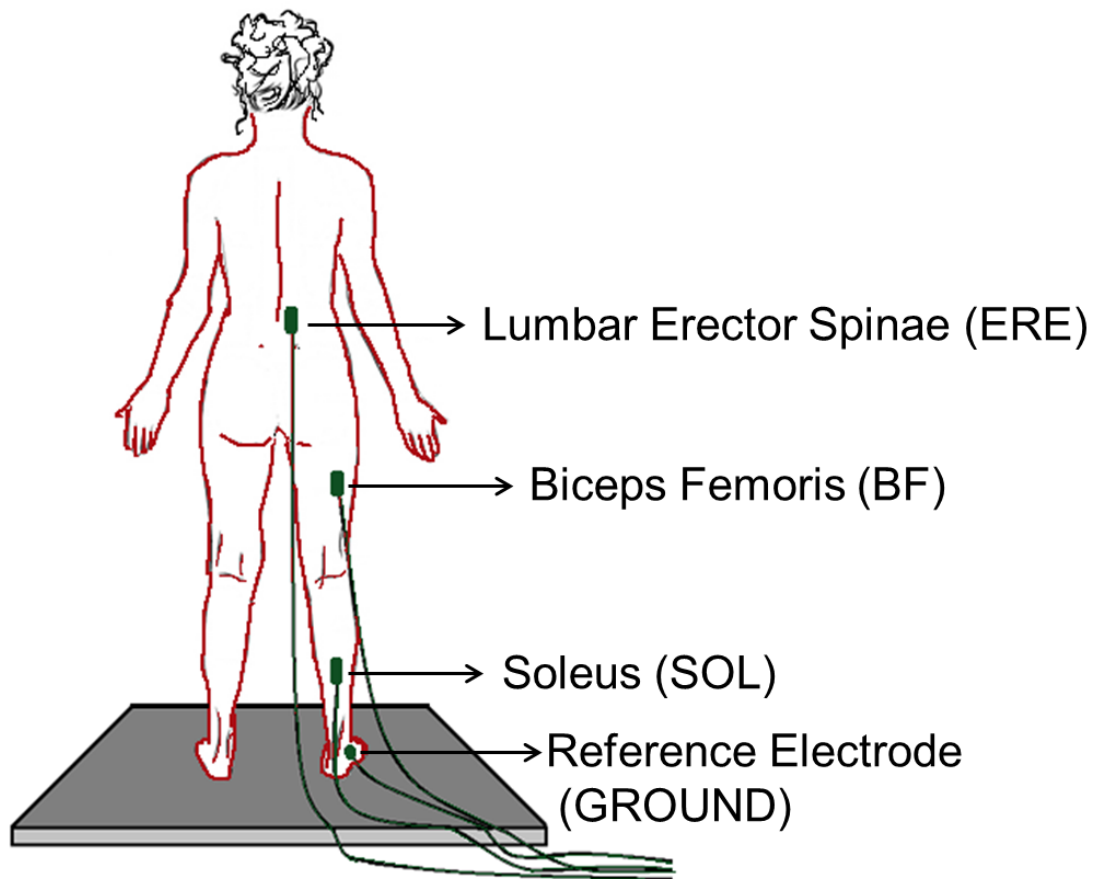
Figure 3.1. Representation of the electrodes placed on postural muscles. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), and reference electrode (GROUND).
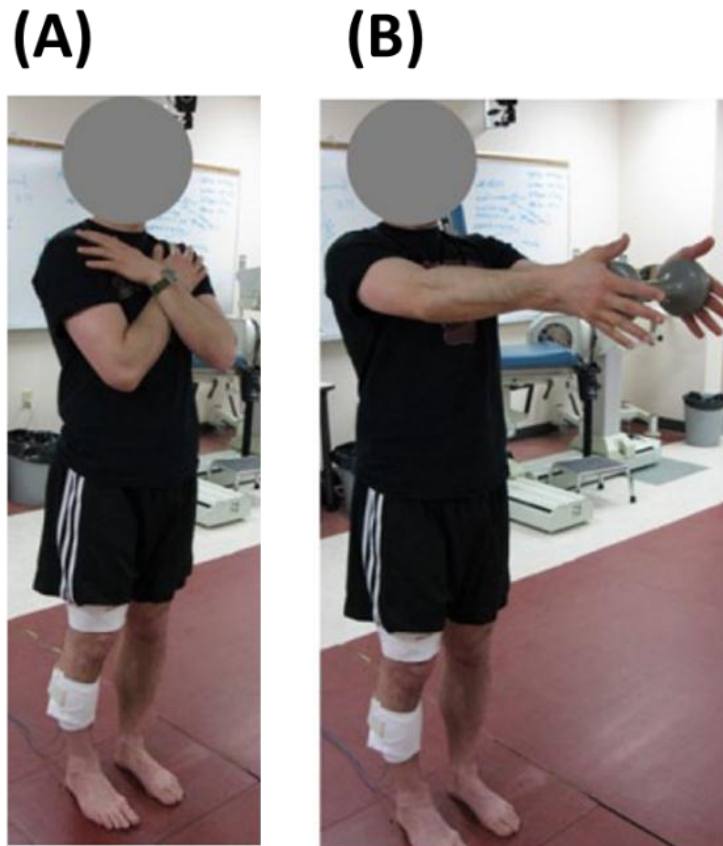
Figure 3.2. Posture adopted during **(A)** the control task (unperturbed stance, *UStance*), and **(B)** the experimental task (holding a load, *LOAD*).

3.3. DATA PROCESSING

Center of pressure coordinates and EMG signals were analyzed off-line with custom-written software routines (Matlab R2012b, The MathWorks).

Postural sway behavior. COPap and COPml coordinate signals were filtered with a 20 Hz low-pass, second-order, and zero lag Butterworth filter. Next, the relative position of COPap and COPml during each *LOAD* trial was determined with respect to the normalized position of these same coordinates obtained during the *UStance* trial. Normalization of coordinates recorded during the *UStance* trial was obtained by subtracting their average position from its initial values. This normalization allowed the

researcher to position COP coordinates at the center of an xy coordinate system and, therefore, draw any comparisons of basic COP features across participants. In a sequence, the following postural indices were extracted: the elliptical area containing 95 % of the entire COP path (*Area*); the ranges and mean velocities of the COP displacement in both anterior-posterior and medio-lateral directions (*Range$_{AP}$, Range$_{ML}$, MV$_{AP}$*, and *MV$_{ML}$*, respectively); and the mean power frequency and the maximum frequency containing 80% of the power spectral density of the *COPap* signal (*Fmean$_{AP}$* and *F80$_{AP}$*, respectively). Measures of frequency were only extracted from the anterior-posterior component of the COP because the muscles recorded act mostly to move the body's COP in this direction. A more detailed description of these postural indices can be found in *Chapter 7* of the dissertation.

Levels and patterns of muscle activation. The EMG signals recorded from all three muscles were submitted to two separate analyses.

First, the relative index of muscle activation (*Index$_{EMG}$*) was computed to quantify the **amplitude of muscle activation** of each muscle during the experimental trials (*LOAD*). EMG signals from both control and experimental trials were visually inspected to verify the presence of any signal artifacts and then filtered (20 Hz high-pass, second-order, zero-lag Butterworth filter) and full-wave rectified. Each resulting signal from the *LOAD* trials was integrated over its entire trial length and normalized by similar integrals performed on the EMG signals recorded during the control trial (*UStance*). Since the duration of the *UStance* trial was 30 s, only the intermediate 10 s of the trials were integrated. *Index$_{EMG}$* was used to confirm that all muscles recorded were activated during the execution of the experimental task. *Figure 3.3* shows the sequence of steps used to compute *Index$_{EMG}$*. This index indicates the relative amount of EMG activation for a given muscle during the *LOAD* trial, compared to the *UStance* trial. For example, a final *Index$_{EMG}$* of value 2 indicates that the muscle activation during the *LOAD* trial was twice the muscle activation during the *UStance* trial.
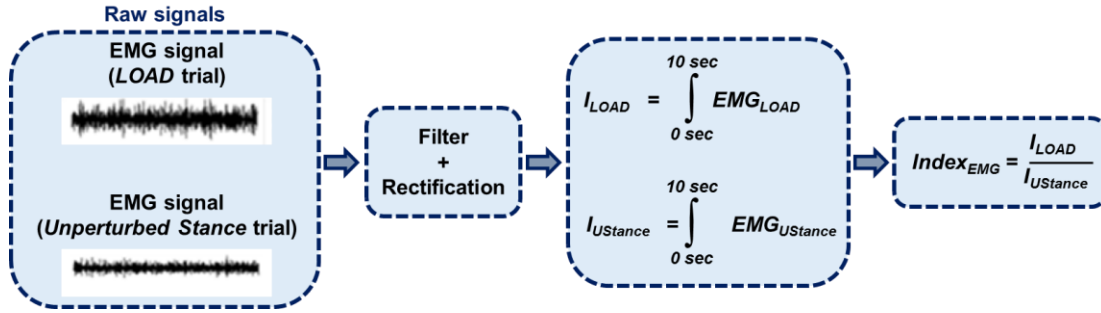
Figure 3.3. The sequence of steps used to compute the relative index of muscle activation (*Index_EMG*).

In addition, the **degree of similarity of muscle activation patterns** during the experimental (*LOAD*) trial across participants was computed based on the resultant muscle activation vectors of the participants. The resultant muscle activation vector of each participant was computed as the resultant vector formed by three muscle activation vectors: $\overrightarrow{Index_{EMG\_SOL}}$, $\overrightarrow{Index_{EMG\_BF}}$, and $\overrightarrow{Index_{EMG\_ERE}}$. These three muscle activation vectors were orthogonally assembled and their magnitude was the respective normalized *Index_EMG*. *Figure 3.4AB* illustrates the resultant muscle activation vectors of two representative participants. Next, the cosine of the angle between resultant vectors of every two participants was computed (*Figure 3.4C*). The cosine values, ranging from 0 to 1, were used to quantify the degree of similarity of the muscle activation pattern between two participants. Cosine values close to 0 indicates perpendicularity or dissimilarity of the patterns of muscle activation across participants, whereas cosine values close to 1 indicate parallelism or similarity in the patterns of muscle activation across participants (see Danna-dos-Santos et al 2007 and Poston et al 2010 for more details for the computation of the cosines). This analysis was important to detect whether participants were using similar patterns of muscle activation throughout their experimental trials.

A) Resultant muscle activation vector (participant # 5)

B) Resultant muscle activation vector (participant # 9)

C) Angle, and respective cosine, between two resultant muscle activation vectors

Angle = 13.17°
Cosine = 0.97
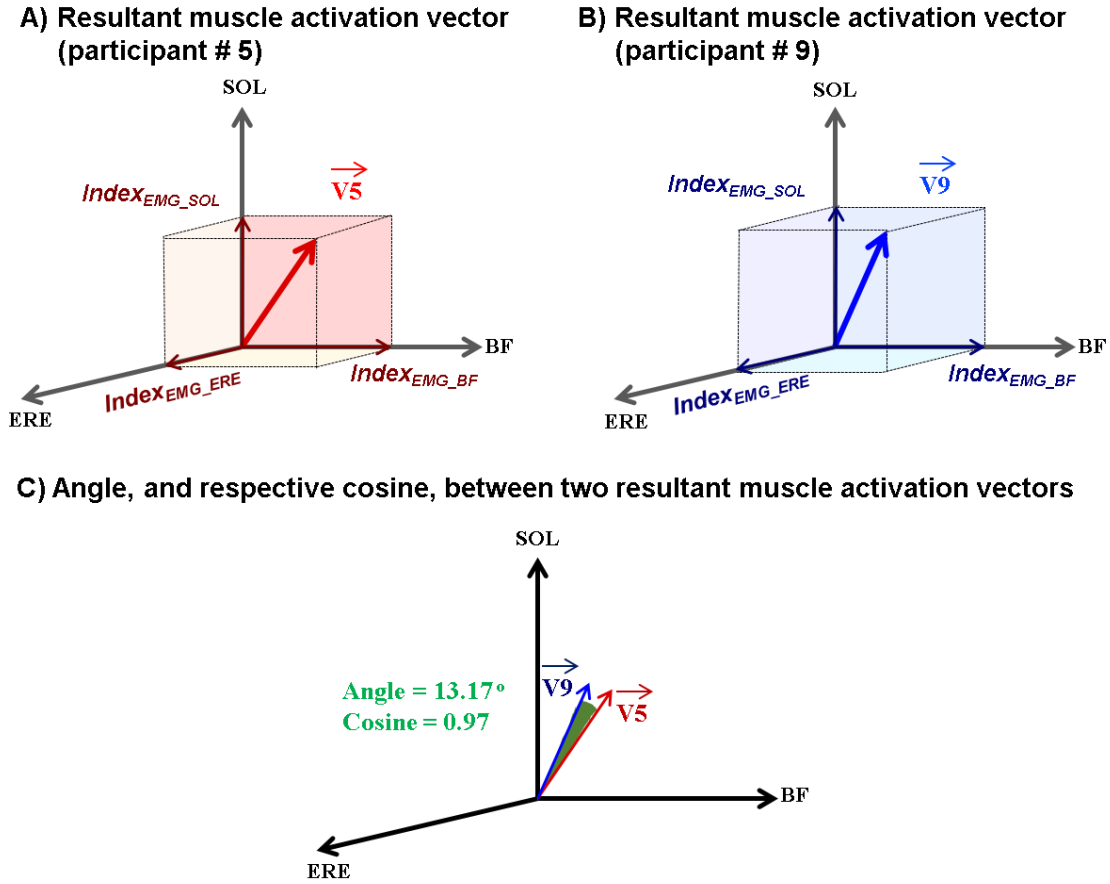
Figure 3.4. Resultant muscle activation vectors (*V5* and *V9*) of two representative participants performing the experimental trial (panels *A* and *B*, respectively). The angle, and respective cosine, between the resultant muscle activation vectors of these two participants (panel *C*).

The second analysis quantified the correlation of muscle activation in the frequency domain during the experimental trials (*LOAD*) by using **intermuscular coherence analyses**. Filtered EMG signals (20 Hz high-pass, second-order, zero lag Butterworth filter) recorded from all ten experimental trials were concatenated to create a long single data series (100 s; 120,000 data points). This procedure (concatenating) was used to increase the number of segments and the reliability of the coherence estimation (Maris et al 2007, Poston et al 2010). Intermuscular coherence was then estimated separately for each pair of EMG signals (SOL/BF, SOL/ERE, and BF/ERE) by using the cross-spectrum of two EMG signals (f_{xy}) squared and normalized by the product of the autospectrum of each signal (f_{xx} and f_{yy}) at each frequency (λ), as follows:

$$\left|R_{xy}(\lambda)\right|^2 = \frac{\left|f_{xy}(\lambda)\right|^2}{\left|f_{xx}(\lambda)f_{yy}(\lambda)\right|} \tag{3.3}$$

Coherence was estimated from segments of 1 s duration (i.e., 1,200 samples per segment), resulting in a frequency resolution of 1 Hz. Since the synchronization at lower frequency bands was of interest of the researcher, coherence was analyzed within 0 to 55 Hz (a more detailed description can be found in Danna-dos-Santos et al 2010 and Poston et al 2010). In order to compare coherence across participants and frequency intervals, all estimates were z-transformed using the Fisher transformation (Rosenberg et al 1989, Amjad et al 1997, Danna-dos-Santos et al 2010). Coherence was considered statistically significant when it exceeded the significance limit of the null distribution, computed as proposed by Rosenberg et al (1989). Analysis of the *frequency distribution* of correlated neural inputs was performed by identifying frequency intervals with significant z-scored coherence estimates across participants. Next, the analysis of the *strength of correlated neural inputs* was achieved by comparing integrals computed for the z-scored coherence estimates profiles over the frequency bands of interest. For further comparisons of these integrals among different frequency bands, the integrals were normalized by the length of each interval. *Figure 3.5* shows an example of a typical z-scored coherence profile within the frequency band 0–55 Hz and the significance level obtained from a pair of EMG signals (SOL/BF) of a representative *LOAD* trial. Note the significant z-scored coherence estimates for the frequencies 0 to 4 Hz, 10 to 15 Hz, and 33 to 35 Hz. Note also that the SOL/BF coherence seems stronger within the frequency band 0–4 Hz, compared to that of the other two frequency bands.
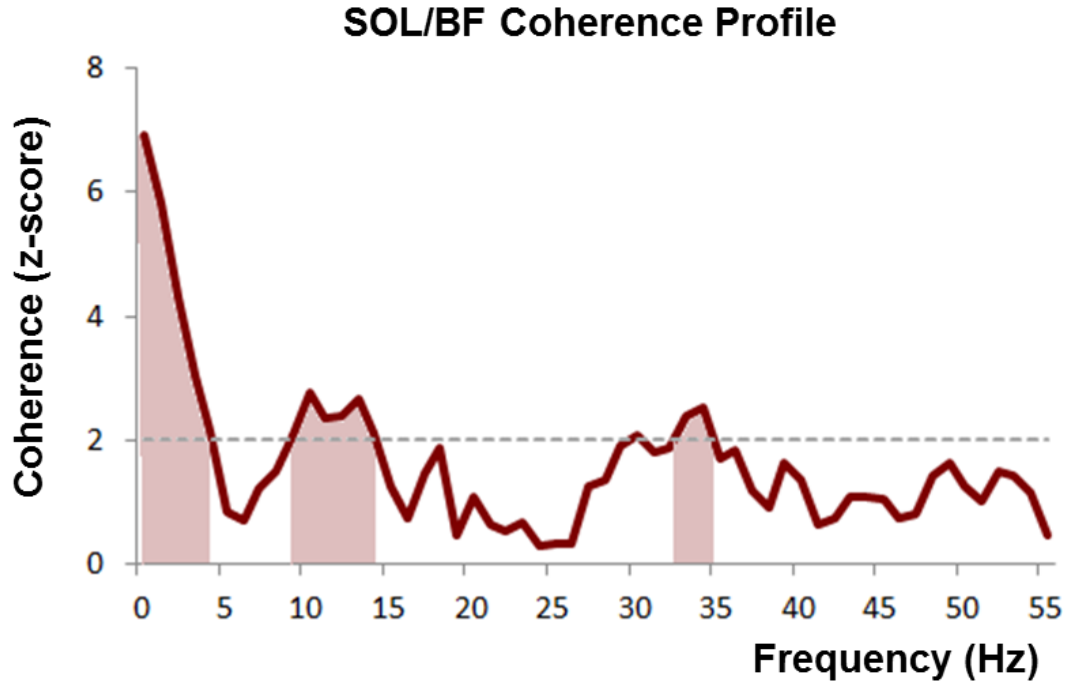
Figure 3.5. Z-scored coherence profile (and significance level, *dashed* line) obtained from a single pair of EMG signals (SOL/BF) computed from a representative *LOAD* trial.

In addition to the intermuscular coherence for pairs of muscles, estimates of coherence obtained for each muscle pair were combined to calculate the pooled coherence estimates across muscle pairs, as proposed by Amjad et al (1997) as follows:

$$\frac{\left| \sum_{i=1}^{k} f_{xy}(\lambda) L_i \right|^2}{\left(\left(\sum_{i=1}^{k} f_{xx}(\lambda) L_i \right) \left(\sum_{i=1}^{k} f_{yy}(\lambda) L_i \right) \right)} \tag{3.4}$$

Pooled coherence estimates are considered a weighted average of individual coherence estimates. They are used to increase statistical power of the estimate. It is assumed that all three muscles form a single muscle mode and, hence, share correlated neural inputs. As coherence between the different muscle combinations reflects the same underlying process, the estimate is improved by pooling coherence across all three pairs

(SOL/BF, SOL/ERE, and BF/ERE). Analyses of the frequency distribution and strength of correlated neural inputs obtained from pooled coherence calculations followed the same procedures as described for single-pair analyses.

Statistical approach. Summaries on the averages and standard deviations are reported along with the text and the tables. Considering the sample size, Shapiro-Wilk tests were performed to verify whether responses were normally distributed. Since the response variables followed a normal distribution, a series of parametric tests were performed. Paired *t*-tests were conducted on the factor *Task (UStance* and *LOAD*) for response variables extracted from the COP signals and for the relative index of muscle activation (*Index$_{EMG}$*). For the *UStance* task, repeated measures ANOVAs followed by post hoc comparisons (paired *t*-tests) were performed on the factors *Frequency Band* and *Muscle Pair* for the variable *Normalized Integrals* (*INT$_{Coh}$*). All parametric tests were conducted by the IBM SPSS statistics software suite (version 20, IBM® SPSS®) and the level of significance was fixed at 5% ($\alpha = 0.05$) for individual tests.

3.4. RESULTS

All nine participants were able to perform both the *UStance* and *LOAD* trials with ease and without any signs of fatigue or discomfort. In addition, all participants were able to easily sustain a vertical position of their axial skeleton during the *LOAD* trials with no visible postural deviations.

Postural sway behavior. In general, the application of the frontal weight to the axial skeleton generated qualitative and quantitative changes on the COP displacement patterns and its frequency content. As expected for the *LOAD* trials, there was an anterior migration of the body's COP (anterior mean displacement = 2.94 cm, *SD* = 2.15). This anterior migration was associated with a small migration to the left (mean displacement to the left = 0.25 cm, *SD* = 0.42). *Figure 3.6* shows the average COP position for each

participant during *LOAD* trials (*open circles*), as well as the averaged COP position during *UStance* (*filled triangle*) and *LOAD* trials (*filled circle*) across participants.
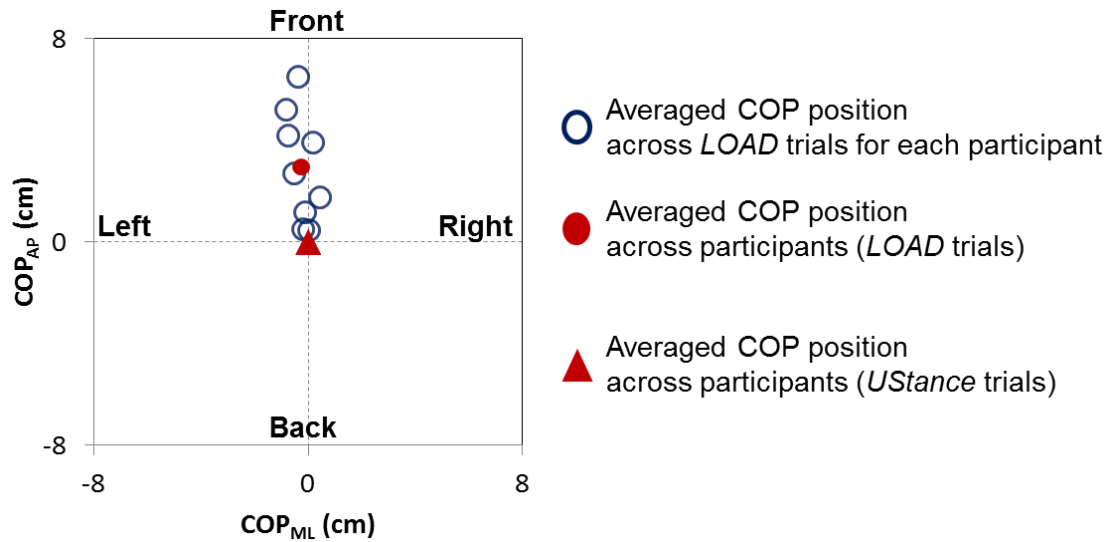
Figure 3.6. The average center of pressure (*COP*) position across trials for each participant during *LOAD* trials (*open circles*), the average COP position across participants during *LOAD* trials (*filled circle*), and the average COP position across participants during *UStance* trials (*filled triangle*).

In addition to this migration, all participants oscillated more and faster when the frontal load was applied. *Figure 3.7* shows the COPap and COPml profiles of a representative participant performing the *UStance* (panels *A* and *B*) and one representative *LOAD* trial (panels *C* and *D*). Note the larger area and amplitude of the COP displacement during the *LOAD* trial. Also visually noticeable is the increased presence of higher frequency oscillations embedded on both COPap and COPml profiles under *LOAD* task (panel *D*) compared to that for the *UStance* task (panel *B*).
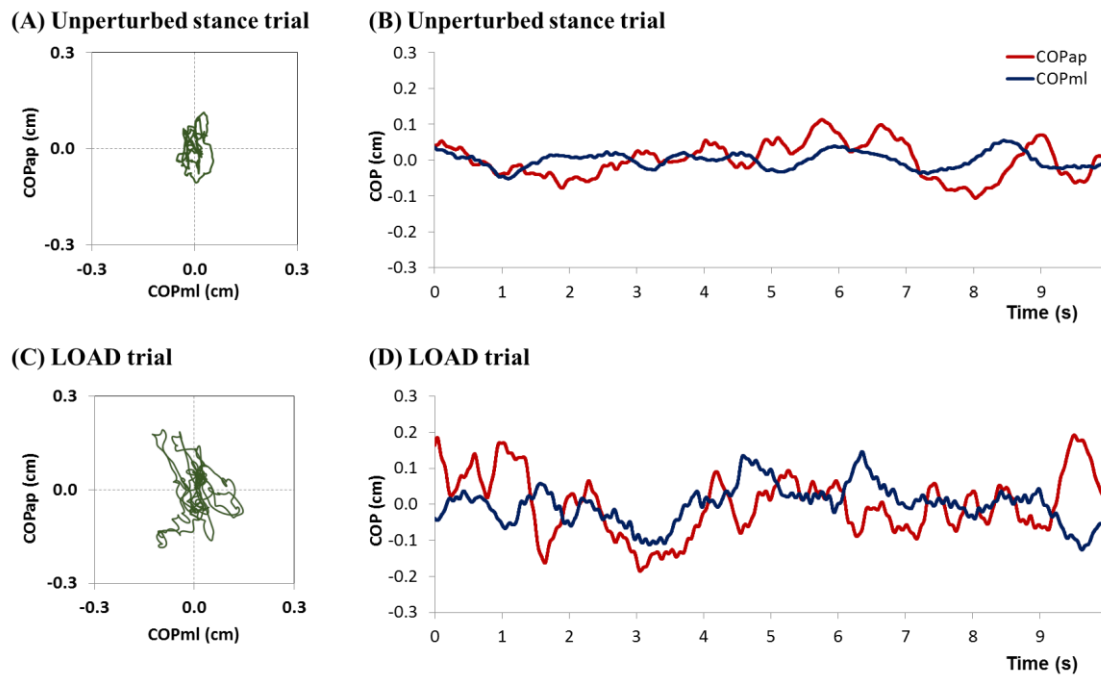
Figure 3.7. Center of pressure (*COP*) displacement of one representative *UStance* trial (panels *A* and *B*) and one representative *LOAD* trial (panels *C* and *D*).

Table 3.1 summarizes the averages across participants and the respective standard deviations for all seven COP variables (*Area*, *Range*$_{AP}$, *Range*$_{ML}$, MV_{AP}, MV_{ML}, *Fmean*$_{AP}$, and *F80*$_{AP}$). In general, participants presented a larger, faster, and increased frequency in their body sway when holding a frontal weight, compared to unperturbed stance. Paired *t*-tests confirmed a significant increase in *Area*, *Range*$_{AP}$, MV_{AP}, MV_{ML}, *Fmean*$_{AP}$, and *F80*$_{AP}$ for the *LOAD* task, compared to that for the *UStance* task (see *p* values in *Table 3.1*).

Table 3.1. Average, standard deviation, and 95% confidence interval of the difference across participants of seven postural indices extracted from the center of pressure coordinates in the anterior–posterior and medio–lateral directions (*COPap* and *COPml*, respectively) during unperturbed stance (*UStance*) and holding a load (*LOAD*) tasks. Note: * indicates a significant effect of *Task* ($p < 0.05$).

	UStance	*LOAD*	95% confidence interval of the difference	Paired *t*-test
Area (cm^2)	0.46 ± 0.20	0.94 ± 0.60	(-0.90, -0.06)	$t_{[8]} = -2.625$ $p = 0.030$ *
Range$_{AP}$ (cm)	1.07 ± 0.37	1.34 ± 0.50	(-0.54, -0.01)	$t_{[8]} = -2.371$ $p = 0.045$ *
Range$_{ML}$ (cm)	0.55 ± 0.22	0.69 ± 0.38	(-0.39, 0.12)	$t_{[8]} = -1.243$ $p = 0.249$
MV$_{AP}$ (cm/s)	0.69 ± 0.21	1.17 ± 0.45	(-0.75, -0.20)	$t_{[8]} = -3.931$ $p = 0.004$ *
MV$_{ML}$ (cm/s)	0.44 ± 0.17	0.67 ± 0.34	(-0.45, -0.01)	$t_{[8]} = -2.373$ $p = 0.045$ *
Fmean$_{AP}$ (Hz)	0.51 ± 0.12	0.68 ± 0.15	(-0.30, -0.06)	$t_{[8]} = -3.421$ $p = 0.009$ *
F80$_{AP}$ (Hz)	0.82 ± 0.21	1.05 ± 0.23	(-0.44, -0.02)	$t_{[8]} = -2.495$ $p = 0.037$ *

Levels of muscle activation. All participants revealed a relative increase in activation levels of all muscles during the execution of the *LOAD* trials. *Figure 3.8* shows the raw EMG recordings for all 3 muscles obtained from a representative participant during the execution of *UStance* and *LOAD* trials (panel *A*), and the index of muscle activation (*Index*$_{EMG}$) of each muscle (panel *B*). Note the larger amplitude of the EMG signals for all three muscles during the *LOAD* trial. Note also that the lumbar erector spinae (ERE) muscle had the largest mean *Index*$_{EMG}$ value across participants (*mean* = 2.52, *SD* = 1.10) compared to that for the muscles BF and SOL (*mean* = 1.81, *SD* = 0.39 and *mean* = 1.64, *SD* = 0.38, respectively). *Table 3.2* shows the average and standard

deviation across participants for the index of muscle activation (*Index_EMG*). A two tailed paired *t*-test comparing *Index_EMG* across participants to a vector of ones representing similar level of muscle activation was performed for each muscle. These tests confirmed an increased muscle activation during the *LOAD* trials ($t_{[8]} = 4.994$, $p = 0.001$; $t_{[8]} = 6.159$, $p < 0.001$; and $t_{[8]} = 4.127$, $p = 0.003$; respectively for SOL, BF, and ERE muscles).
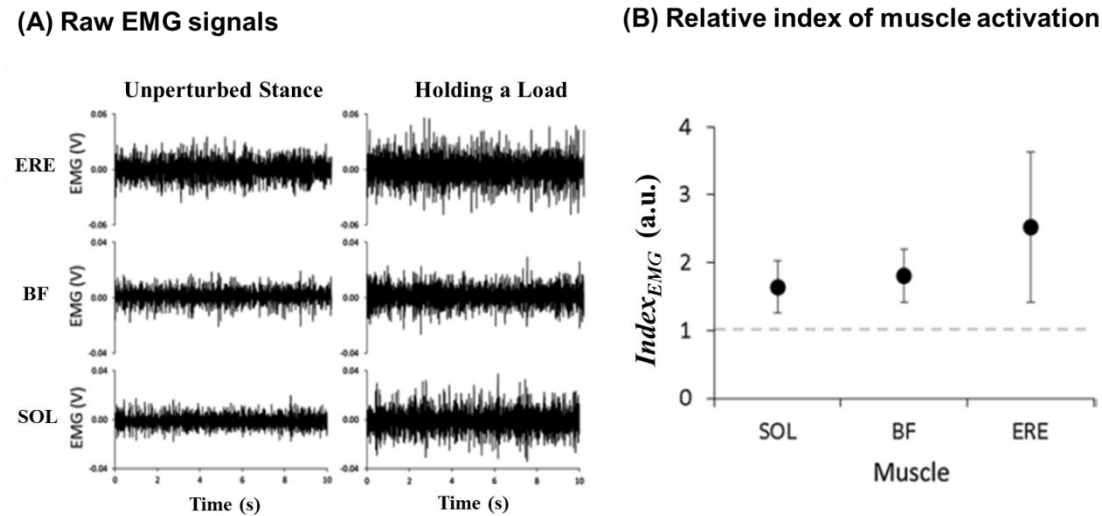
Figure 3.8. (A) Raw EMG signals recorded from the soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE) muscles in a representative participant performing the unperturbed stance trial (*UStance*) and holding an anterior load trial (*LOAD*). **(B)** The averages and standard deviations of the relative index of muscle activation (*Index_EMG*) across participants. *Dashed line* indicates the region of the graph where the relative levels of muscle activation would be equal between *UStance* and *LOAD* trials.

Table 3.2. The average and standard deviation across participants for the index of muscle activation (*Index$_{EMG}$*) for soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE).

	SOL	**BF**	**ERE**
Index$_{EMG}$	1.64 ± 0.38	1.81 ± 0.39	2.52 ± 1.10

Patterns of muscle activation. Participants employed similar patterns of muscle activation during the execution of *LOAD* trials. This similarity of patterns was confirmed by the vector analysis revealing cosine values across participants close to value 1 (*mean = 0.94, SD = 0.04*).

Frequency domain analysis. *Figure 3.9ABC* shows the power spectrum density (PSD) of EMG signals from SOL, BF, and ERE of a representative participant performing a *LOAD* trial. Muscle activation in the frequency domain was analyzed to compute intermuscular coherence. *Figure 3.9D* shows the pooled coherence computed at the lower frequency band of 0–55 Hz. As observed, there was sign of intermuscular coherence within 0–3 Hz.

(A) Soleus

(B) Biceps Femoris

(D) Pooled Coherence within 0-55 Hz

(C) Lumbar Erector Spinae
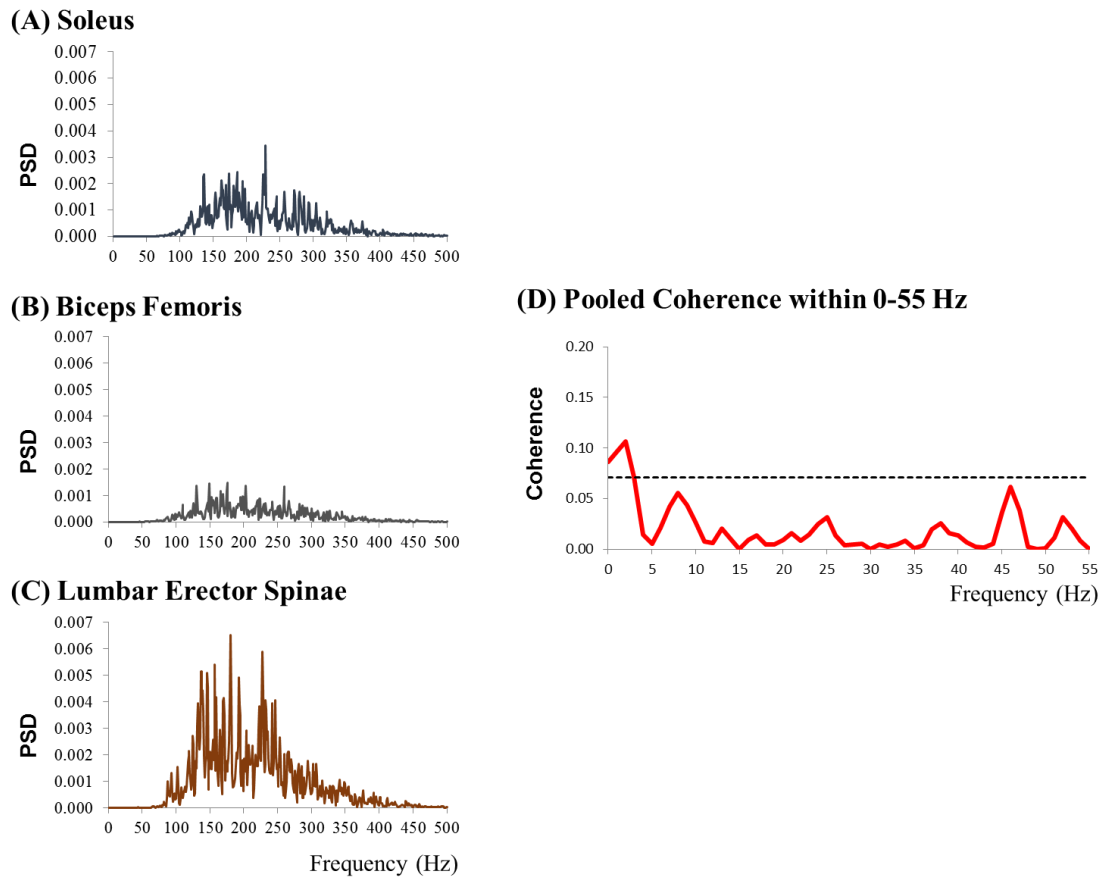
Figure 3.9. The power spectrum density (PSD) of EMG signals from soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE) muscles of a representative participant performing a *LOAD* trial (panels *A* to *C*); and the pooled coherence profile (and significant limit in *dashed line*) within 0–55Hz (panels *D*).

Pooled coherence. *Figure 3.10A* shows the average and standard deviation across participants for the pooled z-scored coherence profiles and the significance limit (*black dashed line*) over the frequency range of 0–55 Hz during the *LOAD* trials. The overall profile was relatively constant across participants, and two frequency intervals were consistently found to reach significant values of coherence: 0–5 Hz and 5–20 Hz.

Figure 3.10B and *Table 3.3* show the average (and standard deviation) across participants for the normalized integrals of these profiles under these two frequency intervals 0–5 Hz and 5–20 Hz (2.51 ± 0.93 and 2.34 ± 0.60, respectively) and under the frequency interval containing the remaining frequency spectrum studied 20–55 Hz (1.52 ± 0.18). As a reminder, these integrals were normalized by the length of each frequency interval mentioned in order to perform comparison of their values. A repeated measures

ANOVA confirmed the effect of *Frequency Band* on the variable *Normalized Integral* (Wilks' Lambda = 0.346, $F_{(2,7)}$ = 6.630, p = 0.024). Paired *t*-tests were used to make post hoc comparisons between frequency bands and revealed stronger pooled coherence within 0–5 Hz and 5–20 Hz, compared to that for the frequency band 20–55 Hz ($t_{[8]}$ = 3.025, p = 0.016; and $t_{[8]}$ = 3.888, p = 0.005; respectively).

(A) Pooled coherence (SOL/BF/ERE)   **(B) Coherence strength**
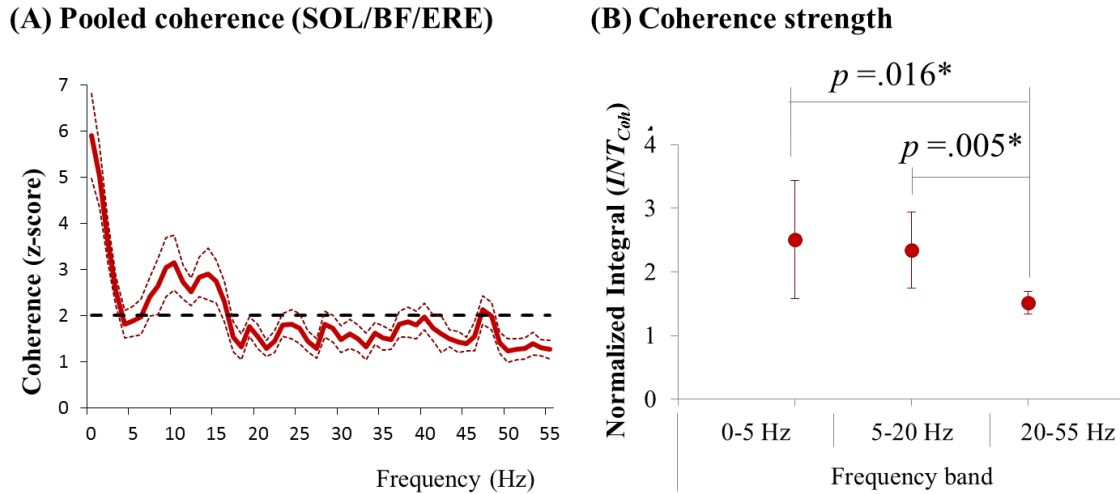
Figure 3.10. (A) Average and standard deviation across participants of the z-scored pooled coherence profile during *LOAD* trials. Note the significance level in *dashed line.* **(B)** Averages and standard deviations across participants for the normalized integrals of the z-scored pooled coherence over the frequency intervals of 0–5 Hz, 5–20 Hz, and 20–55 Hz. Note: * indicates a significant effect of *frequency band* ($p < 0.05$).

Table 3.3. Averages and standard deviations across participants for the normalized integral of the z-scored coherence within the frequency bands 0–5 Hz, 5–20 Hz, and 20–55 Hz ($INT_{Coh0-5Hz}$, $INT_{Coh5-20Hz}$ and $INT_{Coh20-55Hz}$, *respectively*) during *LOAD* trials.

	$INT_{Coh0-5Hz}$	$INT_{Coh5-20Hz}$	$INT_{Coh20-55Hz}$
Pooled coherence (SOL/BF, SOL/ERE, BF/ERE)	2.51 ± 0.93	2.35 ± 0.60	1.52 ± 0.18
Single-pair SOL/BF	2.75 ± 0.67	1.9 ± 0.43	1.41 ± 0.25
Single-pair SOL/ERE	1.38 ± 0.65	1.58 ± 0.32	1.25 ± 0.16
Single-pair BF/ERE	1.39 ± 0.38	1.63 ± 0.32	1.21 ± 0.17

Single-pair coherence. *Figure 3.11A* shows the averages across participants for the z-scored coherence profiles computed separately for each pair of muscles formed between SOL, BF, and ERE over the frequency range of 0–55 Hz. The significance limit (*dashed line*) is also shown. Similar to pooled coherence results, the overall profile obtained for single-pair analyses was kept relatively constant across the participants with two frequency bands reaching levels of significance: 0–5 Hz and 5–20 Hz. *Table 3.3* shows the normalized integral of the z-scored coherence of muscle groups over different frequency bands (0–5 Hz, 5–20 Hz, and 20–55 Hz).

Figure 3.11B shows the averages and standard deviations across participants for the normalized integral computed over the entire 0–55 Hz spectrum for all three muscle pairs. Note that SOL/BF pair had the largest normalized integral value (1.77 ± 0.29), followed by SOL/ERE and BF/ERE pairs (1.41 ± 0.16 and 1.39 ± 0.13, respectively). A repeated measures ANOVA confirmed the effect of *Muscle Pair* on the variable *Normalized Integral* within 0–55 Hz (Wilks' Lambda = 0.371, $F_{(2,7)} = 5.931$, $p = 0.031$). Post hoc comparisons between muscle pairs (paired *t*-tests) revealed stronger coherence within 0–55 Hz for the muscle pair SOL/BF, compared to that for SOL/ERE and BF/ERE ($t_{[8]} = 3.131$, $p = 0.014$; and $t_{[8]} = 3.681$, $p = 0.006$; respectively).

Even though the coherence strength within the frequency band 0–55 Hz for the muscle pair SOL/BF had been found to be significantly higher in comparison to the other two muscle pairs, this increase was entirely related to increased values of coherence found within the frequency interval of 0–5 Hz. *Figure 3.11C* shows the averages and standard deviations across participants for the variable normalized integral computed over the frequency bands 0–5 Hz, 5–20 Hz, and 20–55 Hz. Repeated measures ANOVAs found an effect of *Muscle Pair* on the variable *Normalized Integral* for the frequency band 0–5 Hz (Wilks' Lambda = 0.191, $F_{(2,7)} = 14.800$, $p = 0.003$), and no effect within either 5–20 Hz or 20–55 Hz (Wilks' Lambda = 0.599, $F_{(2,7)} = 2.339$, $p = 0.167$; and Wilks' Lambda = 0.565, $F_{(2,7)} = 2.692$, $p = 0.136$; respectively). Paired *t*-tests showed stronger coherence for the muscle pair SOL/BF compared to that for the pairs SOL/ERE and BF/ERE within 0–5 Hz ($t_{[8]} = 4.794$, $p = 0.001$; and $t_{[8]} = 5.539$, $p = 0.001$; respectively). In addition, repeated measures ANOVAs found an effect of *Frequency Band* on the variable *Normalized Integral* only for the muscle pair SOL/BF (Wilks'

Lambda = 0.124, $F_{(2,7)}$ = 24.634, p = 0.001). Post hoc comparisons between frequency bands (paired *t*-tests) revealed stronger SOL/BF coherence within 0–5 Hz compared to the frequency bands 5–20 and 20–55 Hz ($t_{[8]}$ = -4.966, p = 0.001; and $t_{[8]}$ = -3.626, p = 0.007; respectively) and within 5–20 Hz compared to that within 20–55 Hz ($t_{[8]}$ = 3.419, p = 0.009).

Figure 3.11. **(A)** Averages across participants of the z-scored single-pair coherence profiles obtained for each muscle pair (SOL/BF, SOL/ERE, and BF/ERE). **(B)** Averages and standard deviations across participants of the normalized integrals computed for each muscle pair over the entire frequency spectrum investigated (0–55 Hz). **(C)** Averages and standard deviations across participants of the normalized integrals computed for each muscle pair over three different frequency intervals (0–5 Hz, 5–20 Hz, and 20–55 Hz). Note: * indicates a significant effect of *frequency band* and + represents a significant effect of *muscle pair* ($p < 0.05$).

3.5.   DISCUSSION

The present study investigated the intermuscular coherence as a potential mechanism underlying the organization of multi-muscle synergies. Consistent with the hypothesis of this study, correlated neural inputs were found in three posterior postural muscles (SOL, BF, and ERE) during bipedal stance under the continuous perturbation task of holding a load in front of the body. Specifically, an examination of the frequency bands, at which coherent muscle activity happened, revealed significant coherence within the frequency bands 0–5 Hz and 5–20 Hz. In addition, the results confirm significant correlated inputs in all postural muscles forming a single M-mode, even though these muscles are anatomically remote and act on different joints. Taken together, these findings corroborate the notion that the CNS not only unites motor effectors into smaller sets of controllable units, but may also use correlated neural inputs to assemble these modal (functional) units. In the remainder of this section, findings and their implications on the multi-muscle control of upright stance are discussed.

M-modes and synergies. When any action, static or dynamic, is performed in bipedal stance, the coordination of a large number of muscles is required to counteract the effects of external and internal forces applied onto major body segments and to avoid the collapse of the body's vertical alignment. In terms of control design, this means controlling the position of multiple heavy segments (such as the head, trunk, thighs, and legs) vertically on top of each other and over a small base of support. This mechanical challenge is broadened by the increased complexity of the anatomical distribution of muscles along this vertical axis, where many muscles will cross multiple joints.

To overcome some of these challenges, the proposition of a hierarchical system of control where the controller reduces the number of variables it manipulates was adopted (Gelfand and Tsetlin 1966, Bernstein 1967, Gelfand and Latash 2002). According to this view, at a lower level of the hierarchy, redundant muscles with similar functions are united into smaller groups (M-modes); whereas at a higher level, gains at the M-Modes are co-varied by the controller to ensure stability of the properties of important mechanical variables, such as the body's COP. One may view M-modes as "virtual

muscles" manipulated at a higher level of the control hierarchy for the control of relevant task variables. Under this view, movement control becomes more efficient by freeing neuronal computational power and, therefore, allowing the controller to manage the execution of concurrent secondary tasks, including those related to responses to unexpected perturbations (principle of abundance reviewed in Gelfand and Latash 2002).

The present study was specifically interested in possible neural mechanisms involved in the formation of a previously recognized M-mode. Hypothetically, different sets of M-Modes can be created within the same space of muscle activations, and their formation depends mostly on the nature of the planned action (Danna-dos-Santos et al 2008). For example, the same leg or trunk muscles can be united into sets of optimal M-modes to ensure stabilization of a certain relevant performance variable (M-modes assembled for the execution of a sit-to-stand task will likely to be different from those M-modes assembled during walking). In situations where investigations focused on upright standing, similar M-mode compositions have been reported and commonly include the emergence of three main M-modes: a posterior M-mode formed by soleus, biceps femoris, semitendinosus, lumbar erector spinae, and gastrocnemius; an anterior M-mode formed by tibialis anterior, vastus medialis, vastus lateralis, and rectus femoris; and an M-mode often formed solely by the rectus abdominis muscle (Krishnamoorthy et al 2003a, Wang et al 2005,2006, Danna-dos-Santos et al 2007). These previous studies have provided support for the existence of a harmonic co-variation of M-modes modulation as a way to control the stability of the body's COP along the execution of standing tasks.

The present study focused on the posterior M-mode based on the facility of inducing an isometric contraction of all its components by simply adding a weight to the body's anterior aspect. Therefore, no major task complications were introduced allowing all participants to perform the task with minimum constraints. However, the addition of the load disregarding variation on participant's anthropometrics ensured that all participants used similar patterns of muscle activation to maintain upright posture.

In order to further understand the possible neural mechanisms underlying the coordination of muscles forming the posterior M-mode, intermuscular coherence was computed across all muscle pair combinations formed among the SOL, BF, and ERE. The strength of the coherence pooled at lower frequency bands across all muscles pairs

was significant for two frequency intervals (0–5 Hz and 5–20 Hz). Synchronizations within 0–5 Hz have been reported by previous studies and are thought to reflect synchronized contraction patterns of multiple muscles related to slow movements (Karmen and De Luca 1992, Farmer et al 1993, Vallbo and Wessberg 1993, De Luca and Erim 2002). According to these same studies, the neural sites responsible for the generation of such synchronizations at lower frequency bands are still unclear and unlikely to originate within the corticospinal system (Grosse et al 2002, Mochizuki et al 2006).

It is important to emphasize that the peak value found for the pooled coherence profiles within this frequency range was below 1 Hz (*Figure 3.10A*) and, therefore, close to the frequency of the COP oscillations (0.68 H z) obtained during the execution of the experimental task. The correlation between the oscillation of the COP and the modulation of the recorded EMG signals was not computed, but Mochizuki et al (2006) have previously demonstrated such a relationship. More specifically, their results suggest that very low range synchronizations (<1 H z) are likely to be the result of the coupling between oscillations of the COP and the EMG signals. This view is corroborated by the results shown for the analysis of coherence profiles obtained for each muscle pair separately (*Figure 3.11A*), in which only the most distal muscle pair exhibited such synchronizations. This difference of strength among the three pairs was attributed to a possible prevalence of the ankle strategy to counteract the body oscillations observed. Under this scenario (ankle strategy), most of the anterior body displacement is driven by angular motion around the ankle joint generated by the distal lower posterior muscles. Note that the mean velocity and ranges of displacement of the body's COP exhibited by the participants were relatively small and unlikely to elicit other strategies, such as the hip strategy (Nashner and Horak 1986). This view is supported by Gatev et al (1999), who examined correlations between postural muscle activity and sway events during unperturbed stance. These researchers also reported a significant positive correlation between the EMG activity of the lateral gastrocnemius muscle and the displacement of the body's COP in anterior–posterior direction. In summary, these studies suggest that the synchronization within this lower frequency band (0–5 Hz) was probably not driven by a

neural circuitry innervating multiple muscles, but may result from a coupling between the COP oscillations and the modulation of muscle activity of the most distal leg muscles.

In addition to this lower frequency band, the present study reported consistent significant values of coherence estimations for the frequency band of 5–20 Hz. Previous studies have also reported intermuscular synchronization within this range for several different tasks and conditions including, but not limited to, grasping, quiet standing, and walking (McAuley et al 1997, Grosse and Brown 2003, Halliday et al 2003, Hansen et al 2005, Sowman et al 2006, Boonstra et al 2007,2008, Danna-dos-Santos et al 2010, Poston et al 2010). Commonly, synchronizations at lower frequency bands found by these studies are most pronounced around 10 Hz and seem not to have originated cortically but by reticulospinal circuitry (Grosse and Brown 2003, Boonstra et al 2009). These results corroborate synchronizations at similar frequencies. In the present study, the strength of coherence was similar across posterior muscle pairs. This is an interesting finding since it reflects a certain independence of the distribution of correlated neural inputs from the anatomical placement of the muscles involved. Note that the three muscle pairs studied have distinct anatomical relations: the SOL–BF muscle pair is formed by two adjacent muscles placed relatively more distal within the M-mode; BF–ERE is a muscle pair formed by adjacent muscles placed proximally; and the SOL–ERE is a muscle pair formed by two distantly placed muscles where they are the farthest and most proximal muscles of the chain, respectively. These findings indicate a functional organization of M-modes. Poston et al (2010) also reported a functional organization for hand muscles involved in isometric tripod grasping. They found two major groups of muscles: one formed only by extrinsic hand muscles and another formed only by intrinsic hand muscles. These two muscle groups exhibited stronger coherence values compared to any other muscle pair combination. This functional modal organization during grasping was shown to be stable for different levels of isometric contraction (Poston et al 2010) and also to be fatigue resistant (Danna-dos-Santos et al 2010). The two groups described by Poston et al (2010) may reflect two distinct hand-dedicated M-modes assembled in a similar fashion to posture-dedicated M-modes. Taken together, these observations suggest that these patterns of intermuscular coherence likely reflect the formation of a dedicated neural circuitry with the goal of assembling these M-modes.

3.6.   CONCLUSIONS

The findings suggest that the coordination of postural muscles likely involve correlated neural inputs to anatomically distinct muscles in order to form modal units that can be manipulated by a high-level controller. The synchronization at lower frequency bands observed between muscle pairs (pooled and separately) were found to be concentrated within two frequency bands: 0–5 Hz and 5–20 Hz. The former frequency band has showed stronger synchronizations for the more distal muscle pair (SOL and BF) and the source for such synchronization requires further experimental testing. The latter frequency band showed similar strength of its synchronization among all three postural muscles recorded and has been interpreted as a sign of a common circuitry underlying multi-muscle control. In addition, the study advanced the knowledge of using a novel approach, the intermuscular coherence analysis, to investigate multi-muscle control during bipedal stance. The next logical step in the development of this approach is to expand it to a larger number of muscles and investigate the influence of postural challenges to the formation of muscle synergies. Therefore, the exploration of possible effects of aging to this mechanism will be possible.

3.7.   REFERENCES

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CHAPTER 4

Multi-muscle control during unperturbed bipedal stance and

the effects of visual input on postural control

4.1.   INTRODUCTION

The execution of bipedal stance posture involves many mechanical challenges imposed by the design of the musculoskeletal system. The vertical orientation of the head–leg–trunk segments, the high center of mass, the large number of joints, and the narrow base of support all together impose mechanical instability that needs to be counterbalanced by the precise activation of multiple muscles spanning across multiple joints (ankle, knee, hip, and intervertebral joints).

Previous studies have provided important insights to the principles of multi-muscle control, including evidence that the central nervous system (CNS) unites muscles into functional groups (synergists) to reduce the number of variables to be controlled (Bernstein 1967, reviewed in Latash 2008). According to this perspective, the control of the human bipedal stance can be represented by a hierarchical scheme composed of at least two levels: a lower level where these functional muscle groups are formed, and a higher level where these groups are activated in a synergistic fashion to control physical variables directly related to task execution (Latash 2008). Several experimental findings support this hypothesis. For example, Krishnamoorthy et al (2003a) reported that, during the execution of a whole-body voluntary sway in the anterior-posterior direction, three major functional muscle groups co-varied their magnitude to provide a stable trajectory of the position of the body's center of pressure (COP). These authors employed the method of principal components analysis to identify these muscle groups and introduced the term muscle mode or M-mode to describe them. One may view M-modes as "virtual muscles" manipulated at a higher level of the control relevant task variables. The M-modes identified were as follows: a posterior M-mode formed by posterior postural muscles (soleus, biceps femoris, semitendinosus, lumbar erector spinae, and gastrocnemius), an anterior M-mode formed by anterior postural muscles (tibialis

anterior, vastus medialis, vastus lateralis, rectus femoris, and rectus abdominis muscle), and a third M-mode often formed solely by the rectus abdominis (Krishnamoorthy et al 2003a). These findings have been replicated in several other studies (Krishnamoorthy et al 2003b, Wang et al 2005, Danna-dos-Santos et al 2007,2008,2009). However, the neural mechanisms related to their formation remain unclear.

Chapter 3 focused on the hypothesis of correlated neural inputs as one of the neural mechanisms used by the CNS to coordinate the activation of synergistic muscles forming one of these M-modes. More specifically, intermuscular coherence between pairs of EMG signals composing a posterior M-mode (soleus, biceps femoris, and lumbar erector spinae muscles) was measured. The results confirmed significant intermuscular coherence at lower frequency bands, revealing the presence of correlated neural inputs to these muscles in healthy young individuals. In order to provide further evidence of this principle of multi-muscle control and better understand the mechanisms underlying the formation of functional muscle groups, studies involving additional muscles and experimental conditions are necessary.

The first goal of the present study was to investigate the role of correlated neural inputs in the formation of multiple synergistic groups. The analyses included two major synergistic muscle groups involved in upright standing: the anterior and posterior M-modes previously reported by Krishnamoorthy et al (2003a). Based on the results presented in *Chapter 3* as well as previous studies (Krishnamoorthy et al 2003b, Wang et al 2005, Danna-dos-Santos et al 2007,2008,2009), it was hypothesized that muscles comprising each of these synergistic groups would be coordinated by correlated neural inputs and, hence, will exhibit significant intermuscular coherence at lower frequency bands. The second goal was to investigate the effects of visual information on the generation of correlated neural inputs to muscles forming the anterior and posterior synergistic muscle groups. The rationale was based on the idea that poor or absent visual input has a detrimental effect on postural stability (Allum and Pfaltz 1985, Fitzpatrick et al 1992, Simoneau et al 1992, Schumann et al 1995, Wood et al 2009). Therefore, it was expected that a temporary disruption of visual information would result in a significant reduction in correlated neural inputs, expressed by a decreased intermuscular coherence at lower frequency bands across synergistic muscles.

4.2. METHODS

Participants. Ten healthy adults (4 females and 6 males, mean age 26.8 years old and $SD = 2.7$, mean height 175.0 cm and $SD = 12.7$, mean weight 80.6 kg and $SD = 22.0$) participated voluntarily in this study. All participants had no history of neurological or muscular disorders. All participants were right-handed and right-footed based on their preferred hand for writing and eating, and foot for kicking a soccer ball. Prior to participation, all participants voluntarily gave their informed consent based on the procedures approved by the Institutional Review Board at The University of Montana and conformed to The Declaration of Helsinki.

Apparatus. Active surface electrodes (Delsys Bagnoli single differential DE-2.1) were used to record the activity of the following muscles: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA). The electrodes were placed on the right side of the participant's body and over the muscle bellies based on recommendations reported in the literature (Basmajian, 1980): TA electrode at one-third on the line between the tip of the fibula and the tip of the medial malleolus; SOL electrode laterally below the heads of the gastrocnemius bellies; RF electrode at 50% on the line from the anterior superior iliac spine to the superior part of the patella; BF electrode at half way between the ischial tuberosity and the lateral epicondyle of the tibia; RA electrode 30 cm lateral to the umbilicus; and ERE electrode 3 cm lateral to L1. A reference electrode was placed over the lateral aspect of the fibular malleolus. *Figure 4.1* shows the placement of the electrodes. Signals from the electrodes were amplified ($1,000\times$) and band-pass filtered (6–500 Hz). All signals were sampled at 1,200 Hz with a 12-bit resolution.
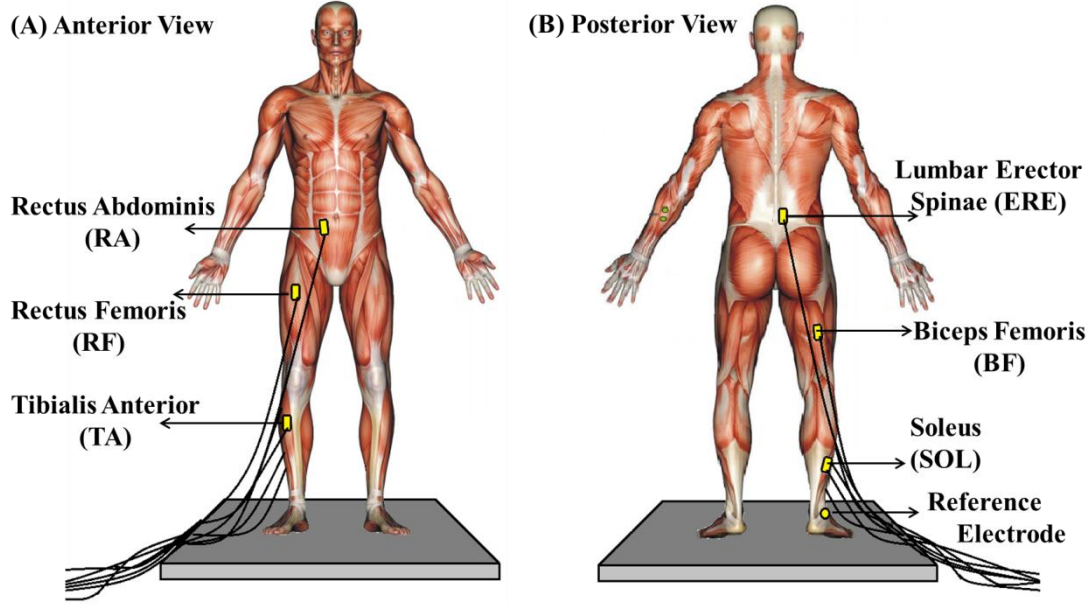
Figure 4.1. Position of the six electrodes and the reference electrode. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

Experimental procedures. All participants performed four independent unperturbed bipedal standing trials. Two trials were performed with opened eyes (*BOE* or *Vision* condition) and two with closed eyes (*BCE* or *No Vision* condition). To avoid any possible effects of BCE trials on the distribution of correlated neural inputs recorded during BOE trials, BOE trials were performed first. For all trials, participants were instructed to stand barefoot on a force platform with their upper limbs crossed against their chest, and their feet parallel and 15 cm apart for 35 s. To avoid any discrepancies between feet positioning across trials, the initial feet position was marked on the top of the platform and participants were asked not to move their feet during the entire data recording. For BOE trials, participants were asked to focus their vision to a physical static point placed at eyes height and at a distance of approximately 2 m, while they were asked to close their eyes during BCE trials. To avoid recording of any transient effects, the first 5 s were discarded and the remaining 30 s were analyzed. An inter-trial interval of 60 s was provided to avoid fatigue or discomfort. The total duration of the experiment, including preparation and placement of electrodes, explanation of the tasks, and data

recording, was approximately 30 min and none of the participants reported fatigue as an issue.

4.3. DATA PROCESING

All EMG signals recorded were analyzed off-line with custom-written software routines (Matlab R2012b, The MathWorks).

4.3.1. Muscle activation: a time domain analysis of EMG signals

EMG signals recorded from all six muscles were submitted to two time domain analyses. The first analysis quantified the relative *amplitude of activation* of each muscle by computing a relative index of muscle activation ($Index_{EMG}$) as follows. First, EMG signals from BOE and BCE trials were visually inspected to verify the presence of any signal artifacts and, then, filtered (20 Hz high-pass, second-order, zero-lag Butterworth filter) and full-wave rectified. Each signal from BCE trials was integrated over its trial length and normalized by similar integrals performed on EMG signals from BOE trials as follows. This analysis was used to confirm that all muscles recorded had comparable magnitude of activation across the execution of the two experimental conditions.

$$Index_{EMG} = \frac{I_{BCE}}{I_{BOE}} = \frac{\int_{0\ sec}^{30\ sec} EMG_{BCE}}{\int_{0\ sec}^{30\ sec} EMG_{BOE}}$$

$$(4.1)$$

The second analysis quantified the *patterns of multi-muscle activation* during BCE trials by running a vectorial comparison separately for the anterior and posterior muscle groups (TA/RF/RA and SOL/BF/ERE, respectively). This analysis was

performed to assure that all participants used similar patterns of muscle activation under both experimental conditions. Therefore, the degree of similarity between patterns of muscle activation was based on the cosine of the angle between pairs of muscle activation patterns vectors obtained for BOE and BCE trials. These vectors were assembled in 3D space using normalized *Index$_{EMG}$*. The cosine of the angle between pairs of these vectors quantified the degree of similarity of their spatial orientation between the two experimental conditions (*BOE* and *BCE*). See Danna-dos-Santos et al 2007, Poston et al 2010, and the *Methods* section of *Chapter 3* for more details on the computation of muscle activation pattern vectors. The cosine values range from 0 to 1, where the former indicates perpendicularity or dissimilarity of muscle activation patterns between the two experimental conditions, and the latter indicates parallelism or similarity of muscle activation patterns.

4.3.2. Muscle activation: a frequency domain analysis of EMG signals

EMG signals were analyzed in the frequency domain by estimating *intermuscular coherence* for muscle pairs separately (single-pair estimations) and combined (pooled estimations) using similar procedures reported by Poston et al (2010) and *Chapter 3*. EMG signals from the two trials collected under each experimental condition were concatenated to create a single longer time series of 60 s, totalizing 72,000 data points. Concatenation is a standard procedure used to increase the reliability of coherence estimation (Amjad et al 1997, Maris et al 2007, Poston et al 2010).

Single‑pair coherence estimations. The study followed the functional relationship of the postural muscles, such as the muscles pushing the body forward (anterior M-mode) and the muscles pushing the body backward (posterior M-mode) as reported by Krishnamoorthy et al (2003a). In order to emphasize the functional role of the distribution of correlated neural inputs, the intermuscular coherence was also estimated between muscle pairs formed by one anterior and one posterior muscle with

and without antagonist relation (named in this study as *antagonist* and *mixed groups*, respectively). *Table 4.1* shows all 15 muscle pairs and their relationships.

Table 4.1. Fifteen muscle pairs formed by solely posterior muscles, solely anterior muscles, antagonist muscles, or mixed muscles (one anterior and one posterior, non-antagonist, muscles). Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

	Muscle pair	Group (anatomic position)
1	SOL – BF	
2	SOL – ERE	Posterior
3	BF – ERE	
4	TA – RF	
5	TA – RA	Anterior
6	RF – RA	
7	TA – SOL	
8	RF – BF	Antagonist
9	RA – ERE	
10	SOL – RF	
11	SOL – RA	
12	BF – TA	Mixed
13	BF – RA	
14	ERE – TA	
15	ERE – RF	

Single-pair coherence was estimated by normalizing the cross-spectrum of two EMG signals (*fxy*) squared by the product of the auto spectrum of each signal (*fxx* and *fyy*) at each frequency (λ) as follows:

$$\left| R_{xy}(\lambda) \right|^2 = \frac{\left| f_{xy}(\lambda) \right|^2}{\left| f_{xx}(\lambda) f_{yy}(\lambda) \right|} \tag{4.2}$$

Intermuscular coherence estimates were obtained from non-overlapping 1 s data segments (i.e., 1,200 samples per segment), resulting in a frequency resolution of 1 Hz. The frequency range analyzed in this study was bounded from 0 to 55 Hz. To avoid the inclusion of the mechanical effects of sway and its coupling to the low-frequency content (0–1 Hz) of the EMG signals recorded reported by Mochizuki et al (2006), all analyses excluded the frequency band of 0–1Hz. Coherence estimates were considered statistically significant when they exceeded the significance limit of the null distribution of no coherence, computed based on Rosenberg et al (1989). The significance limit at $\alpha = 0.05$ was determined by

$$Sig(\alpha) = 1 - (1 - \alpha)^{\frac{1}{L-1}} ,$$
(4.3)

where L is the number of disjoint segments.

In order to compare coherence estimates across participants and under different experimental conditions, all estimates were z-transformed by computing the Fisher transformation of the estimates as proposed by Rosemberg et al (1989) and Amjad et al (1997):

$$\tanh^{-1}(x) = 0.5 \cdot \log \frac{1+x}{1-x} ,$$
(4.4)

where x is the coherence estimate.

Analysis of the *frequency distribution* of correlated neural inputs was achieved by identifying frequency intervals showing significant coherence values. The *strength of correlated neural inputs* was quantified by computing the integrals of the z-scored coherence within the frequency intervals of interest (INT_{Coh}).

***Pooled coherence estimations*.** Four pooled coherence analyses were performed separately. The first included the three pairs formed solely by posterior muscles (SOL/BF, SOL/ERE, and BF/ERE); the second analysis included the three pairs formed

solely by anterior muscles (TA/RF, TA/RA, and RF/RA); the third included the three pairs formed by antagonist muscles (TA/SOL, RF/BF, and RA/ERE); and the fourth analysis included the six remaining pairs formed by one posterior and one anterior, non-antagonist, muscles (SOL/RF, SOL/RA, BF/TA, BF/RA, ERE/TA, and ERE/RF). Pooled coherence estimates are considered a weighted average of individual coherence estimates and can be used to increase statistical power. Estimates of pooled coherence were obtained as proposed by Amjad et al (1997):

$$\frac{\left|\sum_{i=1}^{k} f_{xy}(\lambda)L_i\right|^2}{\left(\left(\sum_{i=1}^{k} f_{xx}(\lambda)L_i\right)\left(\sum_{i=1}^{k} f_{yy}(\lambda)L_i\right)\right)} \tag{4.5}$$

Analysis of the *frequency distribution and strength* of correlated neural inputs obtained from pooled coherence calculations were also based on the determination of frequency bands within significant coherence estimates and the calculation of integrals within these frequency bands, respectively.

Statistical approach. Medians across participants for the integrals of the z-scored coherence profiles within the frequency band of 1–10Hz ($INT_{Coh1\text{-}10}$) are reported and statistical tests were performed using the IBM SPSS statistics software suite (version 22, IBM® SPSS®). Considering the small sample size, Shapiro-Wilk tests were performed to test for normality of the coherence variable. Since some of the response variables did not follow a normal distribution, non-parametric paired tests (Wilcoxon signed-rank tests) were used to investigate the effect of *Vision* (BOE and BCE) on $INT_{Coh1\text{-}10}$. Since multiple comparisons were performed, the significant level was adjusted at 1% ($\alpha = 0.01$) for individual tests.

4.4. RESULTS

4.4.1. Muscle activation: time domain analysis

As expected, all participants performed both visual conditions with ease. They also used similar muscle activation levels across the two experimental conditions. *Figure 4.2* shows the rectified EMG recordings for all six muscles obtained from a representative participant during unperturbed stance with open eyes and closed eyes (panel *A*), and the average across participants for the $Index_{EMG}$ (panel *B*), which describes the ratio between the integrals of EMG signals recorded during the trials. Averages of the $Index_{EMG}$ close to 1 indicate that participants generally employed similar magnitudes of activation in both conditions. Participants also employed comparable patterns of muscle activation in both experimental conditions, as shown by the vector analysis. The mean cosine of the angle between resultant muscle activation vectors across participants were around the value of 1, with very little deviation (mean cosine = 0.979 and *SD* = 0.022 for the posterior muscle group, and mean cosine = 0.971 and *SD* = 0.057 for the anterior muscle group).
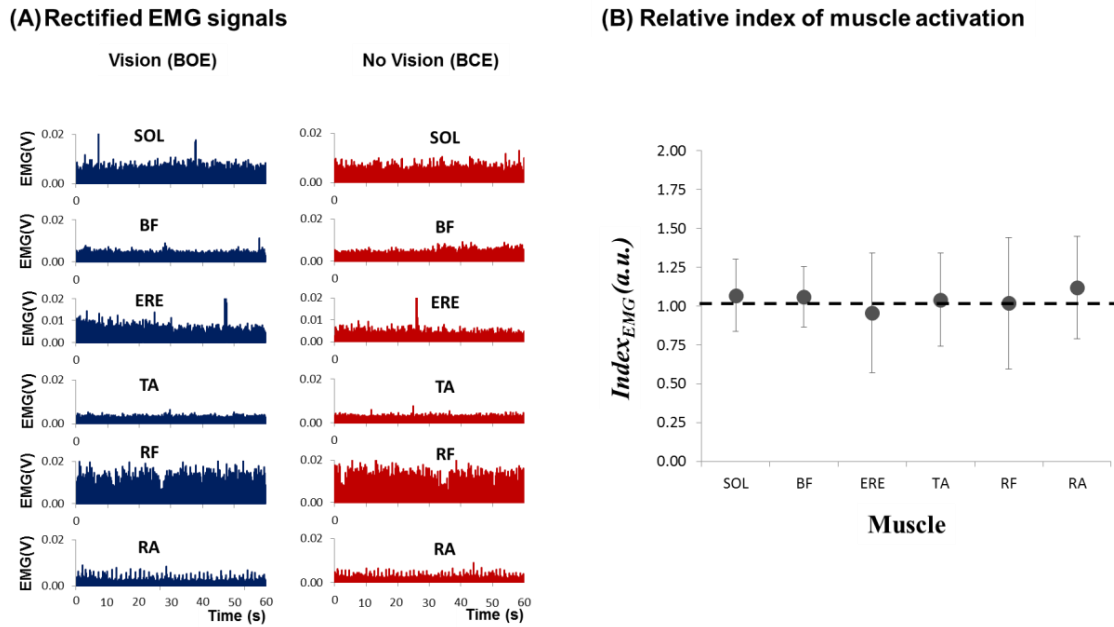
Figure 4.2. (A) Rectified EMG signals of six muscles recorded from a representative participant during unperturbed stance with open and closed eyes (*BOE* and *BCE* conditions, respectively). **(B)** Averages and standard deviations across participants of the ratio between the integrals of EMG signals from *BOE* and *BCE* trials (*Index$_{EMG}$*). Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

4.4.2. Muscle activation: frequency domain analysis

Pooled coherences. *Figure 4.3* illustrates the averaged pooled coherence profile across participants computed for each muscle group (anterior, posterior, antagonist, and mixed) and under both vision and no vision conditions (BOE and BCE, respectively). Note that the pooled coherence for both anterior and posterior muscle groups was significant within the frequency interval of 0–10 Hz during the BOE trials, while it decreased during BCE trials. Regarding the antagonist muscle group, its pooled coherence was significant not only within the frequency band of 0–10 Hz, but also within 10–30 Hz for either BOE or BCE trials. In contrast, no significant intermuscular coherence was observed across all frequencies for the mixed muscle group for either BOE or BCE trials. Based on these findings and discarding the frequency band of 0–1 Hz to exclude the coupling of the mechanical effect of sway to the EMG signals, the frequency of interest was determined to be between 1 and 10 Hz. Medians across participants for the integrals of the z-scored coherence computed within the frequency band of interest ($INT_{Coh1-10Hz}$) are displayed in *Figure 4.4* and *Table 4.2*.

A significant decrease in the strength of the correlated neural inputs during BCE trials was observed for both anterior and posterior muscle groups. Wilcoxon signed-rank tests illustrated the effect of *Vision* (BOE and BCE) on the variable $INT_{Coh1-10Hz}$ for both anterior and posterior muscle groups. No significant effect of *Vision* on $INT_{Coh1-10Hz}$ was detected for either the antagonist or the mixed muscle groups. See *Z* and *p* values in *Table 4.2*.

POOLED COHERENCE (z-score)

(A) Anterior group

(B) Posterior group

(C) Antagonist group

(D) Mixed group
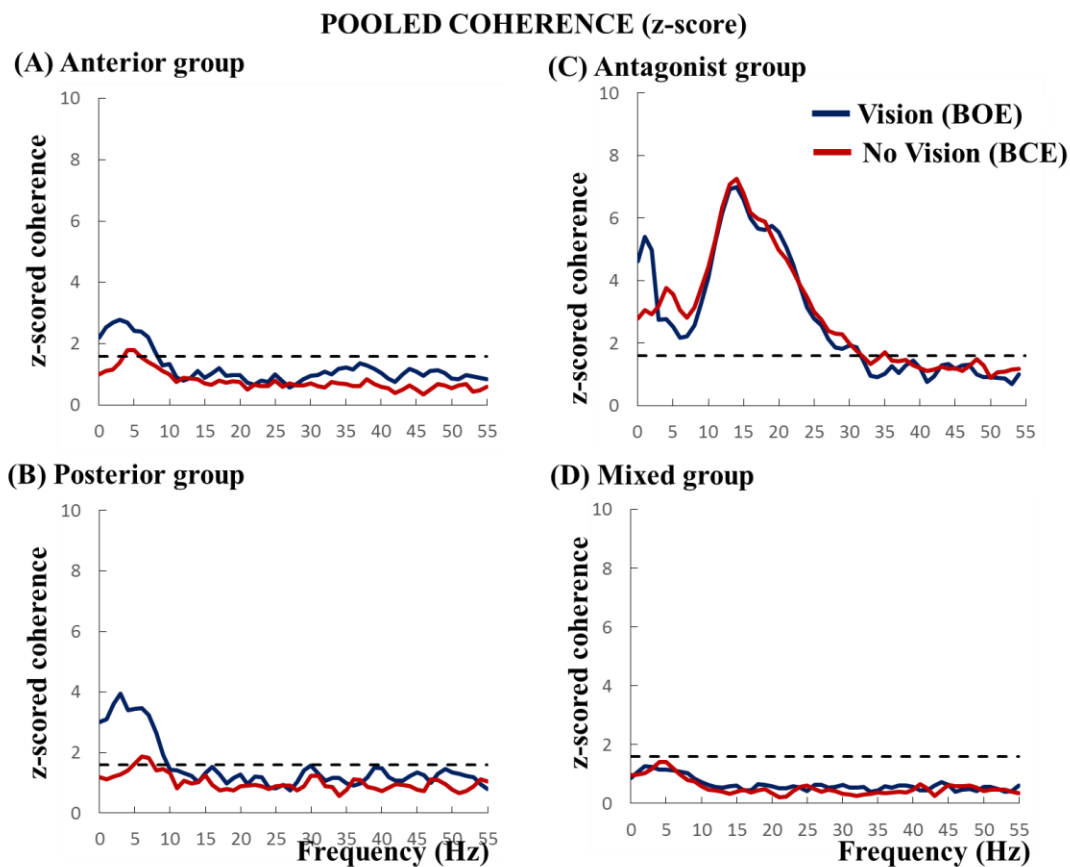
Figure 4.3. Average across participants of the z-scored pooled coherence profiles obtained for all pairs formed between posterior, anterior, antagonist, and mixed muscles during unperturbed stance with and without vision (*BOE* and *BCE* conditions, respectively). Note: *dashed line* represents the significant limit at 5%.
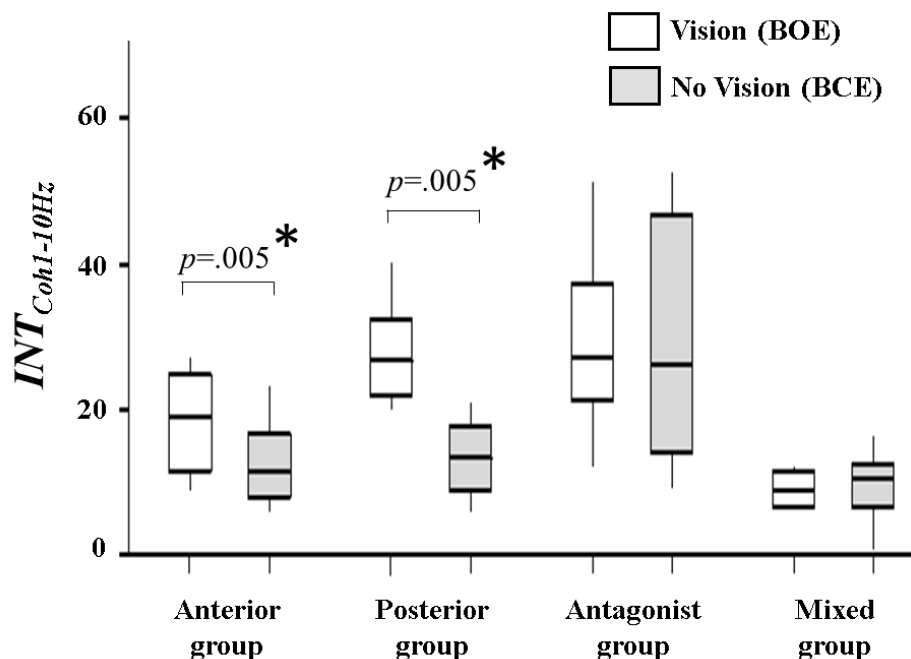
Figure 4.4. Box-plots of the integrals of the z-scored pooled coherence profiles within the 1–10 Hz frequency band ($INT_{Coh1-10Hz}$) for each muscle group (anterior, posterior, antagonist, and mixed) during unperturbed stance with and without vision (*BOE* and *BCE* conditions, respectively). Note: * indicates *Vision* effect ($p < 0.01$).

Table 4.2. Median and 95% confidence interval of the difference of the integrals of the z-scored pooled coherence computed over the frequency interval of 1–10 Hz ($INT_{Coh1-10Hz}$) for each muscle group (anterior, posterior, antagonist, and mixed) during unperturbed stance with and without vision (*BOE* and *BCE* conditions, respectively). Note: [+] represents *Vision* effect ($p < 0.01$).

Muscle Group	$INT_{Coh1-10Hz}$ BOE (Vision)	BCE (No Vision)	95% confidence interval of the difference	Z and p values
Anterior	19.72	11.42	(1.73, 16.56)	$Z = -2.803, p = .005$ [+]
Posterior	26.66	13.12	(8.96, 20.30)	$Z = -2.803, p = .005$ [+]
Antagonist	26.96	26.23	(-12.5, 13.4)	$Z = -0.255, p = .799$
Mixed	8.91	10.53	(-3.58, 4.09)	$Z = -0.153, p = .878$

Single‑pair coherences. *Figure 4.5* shows the average intermuscular coherence spectra for each anterior, posterior, antagonist, and mixed muscle pair within the frequency band of 0–55Hz. *Figure 4.6* and *Table 4.3* present the medians across participants for the integrals of the z-scored coherence computed within the frequency band of interest ($INT_{Coh1-10Hz}$).

Single-pair coherences within the frequency band of 1–10Hz for the pairs formed by solely **anterior muscles** (TA/RF, TA/RA, and RF/RA) were significant for the muscle pair TA/RF under both BOE and BCE conditions, and for the muscle pair TA/RA under the BOE condition. No significant intermuscular coherence within 1–10 Hz was observed for the TA/RA pair under BCE condition or for the RF/RA pair under either BOE or BCE conditions (*Figure 4.5ABC)*. Wilcoxon signed-rank tests revealed a significant decrease in the strength of the correlated neural inputs in the frequency band 1–10 Hz when participants closed their eyes for both TA/RF and TA/RA muscle pairs. No effect was found for the muscle pair RF/RA. See *figure 4.6A* and *Z* and *p*-values in *Table 4.3*.

Single-pair coherence estimates within 1–10Hz for the pairs formed by solely **posterior muscles** (SOL/BF, SOL/ERE, and BF/ERE) were significant for all three muscle pairs under the BOE condition (*Figure 4.5DEF*). A significant decrease in the strength of the correlated neural inputs within this frequency band during BCE trials (*Figure 4.6B* and *Table 4.3*) was confirmed by Wilcoxon signed-rank tests with factor *Vision* on $INT_{Coh1-10Hz}$. See *figure 4.6B* and *Z* and *p*-values in *Table 4.3*.

Interestingly, the single-pair coherence profiles for pairs formed by **antagonist muscles** (TA/SOL, RF/BF, and RA/ERE) were significant within a larger frequency band. *Figure 4.5GHI* shows a significant intermuscular coherence within 1–30 Hz for all three antagonist pairs. Wilcoxon signed-rank tests revealed no effect of *Vision* (BOE and BCE) on the variable $INT_{Coh1-10Hz}$ for any antagonist pair. See *figure 4.6C* and *Z* and *p*-values in *Table 4.3*.

Regarding the muscle pairs formed by one anterior and one posterior, non-antagonist, muscles (the **mixed muscle pairs**), no significant intermuscular coherence was observed (*Figure 4.5, panels J* to *O*) for either BOE or BCE conditions. In addition, no significant effect of *Vision* on the variable $INT_{Coh1-10Hz}$ was observed according to Wilcoxon signed-rank tests. See *Z* and *p*-values in *Table 4.3*.
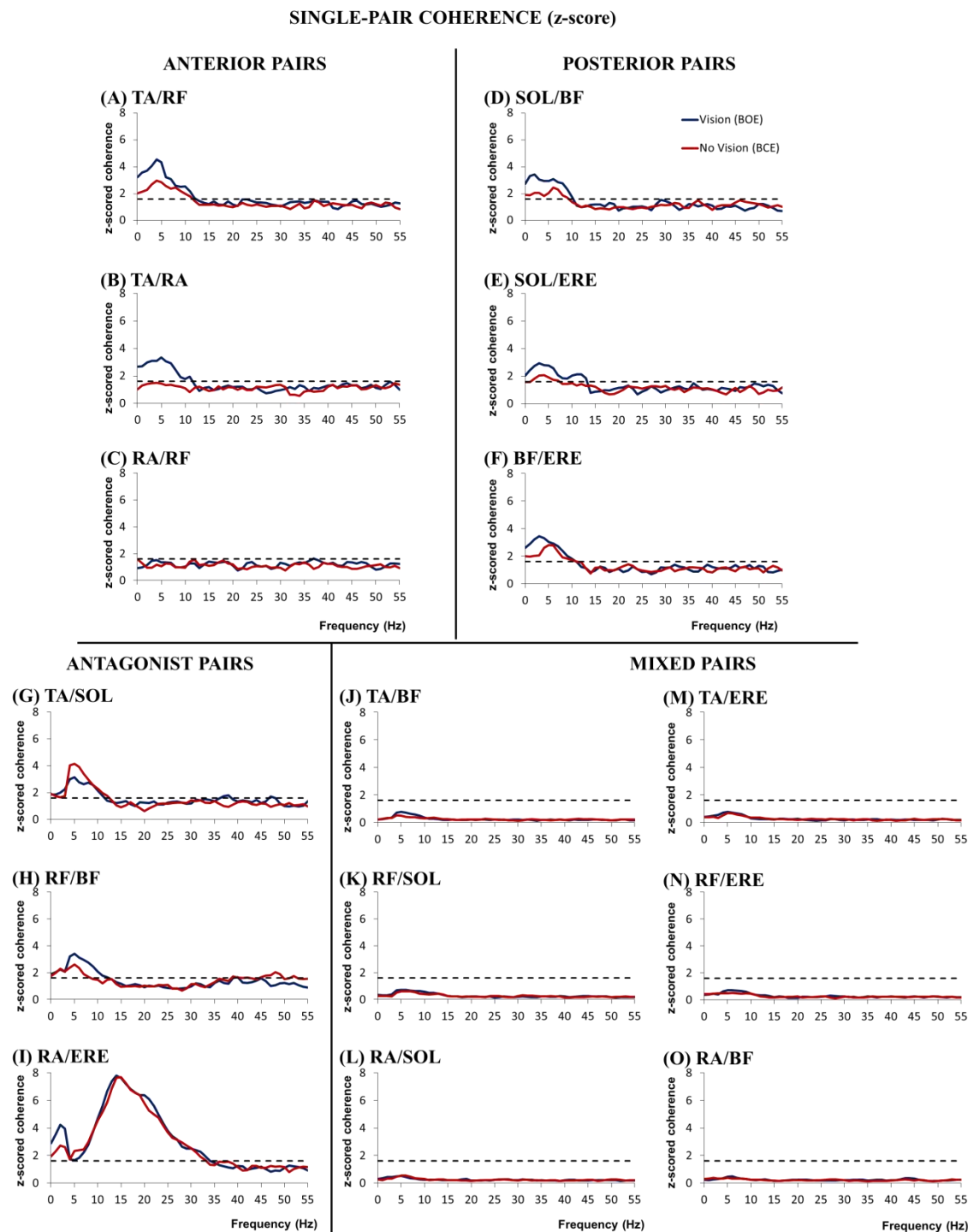
Figure 4.5. Average across participants of the intermuscular coherence profiles obtained separately for each pair of muscles during unperturbed stance with and without vision (*BOE* and *BCE* conditions, respectively). Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA). *Dashed line* represents the significance level for no significant intermuscular coherence.
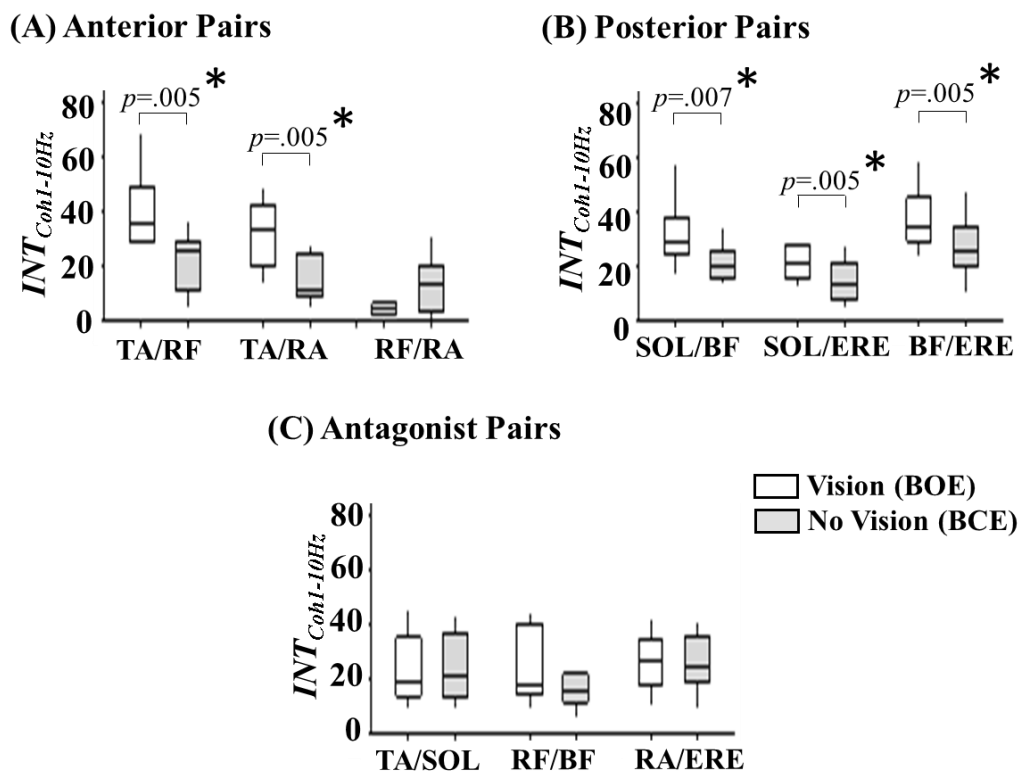
Figure 4.6. Box-plots of the integrals of the z-scored intermuscular coherence profiles within the 1–10Hz frequency band ($INT_{Coh1-10Hz}$) for the muscle pairs formed by solely anterior muscles, solely posterior muscles, and antagonist muscles during unperturbed stance with and without vision (*BOE* and *BCE* conditions). Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA). Note: * indicates *Vision* effect ($p < 0.01$).

Table 4.3. Median and 95% confidence interval of the difference of the integrals of the z-scored intermuscular coherence computed over the frequency interval of 1–10 Hz ($INT_{Coh1-10Hz}$) for the fifteen muscle pairs formed by solely anterior muscles, solely posterior muscles, antagonist, and mixed muscles during unperturbed stance with and without vision (*BOE* and *BCE* conditions, respectively). Note: [+] represents *Vision* effect ($p < 0.01$).

	$INT_{Coh1-10Hz}$		95% confidence interval of the difference	Z and *p* values
	BOE **(Vision)**	**BCE** **(No Vision)**		
Pairs formed by anterior muscles				
TA/RF	36.09	23.82	(8.6, 26.6)	Z = -2.803, p =.005 [+]
TA/RA	33.66	11.17	(8.6, 27.2)	Z = -2.803, p =.005 [+]
RF/RA	4.48	11.91	(-14.1, 5.8)	Z = -1.172, p =.241
Pairs formed by posterior muscles				
SOL/BF	28.97	18.12	(3.2, 20.3)	Z = -2.701, p =.007 [+]
SOL/ERE	21.49	12.20	(2.2, 14.3)	Z = -2.803, p =.005 [+]
BF/ERE	34.54	23.80	(1.7, 21.8)	Z = -2.803, p =.005 [+]
Pairs formed by antagonist muscles				
TA/SOL	18.68	20.01	(-19.2, 12.4)	Z = -0.459, p =.646
RF/BF	18.14	15.18	(-6.6, 18.2)	Z = -1.070, p =.285
RA/ERE	27.00	23.38	(-7.3, 10.9)	Z = -0.255, p =.799
Pairs formed by mixed muscles				
TA/BF	5.49	3.75	(-0.49, 3.19)	Z = -1.580, p =.114
TA/ERE	5.40	4.02	(-1.63, 3.37)	Z = -0.561, p =.575
RF/SOL	4.86	3.64	(-1.70, 3.97)	Z = -0.764, p =.445
RF/ERE	5.88	4.18	(-1.31, 3.21)	Z = -1.070, p =.285
RA/SOL	4.06	3.87	(-0.96, 1.02)	Z = -0.153, p =.878
RA/BF	3.29	2.91	(-1.33, 1.50)	Z = -0.153, p =.878

4.5. DISCUSSION

The present study confirmed the presence of correlated neural inputs to different postural muscles forming synergies intended to control unperturbed stance. Antagonist muscles showed a significant coherence at lower frequency bands within 1–30 Hz; whereas the anterior and posterior muscle groups presented signs of coherence within 1–

10 Hz and the mixed group showed no signs of synchronization at lower frequency bands. Note that the larger distribution of synchronization of antagonist muscles was due to the muscle pair RA/ERE, though. Different distribution and strength of synchronization of antagonist muscles were expected since agonist/antagonist pairs were already recognized for the level of coupling between their EMG signals (Hansen et al 2002). Interestingly, the coherence for both anterior and posterior muscle groups was no longer significant when participants closed their eyes. Therefore, the study confirmed the hypothesis that a short-term interruption of visual input affects the generation of correlated neural inputs to multiple postural muscles.

The control of the human's bipedal stance requires the coordination of multiple postural muscles within temporal, spatial, and magnitude precision. Several studies based on the Uncontrolled Manifold (UCM) Hypothesis had provided evidence of the formation of synergistic muscle groups to control bipedal stance (Krishnamoorthy et al 2003a,b, Danna-dos-Santos et al 2008). These findings supported the idea of the CNS controlling a large number of possible patterns of muscle activation by forming such synergistic muscle groups. More importantly, the study provided initial evidence indicating that the generation of such groups during bipedal stance is driven by correlated neural inputs to different postural muscles.

The use of the intermuscular coherence approach at lower frequency bands revealed significant synchronization of EMG signals within the frequency band of 1–10 Hz for anterior (TA, RF, and RA) and posterior (SOL, BF, and ERE) postural muscles, representing the push-forward and push-back M-modes controlling body sway during unperturbed stance; and a significant synchronization of agonist/antagonist EMG signals within 1–30 Hz, representing another M-mode controlling body sway in the sagittal plane. Interestingly, the frequency profiles of correlated neural inputs were similar for the posterior and anterior muscle groups (*Figure 4.3AB*). One may suggest that these two synergistic muscle groups received the same correlated input. However, intermuscular coherence at lower frequency bands was computed across all time points and, despite similar intermuscular coherence spectra, the posterior and anterior muscle groups may have received correlated input, but at different points in time (e.g., when the body was swaying either backward or forward). Indeed, Boonstra et al (2009a) showed that

bilateral TA muscles receive correlated 10 Hz input only in the most posterior position when swaying in the anterior–posterior direction. In addition, synchronization of EMG signals within this frequency interval (<10 Hz) is consistent with previous studies that reported similar findings during the execution of slow movements (Kamen and De Luca 1992, Farmer et al 1993, Vallbo and Wessberg 1993, De Luca and Erim 2002) and during bipedal stance tasks (Danna-dos-Santos et al 2014).

The neural sites responsible for the generation of correlated inputs to multiple muscles remain unclear, and multiple origins have been proposed in the literature. For example, Farmer et al (1993) studied stroke survivors with damage to motor cortical areas and demonstrated that, despite cortical damage, intermuscular coherence at lower frequency bands was still present within the frequency interval of 1–12 Hz. Their findings suggested that synchronization within this frequency interval was unlikely to originate within the motor cortex. Boonstra et al (2009b) also showed that intermuscular coherence at 7–13 Hz between bilateral hand muscles was not synchronized with cortical activity, further supporting a sub-cortical origin of intermuscular coherence in the lower frequency band. In contrast, the collective results of Mima and colleagues (Mima and Hallett 1999, Mima et al 2000,2001) have not only reported significant values of cortico-muscular coherence within the frequency interval of 3–13 Hz for hand muscles, but they also extended this finding to other muscles, such as the biceps brachii and abductor hallucis (intrinsic muscle of the foot). According to Mima et al (2000), synchronizations in this lower frequency band (3–13 Hz) likely reflect the involvement of the inferior olive and the thalamic cortical loop.

Interestingly, Danna-dos-Santos et al (2014) reported synchronizations of postural muscles at lower frequency bands not only within 1 and 10 Hz, but also extended to frequencies up to 20 Hz. In the present study, participants stood up freely (unperturbed bipedal stance task), whereas participants stood under an induced generalized isometric contraction of their postural muscles (bipedal stance holding an anterior 5 kg load) in the previous study (Danna-dos-Santos et al 2014). Changes in intermuscular synchronization and the presence of multiple significant frequency bands as a function of the type of contraction due to the task have been reported by other studies (Farmer et al 1993,

McAuley et al 1997, Mima et al 2000, Grosse et al 2002, Latash 2008). Therefore, the neural mechanisms used to generate motor outputs seem to be task specific.

The strength of correlated neural inputs within 1–10 Hz reported in the present study was similar across five muscle pairs (all three posterior muscles pairs and two anterior muscle pairs). This was interesting considering the distinct anatomical relations between muscle pairs: muscle pairs SOL/BF and TA/RF are formed by two adjacent muscles placed relatively more distally, whereas muscle pairs BF/ERE and RF/RA are formed by adjacent muscles placed more proximally, and muscle pairs SOL/ERE and RF/RA are formed by one proximal and one distal muscle. This relationship suggests that M-modes are likely formed based on their functional role of moving the body either backward or forward, rather than based on their anatomical location. The functional role of the distribution of correlated neural inputs is also supported by the results showing no significant intermuscular coherence between non-synergistic muscles (mixed group).

Regarding the effects of visual input on the control of multiple postural muscles during unperturbed stance, the present study showed a significant decrease in intermuscular coherence at lower frequency bands when participants stood up in absence of vision (BCE condition). The coherence estimates not only dropped, but they were also not significant when visual information was temporarily removed (except for the group formed by antagonist muscles). The findings suggest that visual information plays an important role in the formation of muscle synergies during unperturbed bipedal stance. Previous studies have shown mixed effects of the removal of visual information on intermuscular coherence. Boonstra et al (2008) reported an increased coherence within the frequency band of 0–5 Hz between bilateral lower leg muscles (soleus and gastrocnemius) during quiet stance in the absence of visual inputs. In contrast, Mochizuki et al (2007) showed that correlated input to individual motor units of bilateral soleus muscles did not differ between standing with eyes opened or closed. These contrasting findings regarding intralimb and inter-limb coherence may reflect a change in the control strategy after removing visual information. In this case, the postural control strategy seems to depend more on proprioceptive information from the ankle joint in the absence of vision (cf. Saffer et al 2008). Despite different effects of vision on postural control reported in the literature, it is suggested that changes in the availability of visual inputs

interfere with the organization of neural drive to synergistic muscles involved in postural control. Considering that visual information appears to play an important role in the generation of correlated neural inputs to different postural muscles, future studies investigating both intra- and inter-limb coherence are needed to map the reorganization of muscle synergies during temporary removal of visual information.

4.6. CONCLUSIONS

These findings suggest that the coordination of postural muscles likely involves the distribution of correlated neural inputs to distinct muscles in order to form modal units (also termed synergistic muscle groups or M-modes) that can be manipulated by the CNS. Synchronizations of postural muscles at lower frequency bands were found to be concentrated within a frequency interval of 1–10 Hz when visual information was available. The synchronizations reported here showed similar strength among six postural muscles and have been interpreted as signs of correlated neural drives to multi-muscle control. These results also suggest that these postural muscle groups are likely formed based on their functional role of moving the body either backward or forward, rather than based on their anatomical location. In addition, the lack of visual input during unperturbed stance not only decreased the correlation between muscle pairs in the frequency domain, but coherence at lower frequency bands was no longer significant. These findings revealed that temporary removal of visual input affected how the CNS organizes correlated neural inputs to generate synergistic muscle groups. Therefore, visual input plays a key role on the neural mechanisms underlying postural control.

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CHAPTER 5

The use of coherence analysis as a novel approach to detect

age-related changes on postural muscle control

5.1.   INTRODUCTION

The central question in human movement control is related to the neurophysiological mechanisms used by the central nervous system (CNS) to control the large and redundant number of degrees of freedom. In 1967, Nicolai Bernstein suggested that the CNS reduces the complexity of the system by uniting motor components into functional groups (Bernstein 1967). Under Bernstein's hypothesis (commonly referred to as the *Motor Redundancy Hypothesis*), the controller is not only able to overcome the redundancy of the system by controlling functional groups instead of each element separately. The controller is also able to rearrange its elements in order to adapt its strategy accordingly to the necessities imposed by the individual characteristics, motor task, and environment. Even though the principle proposed by Bernstein sounds simple and elegant, confirming this hypothesis has proven to be a challenge due to many factors, such as (a) the many levels of analysis one can perform to test its core principle; (b) the difficulty in developing adequate technology to record and analyze the relationships emerging from its basic synergistic elements; and (c) the lack of knowledge of neural mechanisms responsible for the formation of these patterns of muscle activation.

Technological development has allowed progress in the investigation of such functional muscle groups (Scholz and Schöner 1999, Scholz et al 2000,2001, Latash et al 2002,2003,2010, Krishnamoorthy et al 2003a,b,2004, Latash 2008, Danna-dos-Santos et al 2007,2008,2009,2014). The *Uncontrolled Manifold Analysis Method* (*UCM*), for example, has been successfully employed to both identify the emergence of synergistic muscle groups and link this pattern of muscle activation to the control of performance variables, such as the position of the body's center of pressure (Krishnamoorthy et al 2003a,b,2004, Danna-dos-Santos et al 2007,2008,2009). Even though these studies allowed progress in the recognition of such synergistic patterns, they did not reach the

means to explore the possible neural mechanisms related to the implementation of this control strategy.

In *Chapters 3* and *4*, a novel approach (the Intermuscular Coherence Analysis) was used to advance knowledge in examining the mechanisms underlying the organization of multi-muscle synergies. In *Chapter 3*, signs of the presence of correlated neural inputs distributed across soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE) corroborated previous reports that these three muscles form a synergistic group (Krishnamoorthy et al 2003a). This posterior muscle group has been referred to as the "*push-back synergistic muscle group*" or "*push-back M-Mode*", and it is considered one of the important lines of defense to counter-act the tendency of the body to fall forward (Krishnamoorthy et al 2003a). In addition, EMG recordings presented in *Chapter 3* revealed a specific anatomic distribution of correlated neural inputs to these three muscles: a significantly stronger coherence for the most distal muscle pair (soleus and biceps femoris). Interestingly, this finding corroborates the predominant ankle strategy observed in young adults to control upright standing (Horak and McPherson 1996, Horak 2006). Note that the ankle strategy is a simplistic way to describe an emphasis of movement around the ankle joints. However, it does not mean that other joints are not changing their angular position to maintain upright posture. In addition, *Chapter 4* expanded the use of intermuscular coherence by including a larger number of postural muscles as well as a different challenging task (upright stance with closed eyes). Once more, signs of synchronization at lower frequency bands among multiple postural muscles were revealed. These results suggested that this method is robust enough to continue its development in the study of human postural control in health and disease states.

Older adults are well known to adopt a larger repertoire of strategies to maintain upright posture, when compared to the dominant ankle strategy used by young adults. For example, one of the strategies employed by older adults to control upright stance is the increased EMG activity of muscles acting on both ankles and hips (Amiridis et al 2003, Benjuya et al 2004, Nagai et al 2011). The higher activation of ankle muscles was also reported during dynamic stance tasks, such as the functional stability boundary task (Nagai et al 2011). Other studies have shown a higher EMG activity of the tibialis

anterior (pulling the body forward) and semitendinosus (keeping the hip extended) under different static stance tasks (such as wide base of support, narrow base of support, tandem Romberg stance, and unipedal stance) in older adults, compared to that in young adults (Amiridis et al 2003, Benjuya et al 2004). Therefore, older adults seem to use both ankle and hip strategies to maintain upright standing. Another well-recognized strategy commonly used by older adults to control upright stance is the agonist-antagonist cocontraction of postural muscles. This motor strategy is typically used when function is not optimal. Older adults seem to cocontract their postural muscles as a mechanism to stiffen their joints and improve balance (Woollacott et al 1988, Manchester et al 1989, Melzer et al 2001, Benjuya et al 2004, Nagai et al 2011,2013, Papegaaij et al 2014, Lee et al 2015). Benjuya et al (2004) and Nagai et al (2011, 2013) reported a higher co-activation of muscles around the ankle (tibialis anterior and soleus) in older adults performing different static and dynamic standing tasks (Nagai et al 2011,2013). It is important to note, however, that the cocontraction strategy is not used only by older adults and individuals with balance disorders. This strategy has also been reported, for example, in young adults during the experience of fear at heights (Wuehr et al 2014), in elite athletes in response to unexpected perturbations (Mani et al 2014), and other situations.

Taken together, these observations suggest that older individuals may explore other strategies to achieve postural stability when age-related physiological changes affect their ability to generate optimal torque. Therefore, it is hypothesized that the aging CNS may be able to modulate the formation of correlated neural inputs to control postural muscles. The present study was designed to investigate the mechanisms underlying the organization of multi-muscle synergies to control upright stance in older adults using similar procedures of synergy mapping performed in *Chapters 3* and *4*, the intermuscular coherence analysis. It was expected that older adults would reorganize the modulation of the synchronization at lower frequency bands of three posterior postural muscles previously recognized as components of a synergistic muscle group in healthy young adults (soleus, biceps femoris, and lumbar erector spinae).

5.2.   METHODS

Participants. Nine healthy participants (4 females and 5 males, mean age 29.2 years old and *SD* = 6.1, mean height 1.71 m and *SD* = 0.74, mean weight 78.8 kg and *SD* = 20.5) and thirteen healthy older adults (8 females and 5 males, mean age 69.0 years old and *SD* = 3.4, mean height 1.66 m and *SD* = 0.08, mean weight 70.5 kg and *SD* = 8.9) participated in this study (*Control* and *Senior* groups, respectively). The exclusion criteria for both groups included (a) previous history of pathological sensory, musculoskeletal or neurological disorder, (b) history of previous surgeries, (c) history of cardiopulmonary disease, and (d) history of substance abuse. In addition, the senior group included only nonfallers with the age ranging 65 to 74 years (older adults). Prior to their participation, all participants voluntarily gave informed consent based on the procedures approved by the Institutional Review Board at The University of Montana and conformed to The Declaration of Helsinki.

Apparatus. A force platform (AMTI BP400600, AMTI Inc.) was used to acquire the vertical and horizontal components of the ground reaction force as well as the moments of force around the frontal and sagittal axes. These signals were transmitted to a dedicated system (Vycon MX Ultranet and Vycon Nexus version 1.6.1, Vycon®) for the computation of the body's center of pressure coordinates in anterior–posterior and medio–lateral directions (*COPap* and *COPml*, respectively). Features of the COP were recorded due to previous reports relating low-frequency COP modulation to low-frequency EMG modulation (Mochizuki et al 2006). COPap and COPml were defined by

$$COPap = (-h * F_x - M_y) / F_z \tag{5.1}$$
$$COPml = (-h * F_y - M_x) / F_z \ , \tag{5.2}$$

where h = the height of the base of support above the force plate; F_x = horizontal anterior–posterior component of the resultant force applied to the force plate; F_y = horizontal medio–lateral component of the resultant force applied to the force plate; F_z =

vertical component of the resultant force applied to the force plate; Mx = moment of force around the sagittal axis; and My = moment of force around the frontal axis.

Surface electrodes were used to record the activity of the following muscles: soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE) (*Figure 5.1*). The electrodes were placed on the right side of the participant's body over the muscle bellies (see *Chapter 3* for more details). A reference electrode was placed over the lateral aspect of the fibular malleolus. Signals from the control group were sampled at 1200 Hz and signals from the senior group at 2000 Hz, all with 12-bit resolution. Signals from the electrodes were amplified (1,000×) and band-pass filtered (6–500 Hz).
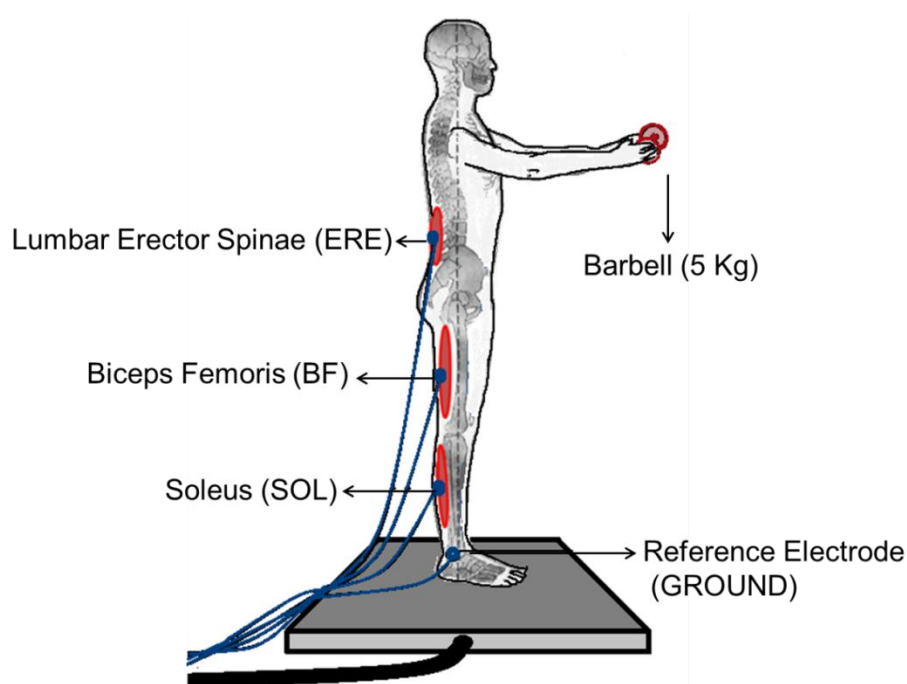
Figure 5.1. Representation of the electrodes placed on postural muscles. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), and reference electrode (GROUND).

Experimental Procedures. All participants performed ten trials of an experimental task. The experimental task consisted of bipedal stance for 15 seconds under the continuous perturbation of holding a load (a barbell of 5 kg) in front of the body with their shoulders flexed 90° and elbows fully extended (*Figure 5.1*). Participants

were also instructed to be barefoot, keep their feet parallel and separated by a distance of 15 cm, and focus their vision at a static point placed 2 m in front and at their eye level.

The first 5 seconds of the task were not recorded to avoid the recording of transient effects usually present in the first seconds of standing posture. The length of the trials (15 s) was stipulated to minimize the chances of back injury, discomfort, and fatigue while testing senior participants. A second measure to reduce these risks was to ensure a resting period of 60 seconds between trials.

The average duration of the entire experimental session was 30 min (including skin preparation, electrodes positioning, and performance of experimental tasks), and none of the participants reported fatigue or discomfort.

5.3. DATA PROCESSING

All COP coordinates and EMG signals recorded were analyzed off-line with custom-written software routines (Matlab R2012b, The MathWorks).

Postural sway behavior. COPap and COPml coordinate signals were filtered with a 20 Hz low-pass, second-order, and zero lag Butterworth filter. Normalization of the COP coordinates was obtained by subtracting their average position from initial values. This normalization allowed the researcher to position COP coordinates at the center of an xy coordinate system and, therefore, draw any comparisons of basic COP features across participants. In a sequence, the following postural indices were extracted from the COP signal: the elliptical area containing 95 % of the entire COP path (*Area*); the ranges and mean velocities of the COP displacement in both anterior-posterior and medio-lateral directions (*$Range_{AP}$, $Range_{ML}$, MV_{AP}*, and *MV_{ML}*, respectively); and the mean power frequency (*$Fmean_{AP}$*) and the maximum frequency containing 80% of the power spectral density (*$F80_{AP}$*) of the *COPap* signal. Measures of frequency were only extracted from the anterior-posterior component of the COP because the muscles recorded act mostly to move the body's COP in this direction. A more detailed description of the methods used

to compute these variables and their importance on the study of COP features can be found in Duarte and Freitas (2010) and *Chapter 7* of the dissertation.

Intermuscular coherence. For each participant, EMG signals from all ten trials were concatenated to create a long single time series (100 s; 120,000 data points). The process of concatenation was performed to increase the reliability of coherence estimations, as proposed by Maris et al (2007) and Poston et al (2010). Once concatenated, all three EMG signals were filtered by a 20 Hz high-pass, second-order, zero lag Butterworth filter.

Due to the lower number of muscles included in this study ($n = 3$), only intermuscular coherence estimates for each pair of EMG signals (SOL/BF, SOL/ERE, and BF/ERE) were calculated. Subsequently, intermuscular coherence (R) was estimated separately for EMGs recorded from each muscle pair by using the cross-spectrum of two EMG signals (f_{xy}) squared and normalized by the product of the auto spectrum of each signal (f_{xx} and f_{yy}) at each frequency (λ), as follows:

$$\left| R_{xy}(\lambda) \right|^2 = \frac{\left| f_{xy}(\lambda) \right|^2}{\left| f_{xx}(\lambda) f_{yy}(\lambda) \right|} \tag{5.3}$$

Intermuscular coherence was estimated from segments of 1 s duration (i.e., 1,200 samples per segment), resulting in a frequency resolution of 1 Hz. The frequency range analyzed in this study was bounded from 0 to 55 Hz. Coherence estimates were considered statistically significant when they exceeded the significance limit of the null distribution, computed as proposed by Rosenberg et al (1989). The significance limit for zero coherence at $\alpha = 0.05$ and for the number of disjoint segments (L) was determined by the following equation:

$$Sig(\alpha) = 1 - (1 - \alpha)^{\frac{1}{L-1}} \tag{5.4}$$

In order to compare coherence estimates across participants, all estimates were z-transformed by computing the arc hyperbolic tangent transformation (*Fisher*

transformation) of the estimates as proposed by Rosemberg et al (1989) and Amjad et al (1997):

$$\tan h^{-1}(x) = 0.5 \cdot \log \frac{1+x}{1-x} \quad , \tag{5.5}$$

where x is the coherence estimate.

Analysis of the *frequency distribution* of correlated neural inputs was performed by identifying frequency intervals with significant z-scored coherence estimates across participants. Next, the analysis of the *strength of correlated neural inputs* was achieved by comparing integrals computed for the z-scored coherence estimates profiles over the frequency band of interest. A more detailed description of intermuscular coherence analysis can be found in Danna-dos-Santos et al (2010,2014), Poston et al (2010), and in *Chapter 3*.

Statistical approach. Averages and standard deviations of response variables were reported. Considering the small sample sizes, normality tests (Shapiro-Wilk tests) were performed on these variables. Since responses were found to be normally distributed, parametric tests were performed. A one-way MANOVA on factor *Age* was used to compare variables extracted from COP signals and normalized integrals of the intermuscular coherence within 0–10 Hz. In addition, a repeated measures ANOVA on factor *Muscle Pair* was performed to compare normalized integrals of the intermuscular coherence within 0–10 Hz in older adults (senior group). All parametric tests were performed by the IBM SPSS statistics software suite (version 20, IBM® SPSS®) and the level of significance fixed at 5 % ($\alpha = 0.05$) for an individual test.

5.4. RESULTS

Postural sway behavior. All participants were able to perform every trial with relative ease and without any signs of fatigue or discomfort. In addition, all participants

were able to easily sustain a vertical position of their axial skeleton with no visible postural deviations. *Table 5.1* and *Figure 5.2* show the averages across participants and respective standard deviations of all nine COP variables (*Area*, $Range_{AP}$, $Range_{ML}$, MV_{AP}, MV_{ML}, $Fmean_{AP}$, $Fmean_{ML}$, $F80_{AP}$, and $F80_{ML}$) for both control and senior groups.

As expected, the postural behavior of young and older adults was found to be dissimilar. In general, older adults swayed more and faster compared to young adults during the holding the load trials. This effect of aging on the behavior of body sway was confirmed by a one-way MANOVA on all nine COP variables ($F_{[9,12]}$ = 7.535, Wilks' Lambda = 0.150, p = 0.001). Follow-up univariate analyses confirmed significant larger values for *Area*, $Range_{AP}$, $Range_{ML}$, MV_{AP}, and MV_{ML}, and significant smaller values for $Fmean_{AP}$, $Fmean_{ML}$, $F80_{AP}$, and $F80_{ML}$ for older adults (see *p*-values in *Table 5.1*).

Table 5.1. Averages and standard deviations across participants of the **(A)** COP area (*Area*), **(B)** range (*Range_{ap}* and *Range_{ml}*), **(C)** mean velocity (*MVap* and *MVml*), **(D)** mean frequency (*Fmean_{AP}* and *Fmean_{ML}*), and **(E)** frequency at which 80% of the COP spectral power is lower than (*F80ap* and *F80ml*) for young (*control*) and older (*senior*) adults. Note: ***** indicates a significant *Age* effect ($p < 0.05$).

	Control group	Senior group	95% confidence interval of the difference	ANOVA
Area (cm^2)	0.94 ± 0.60	3.38 ± 2.43	(-4.17,-0.70)	$F_{[1,20]}$ = 8.560, p =.008*
Range$_{AP}$ (cm)	1.34 ± 0.50	3.26 ± 1.11	(-2.75,-1.09)	$F_{[1,20]}$ = 23.22, p <.001*
Range$_{ML}$ (cm)	0.69 ± 0.38	2.37 ± 1.25	(-2.58,-0.78)	$F_{[1,20]}$ =15.167, p =.001*
MVap (cm/s)	1.17 ± 0.45	1.60 ± 0.41	(-0.82,-0.05)	$F_{[1,20]}$ = 5.488, p =.030*
MVml (cm/s)	0.67 ± 0.34	0.91 ± 0.19	(-0.47,-0.00)	$F_{[1,20]}$ = 4.440, p =.048*
Fmean$_{AP}$ (Hz)	0.68 ± 0.15	0.49 ± 0.16	(0.05,0.34)	$F_{[1,20]}$ = 7.998, p =.010*
Fmean$_{ML}$ (Hz)	0.69 ± 0.23	0.37 ± 0.14	(0.16,0.48)	$F_{[1,20]}$ =17.186, p =.001*
F80ap (Hz)	1.05 ± 0.23	0.74 ± 0.29	(0.07,0.55)	$F_{[1,20]}$ = 7.185, p =.014*
F80ml (Hz)	0.96 ± 0.30	0.42 ± 0.13	(0.34,0.73)	$F_{[1,20]}$ = 33.467, p <.001*
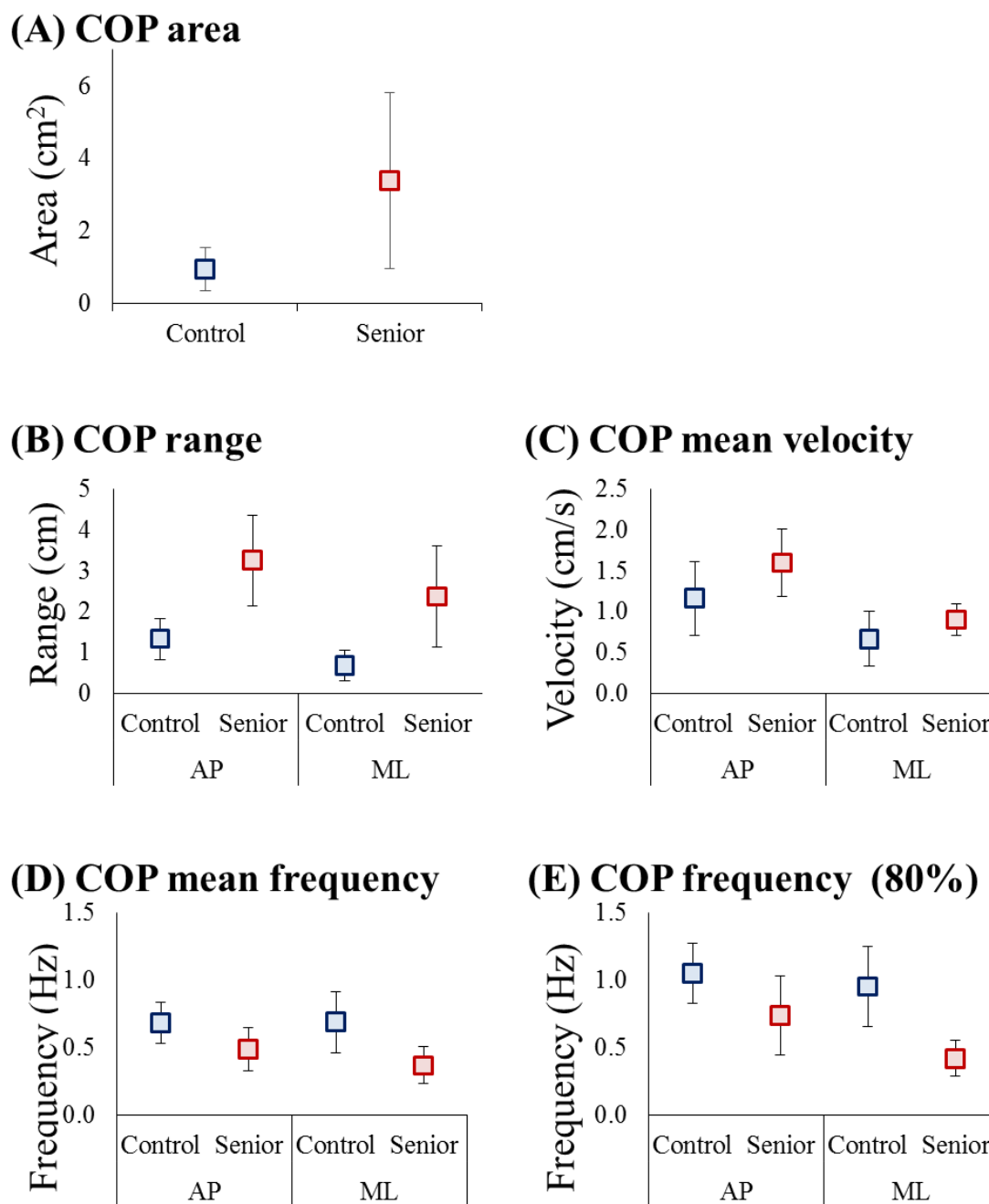
Figure 5.2. Average and standard deviation across participants of the **(A)** COP area (*Area*), **(B)** range (*Range_ap* and *Range_ml*), **(C)** mean velocity (*MVap* and *MVml*), **(D)** mean frequency (*Fmean_AP* and *Fmean_ML*), and **(E)** frequency at which 80% of the COP spectral power is below (*F80ap* and *F80ml*) for young (*control*) and older (*senior*) adults. Note: anterior-posterior direction (AP) and medio-lateral direction (ML).

Intermuscular coherence estimates. *Figure 5.3A,B,C* shows the average across participants for the z-scored coherence profiles obtained for all three pairs of muscles studied (SOL/BF, SOL/ERE, and BF/ERE) in both young and older adults (*gray* and *black* lines, respectively). Note the presence of significant estimates within the frequency band of 0–10 Hz. Due to this focal distribution, further computations of integrals aiming to quantify the strength of correlated neural inputs were performed within this frequency band and referred to as $INT_{Coh0-10Hz}$.

Table 5.2 and *Figure 5.3D* show the $INT_{Coh0-10Hz}$ of each muscle pairs in young and older adults. A one-way MANOVA revealed a significant effect of *Age* (*Control* and *Senior*) on the $INT_{Coh0-10Hz}$ ($F_{[3,18]}$ = 4.512, Wilks' Lambda = 0.571, p = 0.016). Follow-up univariate analyses confirmed significantly larger $INT_{Coh0-10Hz}$ in older adults for both muscle pairs SOL/ERE and BF/ERE, compared to young adults ($F_{[1,20]}$ = 5.582, p = 0.028, and $F_{[1,20]}$ = 5.582, p = 0.004, respectively). No significant effect of *Age* on $INT_{Coh0-10Hz}$ was observed for the muscle pair SOL/BF ($F_{[1,20]}$ = 0.902, p = 0.354). In addition, a repeated measures ANOVA revealed no effect of *Muscle pair* (SOL/BF, SOL/ERE, and BF/ERE) on the $INT_{Coh0-10Hz}$ for older adults (Wilks' Lambda = 0.960, $F_{(2,11)}$ = 0.230, p = 0.798).

Table 5.2. Averages and standard deviations across participants for the integrals of the z-scored coherence within 0–10 Hz ($INT_{Coh0-10Hz}$) for each muscle pair. Note: soleus (SOL), biceps femoris (BF), and lumbar erector spinae (ERE). Note: * indicates a significant effect of *Age* ($p < 0.05$).

	CONTROL	**SENIOR**	**95% confidence interval of the difference**	*p* **value**
$INT_{Coh0-10Hz}$ (SOL/BF)	25.74 ± 5.22	32.95 ± 22.20	(-23.0, 8.6)	.354
$INT_{Coh0-10Hz}$ (SOL/ERE)	17.68 ± 7.17	37.02 ± 23.66	(-36.4, -2.3)	.028 *
$INT_{Coh0-10Hz}$ (BF/ERE)	18.35 ± 3.05	38.08 ± 17.84	(-32.4, -7.1)	.004 *
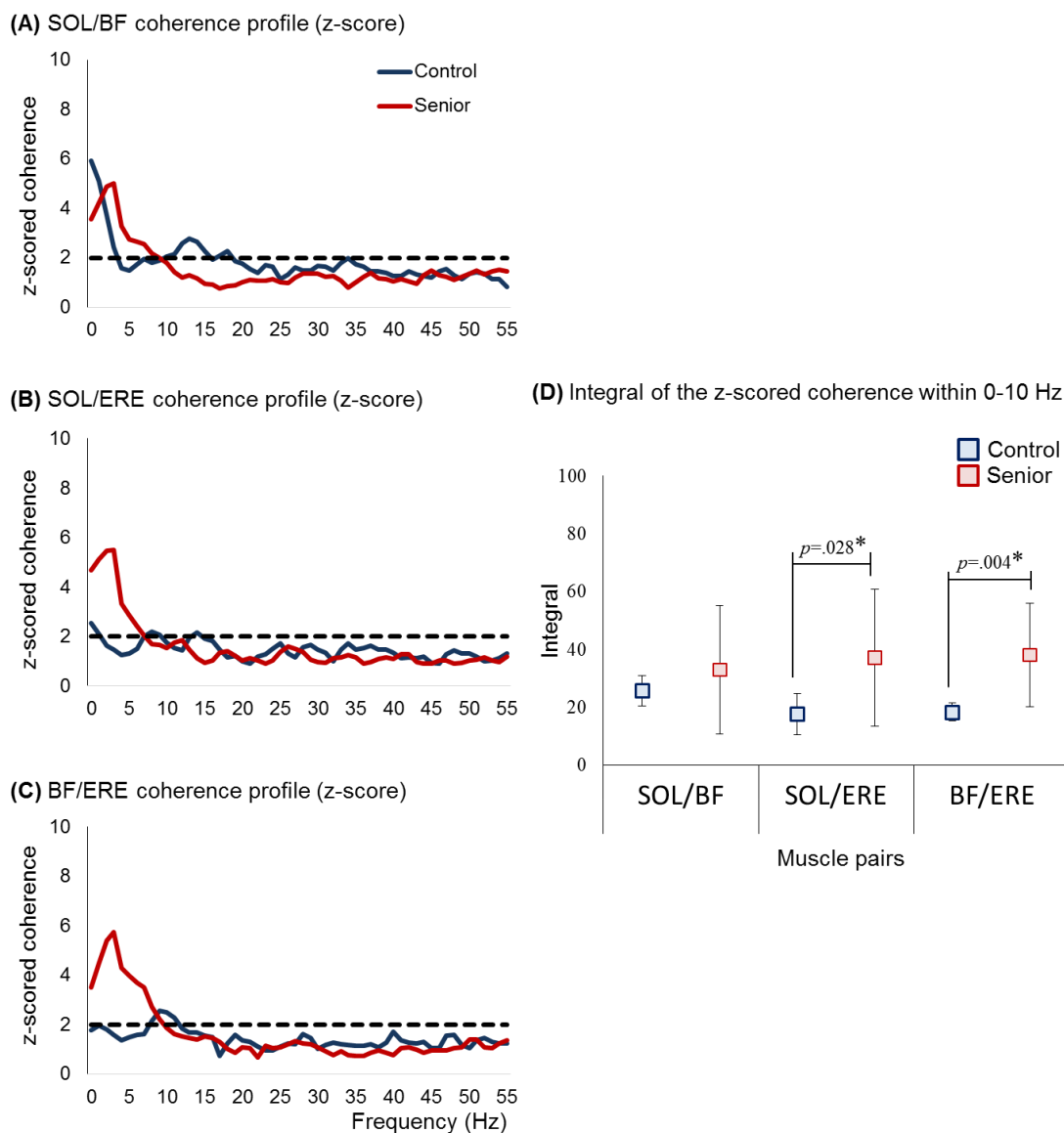
Figure 5.3. (A) (B) (C) The average z-scored coherence profiles across participants over the frequency band of 0–55 Hz for the muscle pairs SOL/BF, SOL/ERE, and BF/ERE, respectively. (D) Integrals of the z-scored coherence within 0–10 Hz ($INT_{Coh0-10Hz}$) for the three muscle pairs in young (*control*) and older (*senior*) participants. Note: * indicates a significant effect of *Age* ($p < 0.05$).

5.5. DISCUSSION

This study was designed to test the hypothesis that the natural process of aging modulates the distribution of correlated neural inputs generated by the CNS and sent to postural muscles. In fact, the results presented provide evidence to support this hypothesis.

This study corroborates changes in postural sway behavior as the individual grows older. Healthy nonfaller seniors presented a larger and faster upright body sway during the task of holding an anterior load, compared to young adults. Older adults also showed reductions in the energy content of the power spectral density of both COP signals (COPap and COPml). The effects of aging on postural sway behavior are discussed at length in *Chapter 7* of the dissertation. Many age-related factors may be responsible for changes in postural sway, such as muscle weakness and vision impairments. The results of the present study corroborate the idea of the relationship between the extensive structural changes on COP signals and the basic neural mechanisms driving postural sway suggested by Zatsiorsky and Duarte (2000). Changes in postural sway expressed by changes in the behavior of the COP displacement were accompanied by changes in motor outputs generated by the CNS. The effects of aging on the distribution of correlated neural inputs to posterior postural muscles are discussed as follows.

Results from intermuscular coherence analysis in older adults are in line with the principle proposed by Bernstein (1967), in which muscles with a similar function are grouped into functional groups controlled by the CNS. Considering that all three posterior muscles studied have the similar function of counter-acting the tendency of the body to fall forward, one could expect that correlated neural inputs to these muscles would be embedded in their EMG signals. This was the case for both young and older participants. In general, significant intermuscular coherence at lower frequency bands was found within 0–10Hz. Compared to young adults, older adults presented a stronger synchronization of the muscle pairs SOL/ERE and BF/ERE within this frequency band (0–10 Hz). In addition to the increased coherence for these muscle pairs, significant correlated neural inputs for these muscle pairs emerged in older adults. These observations suggest that older adults also generate correlated neural inputs to postural muscles to control posture. However, the strategies used by older individuals to organize synergistic muscle groups seem to change.

Note that the general spectra of the coherence at lower frequency bands for all three muscle pairs became similar in older adults: a significant intermuscular coherence within 0–10 Hz with similar peak magnitude around 4 Hz (*Figure 5.3ABC*). This result

may indicate the tendency of the aging CNS to act conservatively by reinforcing a synergistic pattern with distal and proximal muscles, corroborating the hypothesis of older adults using ankle and hip strategies to control upright posture. This finding may also be associated with the increase in cocontraction of all postural muscles during perturbed situations observed in older adults. Lee et al (2015) reported age-related changes on compensatory postural adjustments (CPAs) under the continuous perturbed task of pushing a load. They observed an agonist-antagonist co-activation as a compensatory mechanism to overcome balance deficits as the individual grows older, whereas young adults showed reciprocal muscle activation patterns as their CPAs to the continuous perturbation. It is important to note, however, that results showed in the present study comes from coherence analysis at lower frequency bands, rather than from frequency analysis at higher frequency bands involving firing rates of muscle activation.

Considering that the experimental task of upright stance holding a load is, by nature, a continuous perturbation trying to displace the axial body forward, the results also suggest that older adults adopted similar distributions of correlated neural inputs to all three posterior muscles as a compensatory adjustment to the constant mechanical perturbation imposed by the load. Even though this suggestion is merely speculative, it finds a basis for its rationale in the fact that many of the sensory functions become partially impaired after the 6th decade of life. This factor alone can induce longer delays to the generation of a motor response intended to counter-act mechanical perturbation. In addition, the EMG activity in postural muscles increases asymmetrically in response to perturbation (Tsai et al 2014). This asymmetric pattern might be another contributing factor to unstable postural responses in older individuals. Therefore, changes in the organization of postural muscle synergies may be the solution adopted by the aging CNS to compensate for its progressive sensorimotor impairments.

5.6. CONCLUSIONS

This exploratory study advances the understanding of multi-muscle control principles based on the distribution of correlated neural inputs to postural muscles.

Findings using a novel approach (the Intermuscular Coherence Analysis) to detect age-related changes on postural muscle control suggest the presence of correlated neural inputs within 0–10 Hz for both young and older individuals. Interestingly, older adults showed significant synchronization at lower frequency bands not only for the most distal muscle pair (SOL/BF), as observed in young adults. Moreover, they revealed significant synchronization at lower frequency bands for two other muscle pairs: SOL/ERE and BF/ERE. This finding not only corroborates the functional drive of the formation of correlated inputs, it also corroborates the use of both ankle and hip strategies by older adults to control upright stance.

Therefore, it is suggested that neurophysiological age-related changes affect the organization and strength of neural drive to multiple postural muscles. Further studies are necessary to explore how the CNS organizes multiple muscles into functional groups to control upright stance and what the effects of the natural aging on this process are.

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CHAPTER 6

The effects of aging on the distribution and strength of
correlated neural inputs to postural muscles during unperturbed stance

6.1.   INTRODUCTION

Loss of skeletal muscle strength is a commonly recognized consequence of the aging of the human neuromuscular system (Doherty 2003). Previous studies have reported age-related declines in the strength of skeletal muscles to be between 20% and 40% for individuals after their sixth decade of life, and up to 50% for those in their ninth decade (Larsson et al 1979, Murray et al 1980,1985). The causes of this progressive weakening include the progressive loss of muscle mass and strength (sarcopenia), reductions in the average size of the remaining muscle fibers, and reposition of contractile proteins by connective tissue (Zimmerman et al 1993, Bemben 1998). In addition, these declines occur in a non-homogenous fashion, where distal muscles weaken faster compared to proximal groups (Vitasalo et al 1985, Nakao et al 1989, Shinohara et al 2003). The fact that some muscles weaken faster than others creates a scenario where forces applied to body segments become unbalanced. As a result, motor performance is affected and the controller (CNS) is forced to adapt its muscle control strategies. In fact, this progressive and unbalanced decline in muscle strength is considered to be one of the contributing factors to the development of motor impairments among older adults (O'Sullivan and Schmitz 2006, Shumway-Cook and Woollacott 2011), leading to potential episodes of fall and its consequent comorbidities, such as hip fractures and traumatic brain injuries.

It is also important to emphasize previous observations indicating that the healthy aged body is indeed able to adapt its postural strategies to cope with such age-related declines (Inglin and Woollacott 1988, Woollacott et al 1988, Tang and Woollacott 1998). A typical neuromotor adaptation is the agonist/antagonist cocontraction mechanism adopted by older adults either to maintain unperturbed standing or to prepare and respond to an external mechanical perturbation applied to the body (Inglin and Woollacott 1988,

Woollacott et al 1988, Tang and Woollacott 1998, Laughton et al 2003, Lee et al 2015). Another adaptive change in motor behavior reported in the literature is related to the postural reaction to a sudden translation of the body's base of support. In general, young adults use the ankle strategy in response to such small perturbation, while older adults might also generate a hip strategy or even a step strategy to avoid falls and restore equilibrium under the same challenging situation. The larger proximal joint rotation and larger center of mass sway in older adults in response to postural perturbations, reported by Tsai et al (2014), corroborate the use of the hip strategy in conjunction with the ankle strategy by older adults.

Although these behavioral adaptations have been well reported, the mechanisms underlying the strategies adopted by the aging CNS to generate and distribute the neural commands to multiple postural muscles have not yet been explored. Therefore, the present study was centered on the hypothesis that the CNS uses correlated neural inputs to coordinate the activation of multiple muscles forming a synergistic group (Farmer 1998, De Luca and Erim, 2002, Santello and Fuglevand 2004, Semmler et al 2004, Johnston et al 2005, Winges et al 2008, Boonstra et al 2009, Poston et al 2010, Danna-dos-Santos et al 2010,2014,2015). According to this idea, the synchronization of neural oscillations at lower frequency bands may also be the mechanism used by the aging CNS to achieve large-scale integration among its cortical and subcortical components, including those involved in the generation and control of movements. According to Farmer (1998) and De Luca and Erim (2002), traces of underlying synchronization at lower frequency bands occurring within the CNS are embedded in the electromyographic signals of the targeted muscles and synchronization features, such as the coherence estimations between pairs of electromyography (EMG) signals, can be used to investigate the formation (or dissolution) of multi-muscle synergies. In fact, the intermuscular coherence analysis at lower frequency bands has been successfully used to investigate the distribution of correlated neural inputs to skeletal muscles during the execution of whole-body tasks (Boonstra et al 2008,2009,2015). *Chapters 3* and *4* also indicated the presence of correlated neural inputs to postural muscles crossing the ankle, knee, hip, and lower lumbar intervertebral joints during unperturbed standing in healthy young adults. Taken together, the findings corroborate observations from Boonstra and colleagues, indicating

that correlated neural inputs to skeletal muscles are not only present, but they also have specific spatial distributions and strength .

To further advance knowledge on age-related changes in the organization of postural multi-muscle synergies, the present study aimed to investigate the distribution and strength of correlated neural inputs among six aging postural muscles (Tibialis Anterior - TA, Soleus - SOL, Rectus Femoris - RF, Biceps Femoris - BF, Rectus Abdominis – RA, and Erector Lumbar Spinae – ERE). These muscles were selected for several reasons. First, these muscles act mainly in the sagittal plane, which is the plane where most of the body sway happens during upright stance. Second, they have a crucial role on the control of the vertical posture since they cross the ankle, knee, hip, and lower trunk joints. They are also considered the primary movers for these joints. Third, these muscles act on joints placed at different proximity levels of the body's center of mass and base of support, making them ideal to study the spatial distribution of correlated neural inputs to multiple muscles. And fourth, these muscles have been previously reported as the primary components of synergistic groups (often referred to as muscle modes or M-modes) in healthy young adults, acting in the stabilization of the center of pressure (COP) trajectory within the base of support (Krishnamoorthy et al 2003a,b, Danna-dos-Santos et al 2007,2008,2009).

Based on the observations that postural muscles are organized synergistically according to their mechanical function, it was expected that aging postural muscles would exhibit signs of correlated neural inputs in the form of significant intermuscular coherence at lower frequency bands (*hypothesis #1*). In addition, it was expected that the physiological neuromuscular decline due to the natural process of aging would affect the organization of muscle synergies. More specifically, the researcher expected (a) a significant general increase in the strength of correlated neural inputs to postural muscles in older adults reflected in significant increased intermuscular coherence at lower frequency bands across postural muscle pairs (*hypothesis #2*); and (b) a different distribution of correlated neural inputs between distal and proximal muscles in older adults, compared to young adults (*hypothesis #3*).

6.2. METHODS

Participants. Ten (10) healthy young adults (mean age 26.8 years old and *SD* = 2.7) and ten (10) healthy older adults (mean age 68.7 years old and *SD* = 3.5) volunteered in this study. The exclusion criteria for both *Control* and *Senior* groups included previous history of sensory, musculoskeletal, neurological, or cardiopulmonary disorder; surgeries; or substance abuse. In addition, the senior group included only nonfallers with the age ranging 65 to 74 years (older adults). Therefore, exclusion criteria for the senior group included history of falls (fallers) and age between 75–84 years (older old adult) or over 85 years (oldest old adult).

All volunteers gave their informed consent based on the procedures approved by the local Institutional Review Board and conforming to The Declaration of Helsinki prior to their participation in the study. *Table 6.1* describes the general demographics of the participants.

Table 6.1. General demographics of the participants (average and standard deviation across participants).

Group	Age (years)	Weight (kg)	Height (cm)	Gender
Control	26.8 ± 2.7	80.6 ± 22.0	175.0 ± 12.7	4 females and 6 males
Senior	68.7 ± 3.5	71.9 ± 7.7	168.5 ± 8.5	5 females and 5 males

Apparatus. Six active surface electrodes (Delsys Bagnoli single differential DE-2.1) were used to record electromyographic signals (EMG) of the following muscles: tibialis anterior (TA), soleus (SOL), rectus femoris (RF), biceps femoris (BF), rectus abdominis (RA), and lumbar erector spinae (ERE). These electrodes were placed on the right side of the body over the muscle bellies according to manufacturer instructions and Criswell (2010). The distance between electrode pairs was kept at 1 cm, and total area of surface recording was 10 mm^2 for each electrode. A reference electrode was placed over the lateral aspect of the fibular malleolus. EMG signals were amplified (1,000×) and band-pass filtered (6–450 Hz). Signals from the control group were sampled at 1200 Hz

and signals from the senior group were sampled at 2000 Hz, all with a 12-bit resolution. *Figure 6.1* illustrates the position of the electrodes.
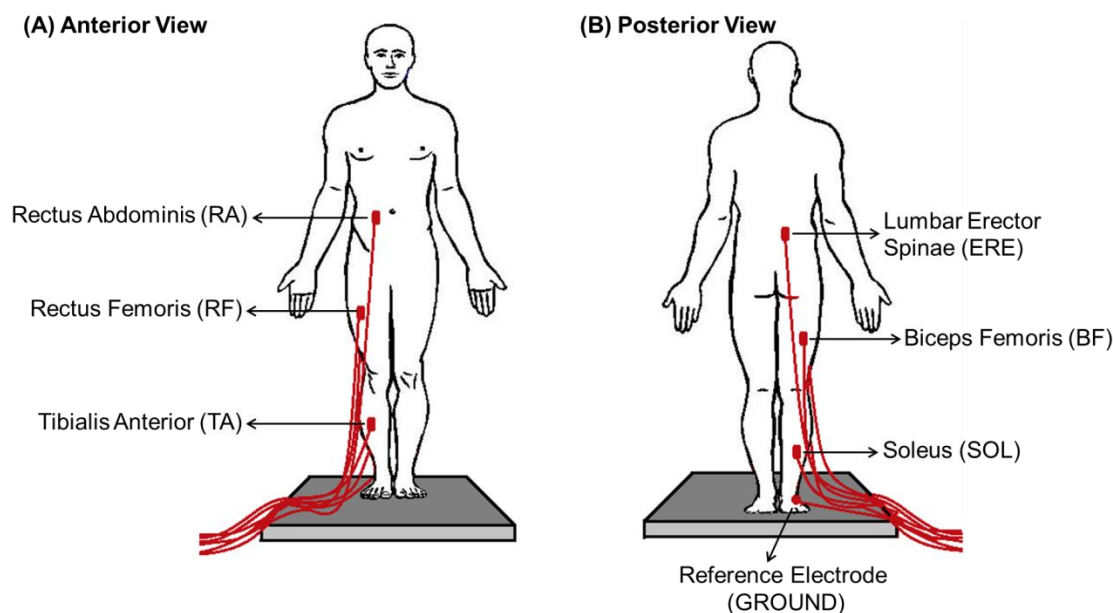
Figure 6.1. Representation of the electrodes placed on **(A)** anterior and **(B)** posterior postural muscles. Note: tibialis anterior (TA), soleus (SOL), rectus femoris (RF), biceps femoris (BF), rectus abdominis (RA), lumbar erector spinae (ERE), and reference electrode (GROUND).

Experimental procedures. All participants performed two tasks: unperturbed bipedal stance with open eyes (*BOE* or *Vision* condition*)* and unperturbed bipedal stance with closed eyes (*BCE* or *No Vision* condition). For both tasks, participants were instructed to stand barefoot with their feet 10 cm apart and parallel for 60 seconds. For the *BOE* task, participants were asked to focus their vision on a physical static point placed at eye level and at a distance of approximately two meters; the *BCE* task, however, required participants to close their eyes. An interval of 60 seconds was provided between the two tasks. The total duration of the experiment was approximately 30 minutes.

6.3. DATA PROCESSING

EMG signals recorded from 6 muscles were submitted to off-line analyses with a series of custom-written software routines in Matlab R2012b (MathWorks Inc., Natick, MA). All EMG signals were down sampled to 1000 Hz and, subsequently, filtered with a band-pass (6–450 Hz), second-order, zero lag Butterworth filter. The correlation of muscle activation in the frequency domain was quantified using similar procedures to Boonstra et al (2009) and Poston et al (2010). Specifically, intermuscular coherence was estimated between all pairs of filtered EMG signals recorded from each participant and organized according to their relationship (*Table 6.2*). The pairs of muscles analyzed in this study included pairs formed by solely anterior muscles, solely posterior muscles, antagonist muscles, and other pairs formed by one anterior and one posterior non-antagonist muscles.

Table 6.2. Fifteen muscle pairs formed by solely anterior muscles, solely posterior muscles, antagonist muscles, or mixed muscles (one anterior and one posterior, non-antagonist, muscles). Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

	Muscle pair	**Group (anatomic position)**
1	TA – RF	
2	TA – RA	Anterior
3	RF – RA	
4	SOL – BF	
5	SOL – ERE	Posterior
6	BF – ERE	
7	TA – SOL	
8	RF – BF	Antagonist
9	RA – ERE	
10	TA – BF	
11	TA – ERE	
12	RF – SOL	Mixed
13	RF – ERE	
14	RA – SOL	
15	RA – BF	

Single-pair coherence estimate. The intermuscular coherence was estimated for pairs of EMG signals by normalizing the cross-spectrum of the two signals (f_{xy}) squared and normalized by the product of the auto spectrum of each signal (f_{xx} and f_{yy}) at each frequency (λ), that is:

$$\left|R_{xy}(\lambda)\right|^2 = \frac{\left|f_{xy}(\lambda)\right|^2}{\left|f_{xx}(\lambda)f_{yy}(\lambda)\right|} \tag{6.1}$$

Intermuscular coherence estimates were obtained from non-overlapping 1 second data segments, resulting in a frequency resolution of 1 Hz. The initial frequency range analyzed in this study was bounded from 0 to 55 Hz. However, to avoid the inclusion of the mechanical effects of sway and its coupling to the low-frequency content previously observed by Mochizuki et al (2006), all analyzes excluded the frequency band of 0–1 Hz.

Coherence estimates were considered statistically significant when they exceeded the significance limit of the null distribution, as proposed by Rosenberg et al (1989). The significance limit at $\alpha = 0.05$ was determined by

$$Sig(\alpha) = 1 - (1-\alpha)^{\frac{1}{L-1}}, \tag{6.2}$$

where L is the number of disjoint segments.

In order to compare coherence estimations across participants and different experimental conditions, all estimates were z-transformed using the Fisher transformation, as proposed by Rosemberg et al (1989) and Amjad et al (1997):

$$\tan h^{-1}(x) = 0.5 \cdot \log \frac{1+x}{1-x}, \tag{6.3}$$

where x is the coherence estimate.

Analysis of the *frequency distribution* of correlated neural inputs was achieved by identifying frequency intervals showing significant z-scored coherence estimations. The

quantification of the *strength of correlated neural inputs* was achieved by computing the integrals within the frequency intervals of interest. Since significant estimates across participants were more prevalent within the frequency interval of 1–10 Hz, the *strength* of correlated neural inputs was obtained by comparing integrals computed for the z-scored coherence estimate profiles over the mentioned frequency bands of interest ($INT_{coh1-10}$).

Pooled coherence estimations. In addition to the fifteen intermuscular coherence analyses, four pooled coherence analyses were performed separately. The first analysis included the three pairs formed by posterior muscles (Posterior group: SOL/BF, SOL/ERE, and BF/ERE); the second analysis included the three pairs formed by anterior muscles (Anterior group: TA/RF, TA/RA, and RF/RA); the third analysis included the three pairs formed by antagonist muscles (Antagonist group: TA/SOL, RF/BF, and RA/ERE); and the fourth analysis included the remaining six pairs formed by one posterior and one anterior non-antagonist muscles (Mixed group: TA/BF, TA/ERE, RF/SOL, RF/ERE, RA/SOL, and RA/BF).

Since pooled coherence estimates are considered a weighted average of individual coherence estimates, they can be used to increase statistical power. That is, it is assumed that all muscles forming a single muscle mode share the same neural inputs. Estimates of pooled coherence were obtained as proposed by Amjad et al (1997):

$$\frac{\left| \sum_{i=1}^{k} f_{xy}(\lambda) L_i \right|^2}{\left(\left(\sum_{i=1}^{k} f_{xx}(\lambda) L_i \right) \left(\sum_{i=1}^{k} f_{yy}(\lambda) L_i \right) \right)} \tag{6.4}$$

Analysis of the *frequency distribution and strength* of correlated neural inputs obtained from pooled coherence estimations was also based on the determination of frequency bands of interest and on the computation of integrals within specific frequency bands.

Statistical Approach. Medians across participants of the integrals computed from the z-scored coherence estimates profiles between 1 and 10 Hz ($INT_{Coh1-10Hz}$) are reported in *Tables 6.3* and *6.4*. Considering the sample size, Shapiro-Wilk tests were performed to test for normality of the coherence variable. Since $INT_{Coh1-10Hz}$ did not follow normal distribution, non-parametric tests for comparisons on the variable $INT_{Coh1-10Hz}$ were performed. More specifically, Mann-Whitney U tests were used to investigate the effect of *Age* (*Control* and *Senior* groups) on $INT_{Coh1-10Hz}$ for each condition (*Vision* and *No Vision*), whereas paired tests (Wilcoxon signed-rank tests) were used to investigate the effect of *Vision* (*BOE* or *Vision* and *BCE* or *No Vision* condition) on $INT_{Coh1-10Hz}$ for each age group (*Control* and *Senior*).

Statistical tests were performed using the IBM SPSS statistics software suite (version 22, IBM® SPSS®). Since multiple comparisons were performed, the significant level was adjusted at 2% ($\alpha = 0.02$) for individual tests.

6.4. RESULTS

Since most results regarding intermuscular coherence in healthy young adults have already been presented and discussed in *Chapters 3* and *4*, this chapter will focus on the findings regarding the distribution and strength of correlated neural inputs in older adults and the effects of vision on the muscle activation in these individuals.

6.4.1. Intermuscular coherence estimations: pooled analyses

Figure 6.2 shows the average pooled coherence profiles (transformed into z-scores) across young and older participants computed for each muscle group (anterior posterior, antagonist, and mixed) and under both vision and no vision conditions (BOE and BCE, respectively). Considering significant coherences those with values above the significance level (*dashed line*), note that significant pooled estimations were found mostly within the frequency interval of 0 to 10 Hz. However, the frequency band 0–1 Hz

was discarded to avoid the inclusion of the coupling of mechanical effects of body sway to the EMG signals. Therefore, the frequency band of interest in the present study was bounded between 1 and 10 Hz.

In general, there was an increase in the strength of correlated neural inputs to multiple postural muscles with age. The effect of *Age* on the pooled coherence estimations is shown in *Figure 6.3* and *Table 6.3*. A significant increase in the strength of the correlated neural inputs within 1–10 Hz in older adults was observed for the anterior, posterior, and antagonist muscle groups. Mann-Whitney U tests confirmed the effect of *Age* on $INT_{Coh1-10Hz}$ for the groups formed by anterior, posterior, and antagonist muscles during upright stance with eyes open, while no significant effect was detected for the mixed muscle group (see *p*-values in *Table 6.3*) The increase in the coherence strength of both anterior and posterior muscle groups in older adults with eyes open was also observed when they performed unperturbed stance with eyes closed. Mann-Whitney U tests revealed a significantly stronger coherence for the anterior and posterior muscle groups in older adults performing unperturbed stance with eyes closed, compared to that for young adults ($p < 0.001$ for both). No significant effect of *Age* on $INT_{Coh1-10Hz}$ during the *BCE* condition was detected for either antagonist or mixed muscle groups ($p = 0.190$ and $p = 0.739$, respectively).

Regarding visual information (*BOE* or *Vision* and *BCE* or *No Vision* conditions), seniors presented similar strength of correlated neural inputs to anterior, posterior, antagonist, and mixed muscle groups when visual input was not allowed. Wilcoxon signed-rank tests confirmed no *Vision* effect for the anterior, posterior, or mixed muscle groups ($p = 0.646$, $p = 0.878$, $p = 0.022$, and $p = 0.333$, respectively). On the other hand, healthy young controls presented similar coherence strength for the antagonist and mixed muscle groups and decreased the strength of correlated neural inputs to their anterior and posterior muscle groups. Wilcoxon signed-rank tests confirmed a significant effect of *Vision* (*BOE* and *BCE* conditions) on $INT_{Coh1-10Hz}$ for both anterior and posterior muscle groups ($p = 0.005$ for both) and no effect for either antagonist or mixed muscle groups ($p = 0.799$ and $p = 0.878$, respectively).

POOLED COHERENCE (z-score)

CONTROL

(A) Anterior muscle group

(B) Posterior muscle group

(C) Antagonist muscle group

(D) Mixed muscle group

SENIOR

(E) Anterior muscle group

(F) Posterior muscle group

(G) Antagonist muscle group

(H) Mixed muscle group
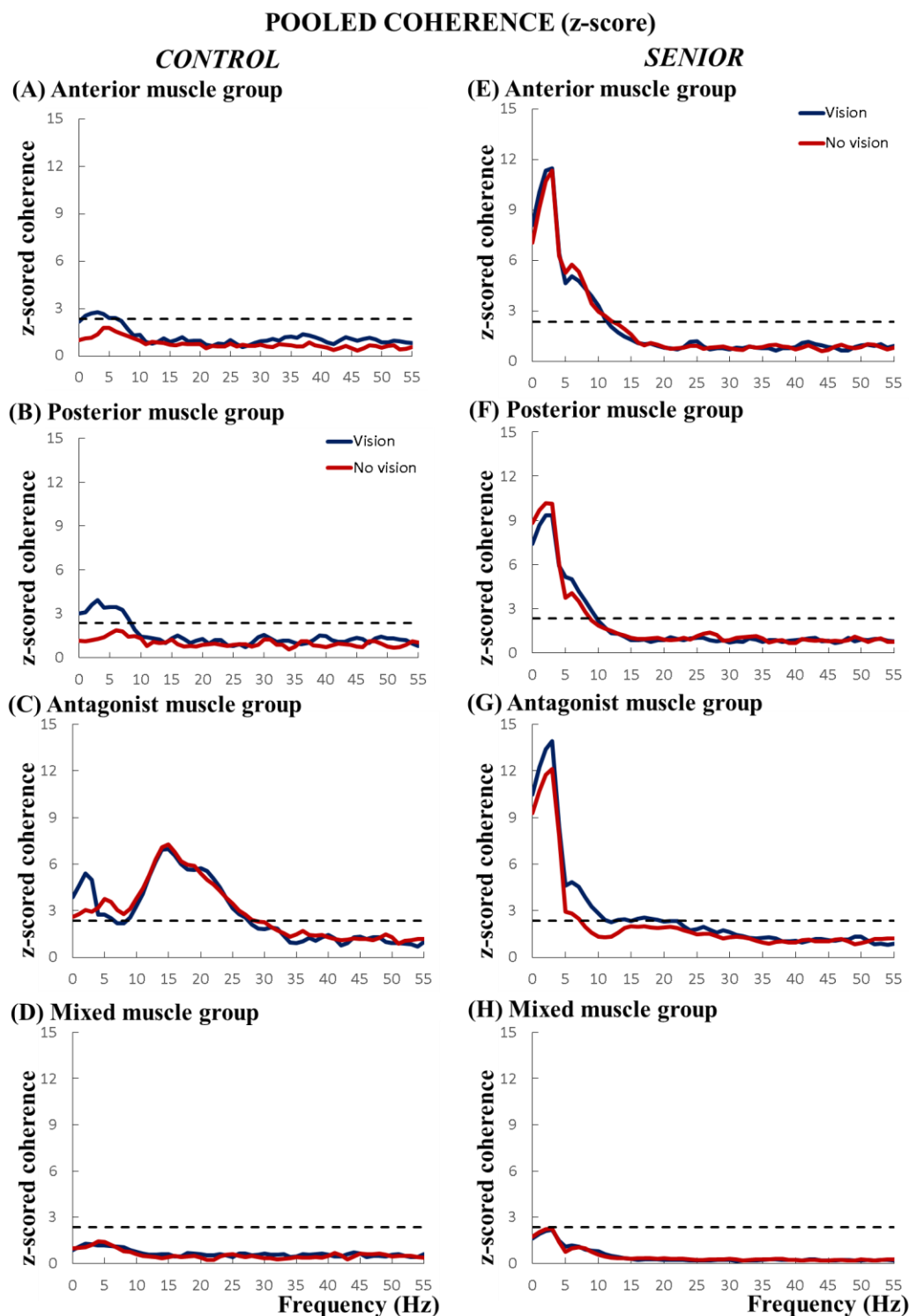
Figure 6.2. Average across young and older participants (*Control* and *Senior* groups, respectively) of the pooled coherence profiles obtained for all pairs formed between posterior, anterior, antagonist, and mixed muscles during unperturbed stance with and without vision (*BOE* and *BCE* conditions, respectively). Note: *dashed line* represents the significance level.

Integrals of the z-scored pooled coherence within the frequency band 1-10 Hz $(INT_{Coh1\text{-}10Hz})$
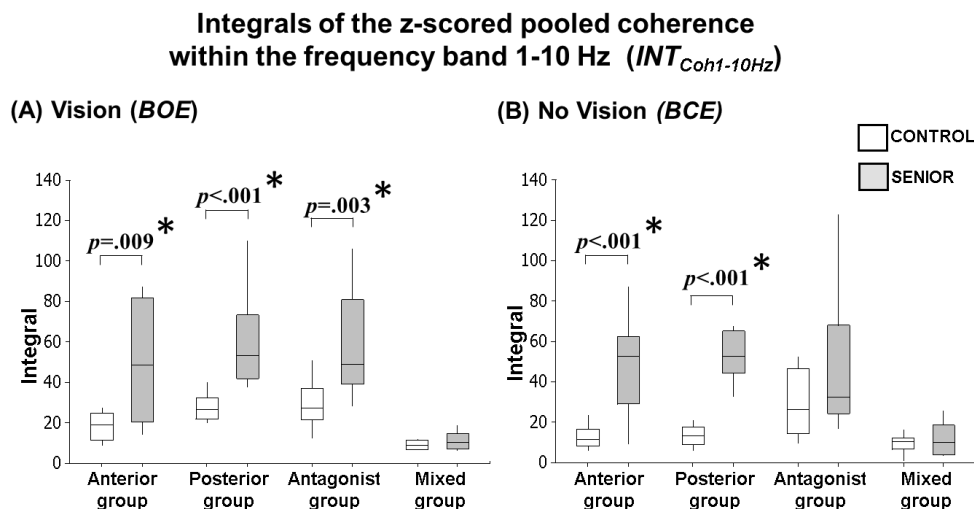
Figure 6.3. Box-plots of the integrals of the z-scored pooled coherence profiles within the 1–10Hz frequency band ($INT_{Coh1\text{-}10Hz}$) for each muscle group (anterior, posterior, antagonist, and mixed) of young and older adults (*control* and *senior* groups, respectively) during unperturbed stance with and without vision (*BOE* and *BCE* conditions). Note: * indicates *age* effect ($p < 0.02$). *Vision* effect is not presented here.

Table 6.3. Median of the integrals of the z-scored pooled coherence computed over the frequency interval of 1–10 Hz ($INT_{Coh1\text{-}10Hz}$) for each muscle group (anterior, posterior, antagonist, and mixed) of young and older adults (*control* and *senior* groups, respectively) during unperturbed stance with and without vision (*BOE* and *BCE* conditions). Note: * indicates *Age* effect ($p < 0.02$). *Vision* effect is not presented here.

Muscle group	$INT_{Coh1\text{-}10Hz}$					
	BOE (Vision)		95% confidence interval of the difference	BCE (No Vision)		95% confidence interval of the difference
	Control	Senior	and *p* value	Control	Senior	and *p* value
Anterior	19.72	48.52	(-50.8,-8.4) .009 *	11.42	52.70	(-52.1,-20.5) <.001 *
Posterior	26.66	53.46	(-46.2,-15.6) <.001 *	13.12	52.80	(-62.2,-28.5) <.001 *
Antagonist	26.96	48.94	(-64.1,-6.0) .003 *	26.23	32.26	(-47.9,6.4) .190
Mixed	8.91	10.45	(-7.6,3.0) .579	10.53	9.89	(-8.0,3.9) .739

6.4.2. Intermuscular coherence estimations: single-pair analyses

Figures 6.4 and *6.5* show the averaged intermuscular coherence spectra (expressed as z-scores) across participants computed separately for all 15 muscle pairs within the frequency band of 0–55 Hz. According to these spectra, there is a clear prevalence of significant estimates within the frequency band of 1–10 Hz. Also, values are clearly greater for the senior group. Note the significant coherence between RA and ERE (an antagonist muscle pair) not only within 0–10 Hz, but also within 10–30 Hz.

Figure 6.6 and *Table 6.4* present the medians across participants for the integrals of the z-scored coherences computed for each muscle pair within the frequency band of 1–10 Hz ($INT_{Coh1-10Hz}$). There is a general increase in the strength of correlated neural inputs to postural muscles with age, in particular for the anterior, posterior, and antagonist muscle pairs. Mann-Whitney U tests showed significant effects of *Age* (control and seniors) on the variable $INT_{Coh1-10Hz}$ for all three anterior and antagonist muscle pairs during both *BOE* and *BCE* conditions, except for the anterior muscle pair TA/RF under the *BCE* condition. Mann-Whitney U tests also showed the effect of *Age* on the variable $INT_{Coh1-10Hz}$ for the posterior pairs SOL/BF and SOL/ERE pairs during the *BOE* condition, and for all three posterior muscle pairs (SOL/BF, SOL/ERE, and BF/ERE) during the *BCE* condition. Regarding mixed pairs, Mann-Whitney U tests showed that only the muscle pair RA/BF during the *BOE* condition and the muscle pair RA/SOL during the *BCE* condition presented significantly stronger coherence in older adults, compared to young adults (see all *p*-values for the effect of *Age* on $INT_{Coh1-10Hz}$ in *Table 6.4*).

Regarding visual information, *Table 6.5* presents the *Vision* effect of the intermuscular coherence for each group (*control* and *senior*). Seniors presented no significant changes on the coherence strength between 1 and 10 Hz when they closed their eyes for any muscle pair. Wilcoxon signed-rank tests confirmed no significant effects of *Vision* (*BOE* and *BCE*) on the variable $INT_{Coh1-10Hz}$ for all anterior, posterior, antagonist, and mixed muscle pairs. The effects of temporary removal of visual input in young adults were presented in *Chapter 5*. As a reminder, Wilcoxon signed-rank tests revealed that young adults presented a significant decrease in the strength of correlated

neural inputs in the frequency range of 1–10 Hz for the anterior muscle pairs TA/RF and TA/RA and for the three posterior muscle pairs SOL/BF, SOL/ERE, and BF/ERE. No significant *Vision* effect was found for the other muscle pairs on the variable $INT_{Coh1-10Hz}$ (see all *p*-values for the effect of *Vision* on $INT_{Coh1-10Hz}$ in *Table 6.5*).
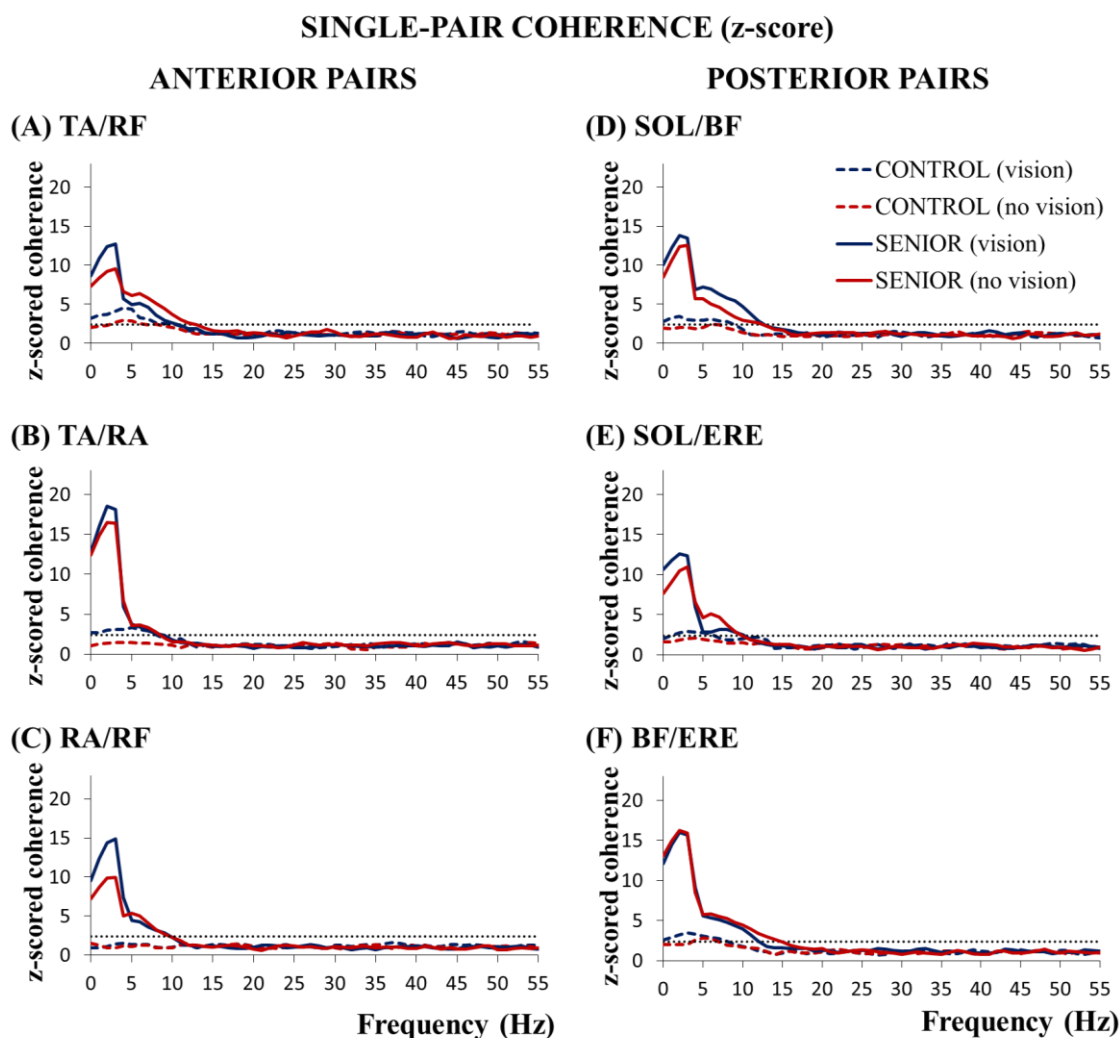
Figure 6.4. Average across young and older participants (*Control* and *Senior* group, respectively) of the coherence profiles obtained for muscle pairs formed by anterior and posterior muscles during unperturbed stance with and without vision (*BOE* or *Vision*, and *BCE* or *No Vision* conditions, respectively). Note: *dashed line* represents the significance level. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

SINGLE-PAIR COHERENCE (z-score)
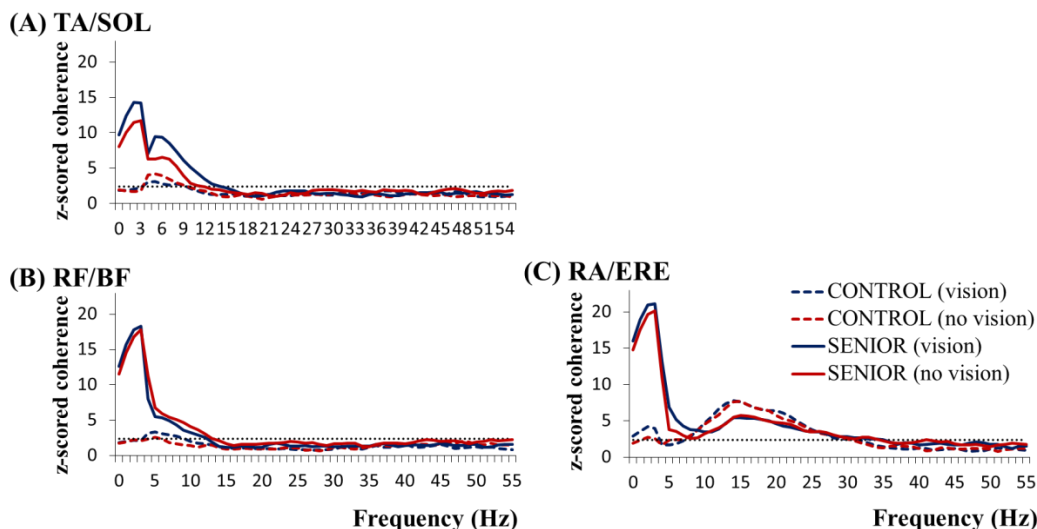
ANTAGONIST PAIRS
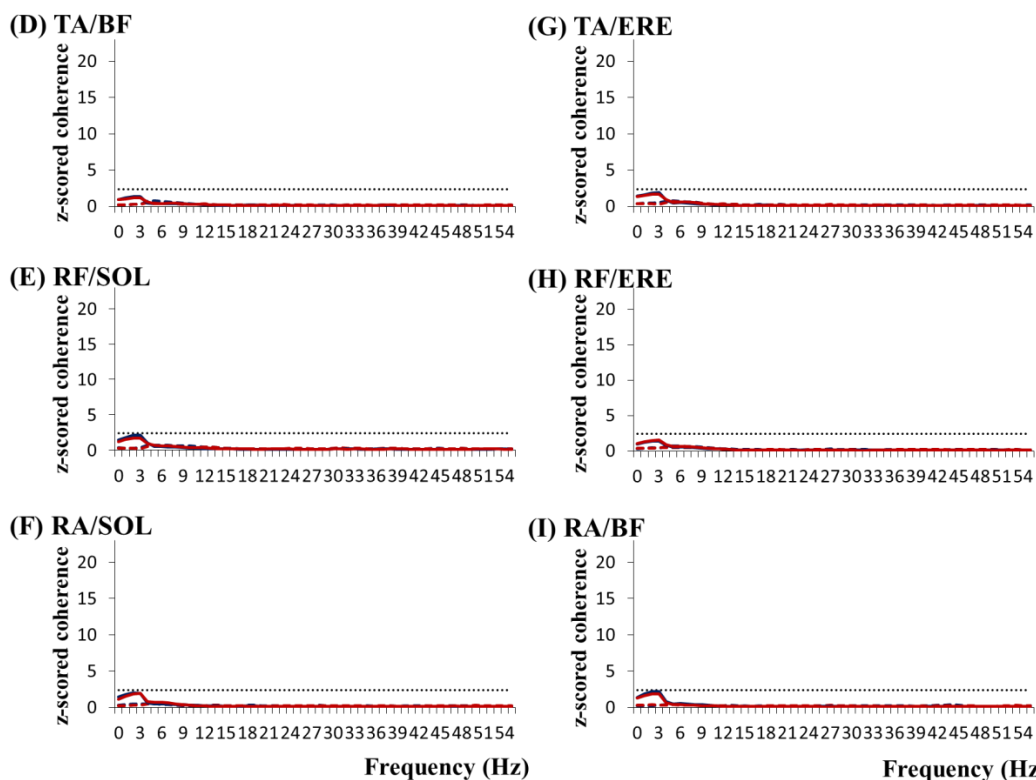
Figure 6.5. Average across young and older participants (*Control* and *Senior* group, respectively) of the coherence profiles obtained for muscle pairs formed by antagonist and mixed muscles during unperturbed stance with and without vision (*BOE* or *Vision*, and *BCE* or *No Vision* conditions, respectively). Note: *dashed line* represents the significance level. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).
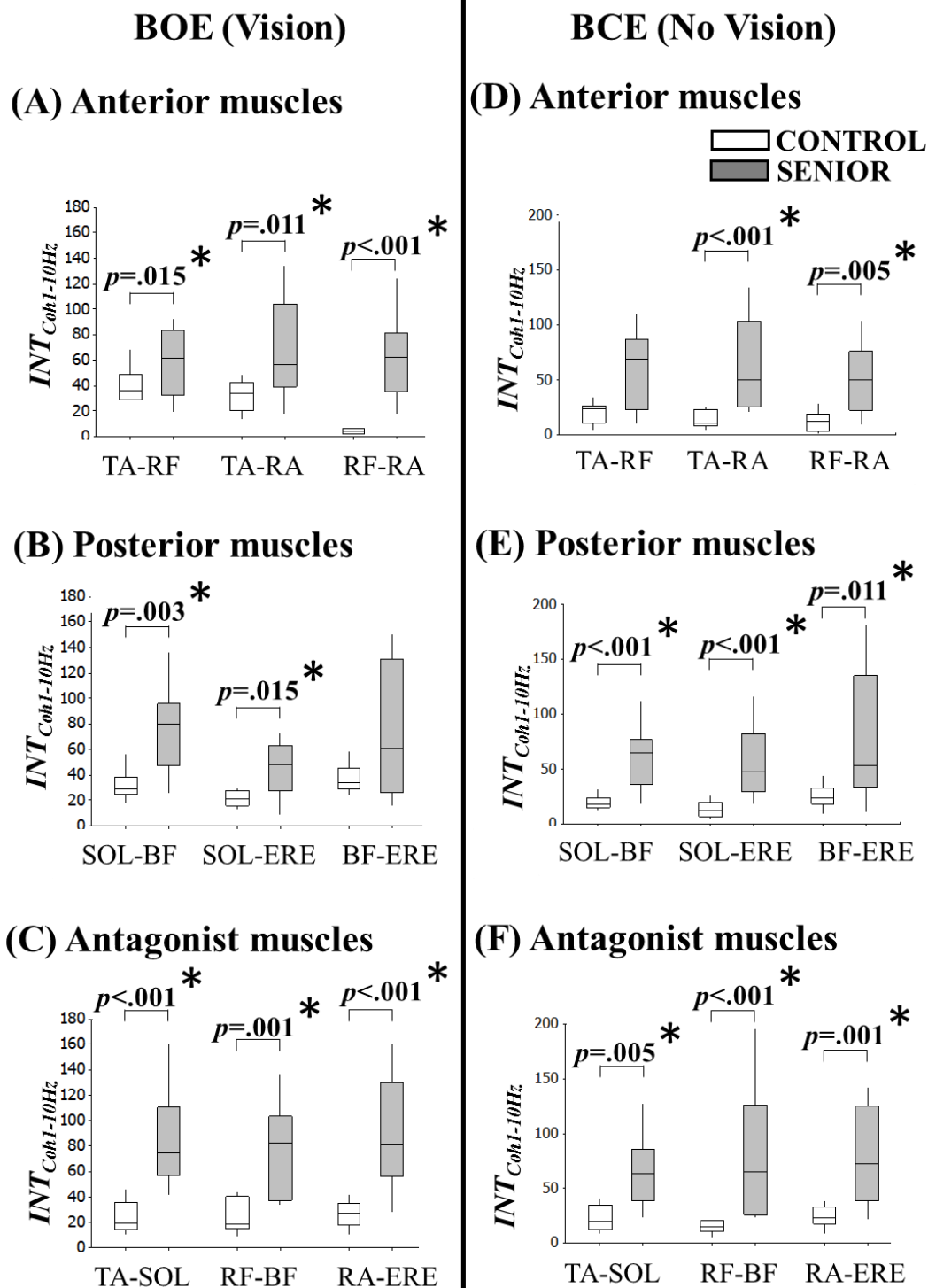
Figure 6.6. Box-plots of the integrals of the z-scored coherence profiles within the 1–10Hz frequency band ($INT_{Coh1-10Hz}$) for each muscle pair of young and older adults (*control* and *senior* groups, respectively) during unperturbed stance with and without vision (*BOE* and *BCE* conditions). Note: * indicates *Age* effect ($p < 0.02$). *Vision* effect is not presented here. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

Table 6.4. Median of the integrals of the z-scored intermuscular coherence computed over the frequency interval of 1–10 Hz ($INT_{Coh1\text{-}10Hz}$) for each muscle pair of young and older adults (*control* and *senior* groups, respectively) during unperturbed stance with and without vision (*BOE* and *BCE* conditions). Note: * indicates *Age* effect ($p < 0.02$). *Vision* effect is not presented here. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

| | $INT_{Coh1\text{-}10Hz}$ | | | | | |
| | **BOE (Vision)** | | | **BCE (No Vision)** | | |
	Control	*Seniors*	95% confidence interval of the difference and *p* value	*Control*	*Seniors*	95% confidence interval of the difference and *p* value
Pairs formed by anterior muscles						
TA/RF	36.09	61.69	(-39.5,-6.4) .015 *	23.82	68.92	(-63.4,-14.9) .023
TA/RA	33.66	56.56	(-60.9,-8.6) .011 *	11.17	49.65	(-76.3,-22.0) <.001 *
RF/RA	4.48	62.19	(-77.3,-31.3) <.001 *	11.91	49.49	(-61.3,-15.9) .005 *
Pairs formed by posterior muscles						
SOL/BF	28.97	79.97	(-66.9,-19.1) .003 *	18.12	64.75	(-59.3,-21.1) <.001 *
SOL/ERE	21.49	48.45	(-56.2,-6.5) .015 *	12.20	47.77	(-63.7,-19.9) <.001 *
BF/ERE	34.54	60.80	(-74.6,-3.8) .032	23.80	53.03	(-94.1,-9.9) .011*
Pairs formed by antagonist muscles						
TA/SOL	18.68	74.95	(-87.8,-35.8) <.001 *	20.01	63.57	(-62.4,-12.7) .005 *
RF/BF	18.14	82.69	(-79.6,-26.2) .001 *	15.18	65.18	(-106.6,-22.6) <.001 *
RA/ERE	27.00	80.62	(-93.6,-34.9) <.001 *	23.38	72.95	(-81.7,-23.0) .001 *
Pairs formed by mixed muscles						
TA/BF	5.49	4.74	(-3.3,1.9) .796	3.75	5.13	(-4.7,0.5) .165
TA/ERE	5.40	6.93	(-5.2,1.4) .280	4.02	7.36	(-6.2,-0.05) .075
RF/SOL	4.86	7.81	(-6.3,0.2) .075	3.64	6.87	(-7.1,-0.5) .023
RF/ERE	5.88	5.14	(-4.8,2.1) .853	4.18	7.47	(-6.3,0.2) .075
RA/SOL	4.06	7.07	(-7.0,-0.9) .035	3.87	8.76	(-6.5,-2.6) <.001 *
RA/BF	3.29	8.41	(-8.3,-1.9) .009 *	2.91	6.78	(-6.4,-0.7) .035

Table 6.5. Median of the integrals of the z-scored intermuscular coherence computed over the frequency interval of 1–10 Hz ($INT_{Coh1-10Hz}$) for each muscle pair of young and older adults (*control* and *senior* groups, respectively) during unperturbed stance with and without vision (*BOE* and *BCE* conditions). Note: [+] represents *Vision* effect ($p < 0.02$). *Age* effect is not presented here. Note: soleus (SOL), biceps femoris (BF), lumbar erector spinae (ERE), tibialis anterior (TA), rectus femoris (RF), and rectus abdominis (RA).

| | $INT_{Coh1-10Hz}$ | | | | | |
| | Control group | | | Senior group | | |
	BOE (Vision)	BCE (No Vision)	95% confidence interval of the difference and *p* value	BOE (Vision)	BCE (No Vision)	95% confidence interval of the difference and *p* value
Pairs formed by anterior muscles						
TA/RF	36.09	23.82	(8.6,26.6) .005 [+]	61.69	68.92	(-26.6,29.3) .878
TA/RA	33.66	11.17	(8.6,27.2) .005 [+]	56.56	49.65	(-33.1,40.0) .333
RF/RA	4.48	11.91	(-14.1,5.8) .241	62.19	49.49	(-19.2,42.3) .169
Pairs formed by posterior muscles						
SOL/BF	28.97	18.12	(3.2,20.3) .007 [+]	79.97	64.75	(-14.8,44.0) .139
SOL/ERE	21.49	12.20	(2.2,14.3) .005 [+]	48.45	47.77	(-34.7,30.4) .646
BF/ERE	34.54	23.80	(1.7,21.8) .005 [+]	60.80	53.03	(-55.1,53.0) .878
Pairs formed by antagonist muscles						
TA/SOL	18.68	20.01	(-19.2,12.4) .646	74.95	63.57	(-11.5,53.2) .114
RF/BF	18.14	15.18	(-6.6,18.2) .285	82.69	65.18	(-54.1,42.2) .959
RA/ERE	27.00	23.38	(-7.3,10.9) .799	80.62	72.95	(-26.7,54.2) .093
Pairs formed by mixed muscles						
TA/BF	5.49	3.75	(-0.5,3.2) .114	4.74	5.13	(-3.22,3.13) .878
TA/ERE	5.40	4.02	(-1.6,3.4) .575	6.93	7.36	(-4.09,3.4) .646
RF/SOL	4.86	3.64	(-1.7,4.0) .445	7.81	6.87	(-3.3,4.1) .575
RF/ERE	5.88	4.18	(-1.3,3.2) .285	5.14	7.47	(-4.9,3.2) .285
RA/SOL	4.06	3.87	(-1.0,1.0) .878	7.07	8.76	(-4.1,2.9) .445
RA/BF	3.29	2.91	(-1.3,1.5) .878	8.41	6.78	(-2.4,5.7) .169

6.5. DISCUSSION

Age-related degenerative processes have a direct impact on the performance of functional daily tasks, such as the simple task of standing upright. As a consequence, balance becomes more challenging and susceptibility to falls increases as the individual grows older. Among several age-related physiological changes, the progressive and non-homogeneous degeneration of both white and gray matter, decline of sensory functions, and sarcopenia, seem to impact motor control in older adults. Several studies have reported not only changes in the magnitude and sequencing of muscle activation in older adults (Howard 1988, Doherty 1993a, Roos 1999, Amiridis et al 2003, Benjuya et al 2004, Nagai et al 2011), but also an increased cortical activation and a decreased modulation of presynaptic inhibition associated with a cocontraction pattern (Woollacott et al 1988, Manchester et al 1989, Melzer et al 2001, Laughton et al 2003, Benjuya et al 2004, Klass et al 2007, Tucker et al 2008, Baudry et al 2010, Nagai et al 2011,2013, Papegaaij et al 2014, Lee et al 2015, Craig et al 2016).

The effects of aforementioned age-related changes on postural sway behavior are discussed in *Chapter 7*. The discussion in this chapter focuses on the effects of physiological age-related changes on multi-muscle control during upright stance. In general, the findings suggest a reorganization of how the aging CNS controls multiple postural muscles during upright stance as a possible way to cope with physiological changes induced by aging.

The effects of aging on multiple postural muscle control.

The formation of synergistic muscle groups was previously reported for healthy young individuals (*Chapters 3* and *4* of the dissertation, Krishnamoorthy et al 2003a,b, Danna-dos-Santos et al 2008). The present study revealed interesting findings regarding changes in the control of multiple postural muscles as the individual grows older. The first finding was the presence of significant levels of muscle synchronization at lower frequency bands, indicating the presence of correlated neural inputs to aging skeletal muscles responsible for controlling upright stance. This finding resonates with the

principles of multi-muscle control based on Bernstein's school of thought and the contemporary developments of his rationale. More specifically, older individuals showed signs of synchronization at lower frequency bands for three distinct postural muscle groups: (a) group formed by solely anterior muscles (tibialis anterior, rectus femoris, and rectus abdominis); (b) group formed by solely posterior muscles (soleus, biceps femoris, and lumbar erector spinae); and (c) group formed by antagonist muscle pairs (tibialis anterior and soleus, rectus femoris and biceps femoris, rectus abdominis and lumbar erector spinae). The presence of signs of synchronization for these three muscle groups was also found in young individuals (see *Results* section in this chapter and in *Chapter 4*). Moreover, these three muscle groups were previously recognized as synergistics and referred to as the "push-forward M-mode", "push-back M-mode", and M-mode controlling body sway in the sagittal plane by Krishnamoorthy et al (2003a,b). On the other hand, no significant coherence at lower frequency bands was found for mixed muscle pairs formed by one anterior and one posterior non-antagonist muscles for either young or older individuals. Therefore, findings described in this chapter suggest that the aging CNS is able to control a large number of degrees of freedom by forming synergistic muscle groups.

Despite the presence of synchronization at lower frequency bands of postural muscles in young and older adults, a second finding in this study revealed the effects of age on the distribution of such synchronizations. During bipedal stance with eyes open, correlated neural inputs to posterior muscles occurred in a larger frequency band (1–10 Hz) in older adults, compared to that in young adults (1–5 Hz); and correlated neural inputs to antagonist muscles occurred in a smaller frequency band (1–10 Hz) in older adults, compared to that in young adults (1–25 Hz). The frequency band of significant intermuscular coherence for anterior muscles was kept similar between young and older adults (1–10 Hz), and no significant intermuscular coherence was found among mixed muscles for either young or older adults. These findings suggest that differences in the frequency distribution of correlated neural inputs to postural muscles may be age specific. In addition, the intermuscular coherence spectra for these three synergistic muscle groups (push-forward, push-back, and antagonist groups) looked similar in older adults. However, this does not mean that the aging CNS sent the same neural input to

these three groups since the coherence profile represents the correlation in the frequency domain rather than in the time domain.

When analyzing the strength of the correlated neural inputs within the frequency band with significant intermuscular coherence (1–10 Hz), all three synergistic muscle groups (push-forward, push-back, and antagonist groups) presented a stronger coherence in older adults, compared to that in young adults. Such increased synchronization may reflect changes in the control strategy of older adults, such as the increased activation of muscles acting on ankles and hips, the use of cocontraction pattern, and the use of both ankle and hip strategies to control upright stance reported in the literature (Amiridis et al 2003, Laughton et al 2003, Benjuya et al 2004, Nagai et al 2011,2013, Lee et al 2015, Craig et al 2016). In addition, age-related changes in the strength of correlated neural inputs to these three synergistic groups may also be related to the asymmetric pattern of increased muscle activation in older adults reported by Tsai et al (2014).

The effects of a sensory challenging task (temporary visual removal) on the formation of synergistic muscle groups in older adults.

When older adults stood upright with eyes closed, only the frequency distribution of correlated neural inputs to antagonist muscle pairs in older adults reduced from the interval 1–10 Hz to 1–5 Hz. The fact that there were no significant effects of temporary disruption of visual information in older adults on the strength of correlated neural inputs to each muscle group suggests minimal effect of visual input in the control strategy in these individuals. Although these results corroborate the hypothesis of the decreased contribution of visual input on balance control as the individual ages suggested by Lord and Ward (1994) and Turano et al (1994), it is important to note that our senior participants were healthy nonfaller individuals aged 65 to 74 years old.

Interestingly, the effects of temporary visual removal on the formation of synergistic muscle groups in healthy nonfaller older adults differed from that in healthy young adults. The effects of visual input on the formation of postural muscle synergies in young adults were previously discussed in *Chapter 4*. In general, young adults presented a decrease in the strength of correlated neural inputs to both anterior and posterior muscle

groups, and only the antagonist muscles presented significant synchronization when vision was temporarily absent. The fact that only antagonist muscles showed significant coherence under both conditions (open and closed eyes) is not surprising, considering a previous report of central input co-activating antagonist motoneurons (Hansen et al 2002). This greater effect of vision on the formation of correlated neural inputs to postural muscles in young adults compared to that in older adults suggests changes in sensory weighting as the individual grows older.

In addition, the researcher investigated the effects of aging on the formation of correlated neural inputs to postural muscles when visual input was not allowed. Under the task of upright stance with closed eyes, older adults showed signs of correlated neural inputs not only to antagonist muscles, but also to anterior muscles and posterior muscles, whereas young adults showed no significant coherence for either anterior muscles or posterior muscles. The temporary removal of visual input in older adults did not abolish neural drive to either push-forward or push-back M-modes. This finding corroborates the idea of differences in sensory reweighting between young and older adults (Horak et al 1989, Teasdale and Simoneau 2001, Wiesmeier et al 2015).

6.6. CONCLUSIONS

The use of intermuscular coherence analysis at lower frequency bands provided new knowledge on the mechanisms underlying strategies adopted by the aging CNS to generate and distribute correlated neural commands to multiple postural muscles, as happens in young adults. However, the frequency distribution of such neural drives in healthy nonfaller older adults differed from that in healthy young adults. In addition, intermuscular coherence at lower frequency bands was stronger in older adults. Therefore, the aging CNS seems to reorganize the formation of such correlated neural inputs to form synergistic muscle groups responsible for controlling upright stance and avoiding falls.

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CHAPTER 7

Age-related changes on postural sway:
a balance assessment using postural indices from multiple domains

7.1. INTRODUCTION

Falls in the elderly are the leading cause of fatal and nonfatal injuries, such as traumatic brain injury and hip fractures (Center for Disease Control and Prevention 2014). Current efforts to reduce falls and fall-related injuries include assessments of fall risk and interventions to improve balance. However, the lack of knowledge about the effects of aging on neurophysiological mechanisms of postural control has limited the advance in the field of fall prevention and rehabilitation of balance disorders.

Aging is associated with progressive and non-homogeneous degeneration of multiple physiological systems, affecting functional activities such as the simple task of upright standing. In an effort to assess balance in older adults, several studies have investigated body sway by extracting features from the body's center of pressure (COP) signals. Traditional measures extracted from COP signals include body sway area, peak-to-peak sway amplitude, signal variability, sway velocity, and signal frequency contents. In general, previous studies have suggested that the upright stance in older adults is characterized by a larger, faster, and more variable body oscillation, when compared to young adults (Prieto et al 1996, Amiridis et al 2003, Benjuya et al 2004, Demura et al 2008, Wiesmeier et al 2015). Although the frequency contents of the COP signal are often overlooked, a few studies reported age-related changes in the power spectrum density of this signal (Maki et al 1990, Vieira et al 2009, Wiesmeier et al 2015).

In the past decades, a few innovative methods for investigating the structure of postural sway were proposed. Richman and Moorman (2000) applied *Sample Entropy Analysis* to investigate and quantify the unpredictability level of the COP fluctuation in time. This analysis has been successfully used to address the irregularity and randomness of body sway (Ramdani et al 2009, Borg and Laxaback 2010, Mei et al 2013, Rigoldi et al 2013, Clark et al 2014, Perez et al 2014, Fino et al 2015, Degani et al 2016). In

general, smaller sample entropy estimates indicate a more predictable and regular pattern of COP displacement in time, whereas higher estimates indicate a more irregular and random postural sway. However, the use of entropy analysis to measure the randomness and irregularity of the COP behavior is still to be explored in both healthy individuals and those with balance disorders.

Another novel method for investigating the structure of postural sway is the *Rambling and Trembling Method*, introduced by Zatsiorsky and Duarte (2000). Based on the hypothesis that the body sways from one equilibrium point to another during upright stance (*Equilibrium Point Hypothesis* of motor control proposed by Feldman 1986, Feldman and Levin 1995), the stabilogram may be decomposed into two components, termed *rambling* and *trembling*. The rambling component of the COP signal represents the migration of the reference point from one instant equilibrium point to the subsequent one, whereas the trembling component represents the oscillation around this moving reference point. A few studies investigated the *Rambling-Trembling Hypothesis* (Mochizuki et al 2006, Danna-dos-Santos et al 2008, Shin et al 2011, Tahayori et al 2012, Sarabon et al 2013) and its reliability was reported by Slomka et al (2013).

In an effort to advance knowledge and increase awareness of the importance of including variables from different domains when assessing balance in the elderly, the present study was designed to investigate body sway in older adults from different perspectives. The study included postural indices from spatio-temporal, frequency, and structural domains. In general, it was hypothesized that the inclusion of multiple domains would allow observations of subtle changes in postural control that are likely missed when a single domain approach is employed. In addition, the use of multiple domains is likely to convey a more comprehensive panel of results leading to a better understanding of the neural mechanisms to be explored in later chapters of this dissertation. It was expected that older individuals would present a more random pattern of postural sway, when compared to young individuals. Changes in both rambling and trembling components of COP sway based on the rationale that physiological age-related changes affect both central and peripheral mechanisms of postural control were also expected.

7.2. METHODS

Participants. A total of twenty five (25) healthy participants volunteered for this study, forming two distinct groups according to their age: eleven (11) healthy young adults forming the *Control Group* (mean age = 27.1 years old, *SD* = 3.8), and fourteen (14) healthy older adults forming the *Senior Group* (mean age = 68.8 years old, *SD* = 3.2). The exclusion criteria for both groups included (a) previous history of sensory, musculoskeletal or neurological disorder, (b) history of previous surgeries, (c) history of cardiopulmonary disease, and (d) history of substance abuse. Additionally, the senior group included only nonfallers with age ranging 65 to 74 years (older adults). Therefore, additional exclusion criteria for the senior group included (a) history of falls (fallers) and (b) age between 75–84 years (older old adults) or over 85 years (oldest old adults). Prior to their participation, all volunteers gave their informed consent based on the procedures approved by the local Institutional Review Board and conforming to The Declaration of Helsinki. *Table 7.1* describes the general demographics of the participants.

Table 7.1. General demographics (mean and standard deviation) of the participants forming both control and senior groups.

Group	Age (years)	Weight (kg)	Height (cm)	Gender
Control	27.1 ± 3.8	69.8 ± 9.9	1.73 ± 0.10	7 females and 4 males
Senior	68.8 ± 3.2	73.0 ± 12. 9	1.68 ± 0.09	8 females and 6 males

Apparatus. A force platform (AMTI BP400600, AMTI Inc.) was used to acquire the horizontal and vertical components of the ground reaction force (*Fx*, *Fy*, and *Fz*) as well as the moments of force around the frontal, sagittal, and vertical axes (*Mx*, *My*, and *Mz*). These signals were used to compute the body's center of pressure coordinates in anterior-posterior and medio-lateral directions (*COPap* and *COPml*, respectively) according to manufacturer's directions:

$$COPap = (\text{ -} h * Fx - My) / Fz \qquad (7.1)$$

$$COPml = (\text{ -} h * Fy - Mx) / Fz, \qquad (7.2)$$

where *h* is the height of the base of support above the force platform.

Data from participants of the control group were collected at a sampling frequency of 50 Hz, while data from participants of the senior group were collected at a sampling frequency of 2000 Hz since muscle activity was also recorded for future studies.

Experimental procedures. All participants were asked to perform three upright standing tasks: bipedal stance with opened eyes for 120 seconds (*BOE* or *Vision* condition), bipedal stance with closed eyes for 120 seconds (*BCE* or *No Vision* condition), and body oscillation to the limits of stability (*BOUNDARIES*). For all tasks, participants were instructed to stand barefoot in the force platform with their arms crossed and their feet 15 cm apart and parallel. For the *BOE* task, participants were asked to focus their vision on a physical static point placed at eye level and at a distance of approximately 2 meters. For the *BCE* task, participants were asked to maintain the same posture adopted for the *BOE* task, but with their eyes closed. The *BCE* task was purposefully performed to impose a sensory challenge on upright stance. The rationale was that the natural unperturbed stance may be too simple to detect subtle age-related balance deficits, though the individual may already have problems with chronic disequilibrium and falls. For the *BOUNDARIES* task, participants were asked to stand on both feet and move their whole body around their ankle joints as far as possible forward, backward, to the right, and to the left. This task was performed to measure the limits of body sway without losing balance. The duration of the entire experimental session lasted approximately twenty minutes, and none of the participants reported fatigue.

7.3. DATA PROCESSING

Center of pressure coordinates in both anterior-posterior and medio-lateral directions (*COPap* and *COPml*, respectively) were analyzed off-line with a series of custom-written software routines (*BalanceLab vs 1.1*) in Matlab R2012b (Mathworks Inc, Natick, MA).

Before extracting the response variables of interest from the COP coordinates, the *COPap* and *COPml* signals were down sampled to 10 Hz, normalized by bringing the average position of the COP to the zero line and, next, detrended by the mean of each time series. These two last processes were performed to allow positioning the COP coordinates at the center of an *xy* coordinate system and draw any comparisons of basic COP features across participants. Postural indices of interest selected to assess postural control were extracted from COP signals and computed using the *BalanceLab* software.

Two spatial variables expressing the maximum magnitude of postural sway were extracted from the *BOUNDARIES* task: the limits of the stability in the anterior-posterior and medio-lateral directions (LOS_{ap} and LOS_{ml}, respectively). These postural indices were computed as the difference between the maximum and minimum COP displacement in each direction. The *LOS* expresses the functional base of support.

For the other two tasks (*BOE* and *BCE*), balance was measured using postural indices in the spatio-temporal, frequency, and structural domains. The **spatio-temporal variables** included the following:

- *StabArea* (**cm^2**): instead of using traditional methods to measure the elliptical area containing 90% to 95% of the COP displacement, the stabilometric area of the entire COP displacement was computed using the approach of the sector formula of Leibniz. The COP plot was divided into equal angles from the center ranging from 0° to 360°. Within each of these 360 sectors, the maximal distance between the center and the COP coordinates was calculated. Subsequently, each maximal position was considered as vertices of a 360-side polygon. Finally, the area of the resulting polygon was computed reflecting the area containing the whole COP trajectory (see Schubert et al 2012 for more details). *Figure 7.1A* illustrates the COP signal and the polygon representing the stabilometric area of the COP sway.

- *TotalSway* (**cm):** the total length of the COP trajectory.

- $Range_{ap}$ and $Range_{ml}$ (**cm):** the amplitude of the COP displacement in each direction, computed by the difference between the maximum and minimum values (*Figure 7.1A*).

- ***RMS$_{ap}$*** and ***RMS$_{ml}$*** **(cm):** the root mean square (*RMS*) of the COP displacement, computed as the square root of the mean of the squares of the COP displacement for each direction. Considering that the RMS represents the variability of the COP around zero and the COP signals were previously normalized and detrended, the RMS computed in this study represented the variability of the COP around its mean value.

- ***TMV*** **(cm/s):** the total mean velocity of the COP displacement, computed as the total length of the COP trajectory (*TotalSway*, in cm) divided by the duration of the trial (120 seconds).

- ***MV$_{ap}$*** and ***MV$_{ml}$*** **(cm/s):** the mean velocity of the COP displacement, computed separately for each direction.

- ***TMJerk*** **(cm/s^3):** the total sway jerkiness of the COP displacement is the rate of change of the COP acceleration. Total mean jerkiness was computed as the third derivative of the COP position with respect to time. Although the use of sway jerkiness is not usual, Mancini et al (2011,2012) showed the sensitivity of this measure in detecting changes in postural sway.

- ***MJerk$_{ap}$*** and ***MJerk$_{ml}$*** **(cm/s^3):** the mean sway jerkiness of the COP displacement, computed separately for each direction.

The selected response variables in the **frequency domain** included the following:

- ***Fmean$_{ap}$*** and ***Fmean$_{ml}$*** **(Hz):** the mean power frequency , computed as the mean frequency on the power spectrum density (PSD) of the COP signal in each direction. The power spectrum density was estimated using Fast Fourier Transformation (see Duarte and Freitas 2010 for more details). *Figure 7.1B* illustrates the PSD of the *COPap* signal in a representative trial, along with variables in the frequency domain, including *Fmean$_{ap}$*.

- ***F80$_{COPap}$*** and ***F80$_{COPml}$*** **(Hz):** the frequency band of the stabilogram containing 80% of the PSD for each direction. This variable was selected based on studies showing that it is one of the best spectral measurements to characterize postural sway (Baratto et al 2002). *Figure 7.1B* illustrates *F80$_{ap}$* in a representative trial.

(A)
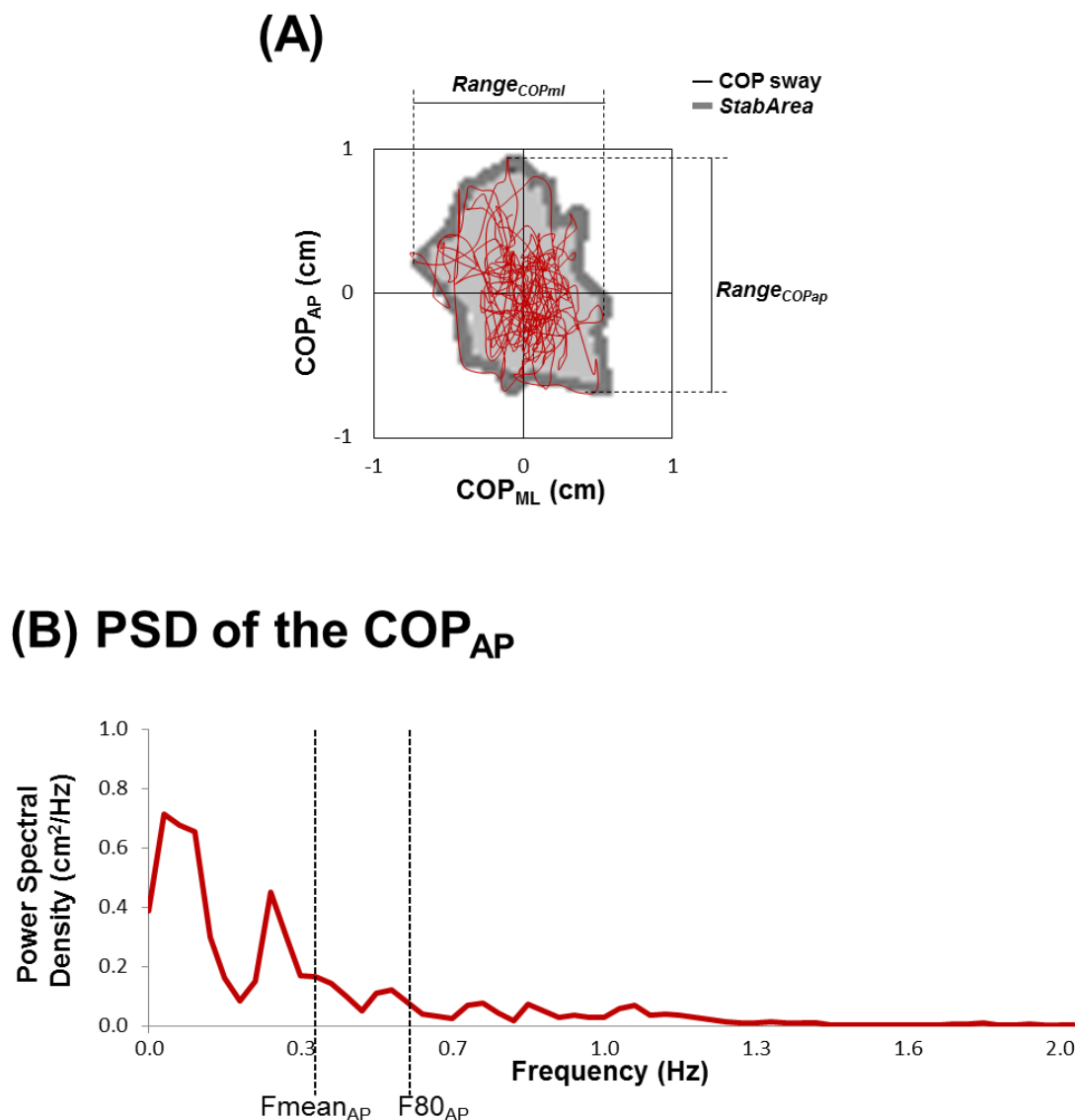
(B) PSD of the COP$_{AP}$

Figure 7.1. (A) The center of pressure (*COP*) displacement, the polygon containing the stabilometric area (*StabArea*) of the COP sway, and the amplitude of the COP displacement in each direction (*Range$_{ap}$* and *Range$_{ml}$*). **(B)** The power spectrum density (PSD) of the anterior-posterior COP oscillation with respective mean power frequency (*Fmean$_{ap}$*), and frequency at which 80% of the COPap spectral power is lower than (*F80$_{ap}$*) in a representative trial.

Finally, the postural sway was assessed in the **structural domain** using the following variables:

- **SEnt$_{ap}$** and **SEnt$_{ml}$:** the sample entropy (*SEnt*) of the COP trajectory in each direction, computed by an algorithm that measures correlation, persistence, and regularity of the COP signal in time. The output of the sample entropy analysis is a single, nonnegative real number indicating the level of irregularity of the time series. Smaller sample entropy estimates indicate many repetitive patterns of COP fluctuation in time, e.g., a more predictable and regular postural sway, whereas larger sample entropy estimates indicate a more irregular and random pattern of COP displacement (see Pincus 1991 and Richman and Moorman 2000 for more details regarding the sample entropy computational method).

- **CrossSEnt:** the cross-sample entropy, computed to measure the degree of asynchrony or dissimilarity between *COPap* and *COPml* signals in time (see Richman and Moorman 2000 for more details). The difference between *SEnt* and *CrossSEnt* computational methods is that the *SEnt* compares a series with itself, whereas the *CrossSEnt* compares two related time series (*COPap* and *COPml*). Higher *CrossSEnt* values indicate more asynchrony of the postural sway between the two directions; lower values, in contrast, indicate more co-dependence of the postural sway dynamics between the two directions.

- ***Variables extracted from the rambling and trembling components of the COP***: the horizontal forces (*Fx* and *Fy*) and the COP signals were filtered (0.9 Hz low-pass, zero-lag Butterworth second-order filter) and, next, *COPap* and *COPml* were decomposed into rambling (*RM*) and trembling (*TR*) components. The rambling trajectory was computed by interpolating the discrete instant equilibrium point trajectory with cubic spline function, while the trembling component was computed as the difference between the approximated rambling trajectory and the original COP trajectory (see Duarte and Zatsiorsky 1999, and Zatsiorsky and Duarte 2000 for more details). *Figure 7.2* illustrates the decomposition of the *COPap* signal into rambling and trembling from a representative senior participant performing a BOE trial. Rambling and trembling signals in each direction were analyzed using spatio-temporal, frequency, and structural measures as follows:

- **StabArea$_{RM}$** and **StabArea$_{TR}$ (cm^2)**: the stabilometric area of the rambling and trembling trajectories, respectively.

- **Range$_{RMap}$, Range$_{TRap}$, Range$_{RMml}$,** and **Range$_{TRml}$ (cm)**: the amplitude of the rambling and trembling displacements in each direction.

- **RMS$_{RMap}$, RMS$_{TRap}$, RMS$_{RMml}$,** and **RMS$_{TRml}$ (cm):** the root mean square (RMS) of the rambling and trembling displacements in each direction.

- **MV$_{RMap}$, MV$_{TRap}$, MV$_{RMml}$,** and **MV$_{TRml}$ (cm/s):** the mean rambling and trembling velocities in each direction.

- **MJerk$_{RMap}$, MJerk$_{TRap}$, MJerk$_{RMml}$,** and **MJerk$_{TRml}$ (cm/s^3):** the mean sway jerkiness of the rambling and trembling displacements in each direction.

- **Fmean$_{RMap}$, Fmean$_{TRap}$, Fmean$_{RMml}$,** and **Fmean$_{TRml}$ (Hz):** the mean power frequency of the rambling and trembling displacements in each direction.

- **F80$_{RMap}$, F80$_{TRap}$, F80$_{RMml}$,** and **F80$_{TRml}$ (Hz):** the frequency at which 80% of the PSD of the rambling and trembling displacements in each direction is reached.

- **SEnt$_{RMap}$, SEnt$_{TRap}$, SEnt$_{RMml}$,** and **SEnt$_{TRml}$:** the sample entropy of the rambling and trembling displacements in each direction.
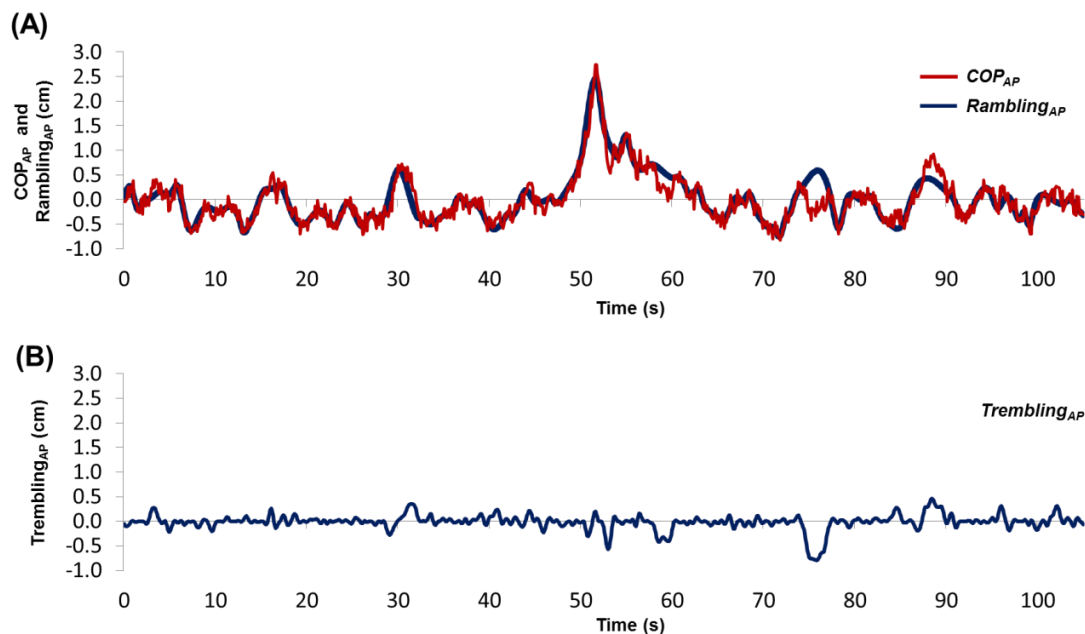
Figure 7.2. (A) The center of pressure displacement and its rambling component in the anterior-posterior direction (*COP$_{AP}$* and *Rambling$_{AP}$*, respectively). **(B)** The trembling component (*Trembling$_{AP}$*) of the *COP$_{AP}$* displacement of a representative participant from the senior group performing unperturbed bipedal stance with eyes open.

Statistical Approach. For all response variables, medians across participants were reported. Statistical tests were performed using the IBM SPSS statistics software suite (version 22, IBM® SPSS®).

Considering the number of variables measured in this study, Shapiro-Wilk tests were used to verify whether data from each domain were normally distributed. Since some of the variables did not follow a normal distribution, non-parametric Mann Whitney U tests were used to investigate the effects of *Age* (*Control* and *Senior* groups), and Wilcoxon signed-rank tests were used to investigate the effects of *Vision* (*BOE* and *BCE* conditions) on the variables of interest.

Since multiple comparisons were performed, an adjustment of the significance level is recommended. However, such correction could be somewhat too conservative considering the large number of tests performed and the positive correlation among these tests. In addition, this adjustment could increase the probability of producing false negatives and, consequently, reduce the statistical power. Therefore, the significant level was fixed at 0.02.

7.4. RESULTS

All participants were able to successfully accomplish all tasks without losing their balance. *Figure 7.3* shows the COP displacement of one typical control participant and one typical senior participant performing the *BOUNDARIES* task. Regarding the limits of stability, the functional base of support for young and older adults was not significantly different for either the anterior-posterior or medio-lateral directions (*Table 7.2*). This finding was confirmed by Mann Whitney U tests with factor *Age* on *LOSap* and *LOSml* ($p=0.702$ and $p=0.125$, respectively).
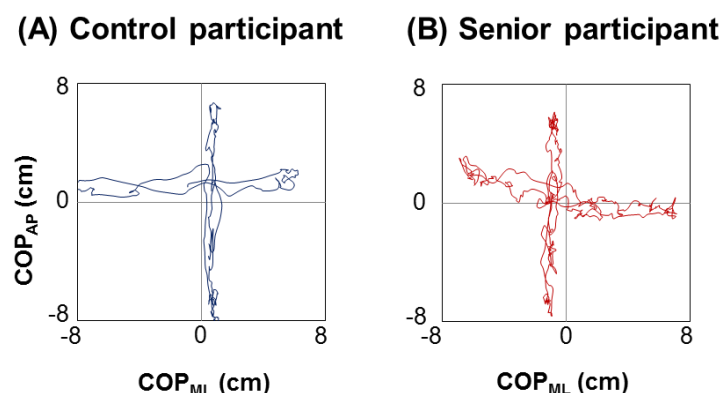
Figure 7.3. The displacement of the center of pressure (COP) of one typical control participant (panel A) and one typical senior participant (panel B) performing the *BOUNDARIES* task.

Table 7.2. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of the limits of stability in the anterior-posterior (AP) and medio-lateral (ML) direction (LOS_{AP} and LOS_{ML}, respectively). Note: ***** indicates significant *Age* effect ($p < 0.02$).

	Control	*Senior*	95% confidence interval of the difference	*p* value
LOS_{AP} (cm)	14.9	13.7	(-1.9, 2.8)	$p = 0.702$
LOS_{ML} (cm)	17.8	15.3	(-0.8, 3.9)	$p = 0.125$

Figure 7.4 shows the COP displacement of one typical control participant and one typical senior participant performing both BOE and BCE tasks. Note visual differences in the magnitude of the postural sway between these two participants and between bipedal stance with eyes open and closed. *Figures 7.5 and 7.6* present boxplots of COP spatio-temporal, frequency, and structural variables of young and older adults performing both tasks.
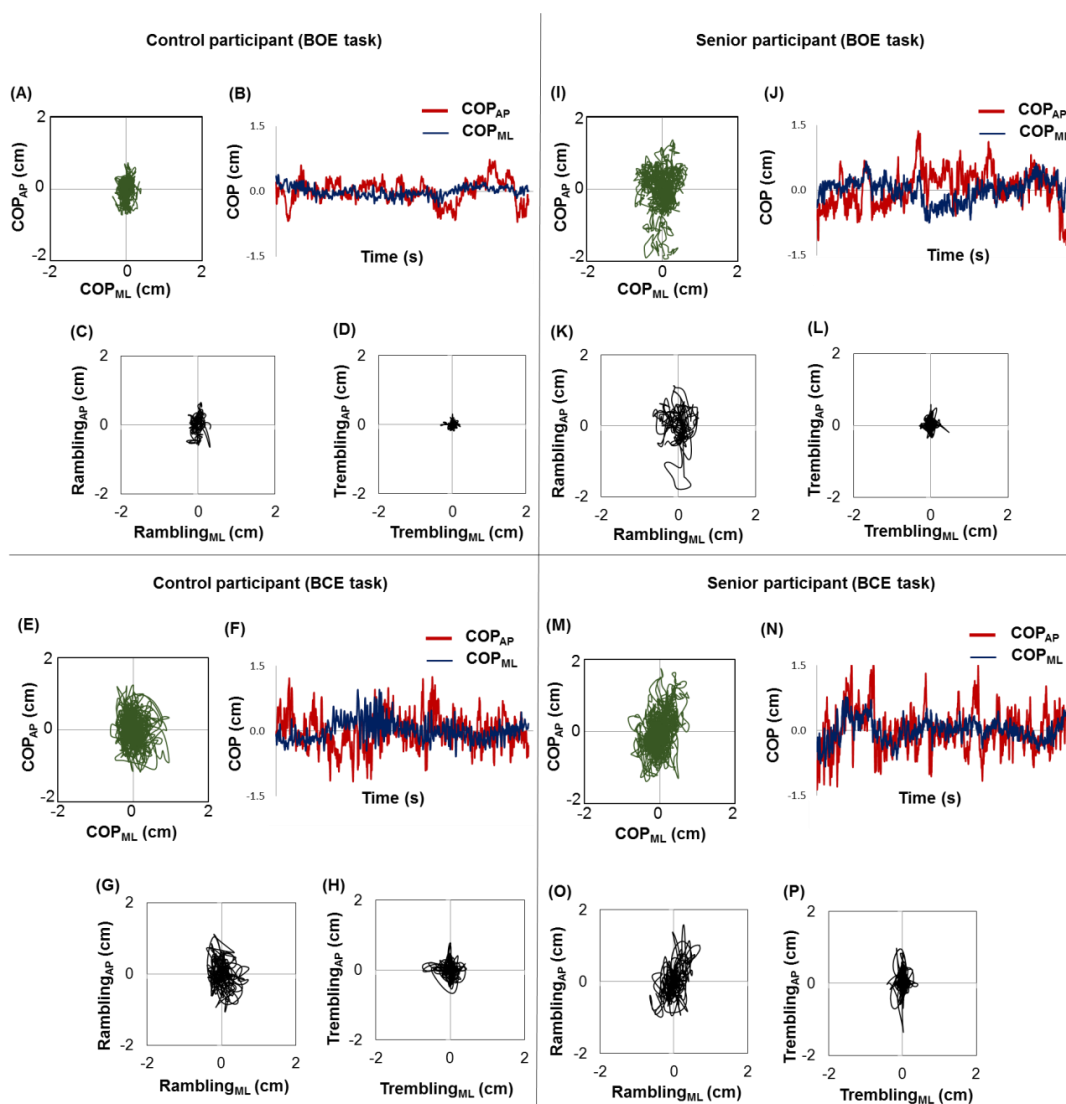
Figure 7.4. The displacement of the center of pressure (COP) and its components (rambling and trembling) of one typical control participant (panels **A** to **H**) and one typical senior participant (panels **I** to **P**) performing bipedal standing tasks with open eyes (*BOE* task) (panels **A** to **D** and **I** to **L**) and closed eyes (*BCE* task) (panels **E** to **H** and **M** to **P**).

SPATIO-TEMPORAL DOMAIN
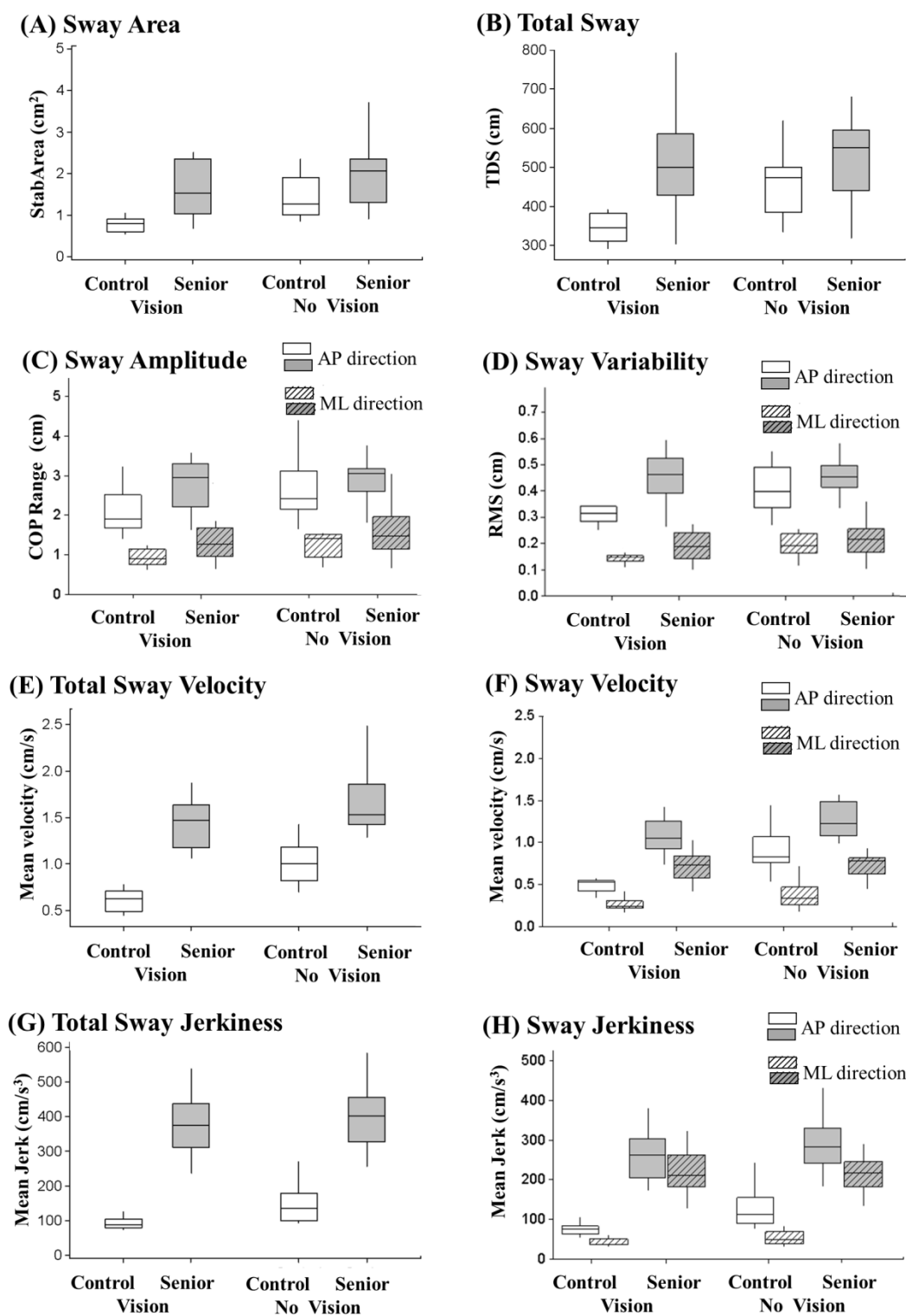
Figure 7.5. Boxplot with spatio-temporal variables of young adults (control group) and older adults (senior group) performing unperturbed stance with eyes open (Vision) and closed (No Vision).

FREQUENCY DOMAIN
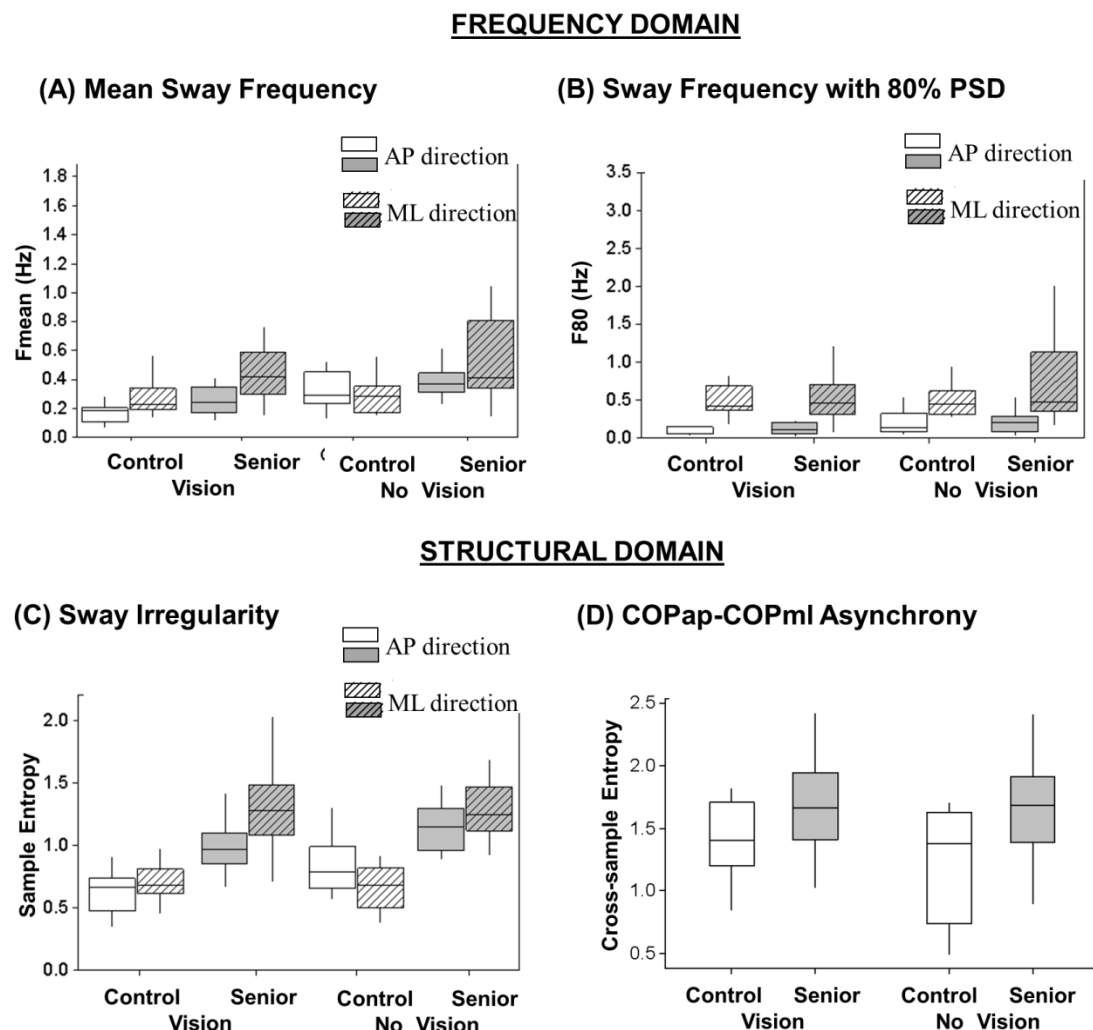
Figure 7.6. Boxplot with frequency (panels A and B) and structural (panels C and D) variables of young adults (control group) and older adults (senior group) performing unperturbed stance with eyes open (Vision) and closed (No Vision).

The effects of aging on balance control.

Tables 7.3, *7.4* and *7.5* present the median across participants (*Control* and *Senior*) of postural indices extracted from the COP signal and from both rambling and trembling components of the COP during upright stance with eyes open and closed (*BOE* and *BCE*, respectively), along with *p*-values from Mann Whitney U tests on factor *Age* for these response variables.

Table 7.3. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of postural indices extracted from the center of pressure signal during upright stance with eyes open and closed (*BOE* and *BCE*, respectively). Note: ***** indicates significant *Age* effect ($p < 0.02$).

	Vision (*BOE*)			No Vision (*BCE*)		
	Control	*Senior*	95% CI of the difference and p value	*Control*	*Senior*	95% CI of the difference and p value
StabArea (cm^2)	0.80	1.53	(-1.58,-0.31) .001 *	1.25	2.07	(-1.10,0.35) .101
TotalSway (cm)	344	499	(-235,-36) .002 *	472	550	(-143,11) .112
Range$_{ap}$ (cm)	1.90	2.96	(-1.26,-0.6) .009 *	2.41	3.05	(-0.87,0.24) .171
Range$_{ml}$ (cm)	0.89	1.25	(-0.78,-0.07) .014 *	1.40	1.47	(-0.78,0.74) .352
RMS$_{ap}$ (cm)	0.32	0.46	(-0.20,-0.03) .003 *	0.40	0.45	(-0.12,0.03) .273
RMS$_{ml}$ (cm)	0.15	0.19	(-0.13,0.00) .027	0.19	0.22	(-0.09,0.05) .547
TMV (cm/s)	0.63	1.47	(-0.99,-0.63) <.001 *	1.00	1.53	(-0.95,-0.34) <.001 *
MV$_{ap}$ (cm/s)	0.53	1.05	(-0.71,-0.44) <.001 *	0.83	1.22	(-0.74,-0.17) .002 *
MV$_{ml}$ (cm/s)	0.24	0.73	(-0.55,-0.33) <.001 *	0.33	0.78	(-0.47,-0.24) <.001 *
TMJerk (cm/s^3)	89	375	(-329,-223) <.001 *	133	404	(-323,-193) <.001 *
MJerk$_{ap}$ (cm/s^3)	74	261	(-219,-143) <.001 *	111	282	(-231,-115) <.001 *
MJerk$_{ml}$ (cm/s^3)	36	199	(-198,-134) <.001 *	46	205	(-179,-124) <.001 *
Fmean$_{ap}$ (Hz)	0.18	0.24	(-0.16,-0.01) .063	0.29	0.37	(-0.16,0.03) .208
Fmean$_{ml}$ (Hz)	0.23	0.42	(-0.51,-0.00) .007 *	0.28	0.41	(-0.40,-0.02) .029
F80$_{ap}$ (Hz)	0.05	0.11	(-0.26,0.12) .316	0.13	0.21	(-0.14,0.13) .956
F80$_{ml}$ (Hz)	0.42	0.46	(-0.74,0.28) .827	0.44	0.48	(-0.64,0.13) .351
SEnt$_{ap}$	0.66	0.96	(-0.51,-0.20) <.001 *	0.78	1.15	(-0.49,-0.14) .003 *
SEnt$_{ml}$	0.68	1.28	(-0.81,-0.35) <.001 *	0.68	1.25	(-0.80,-0.39) <.001 *
CrossSEnt	1.40	1.66	(-0.62,0.02) .080	1.37	1.68	(-0.85,-0.14) .016 *

Table 7.4. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of postural indices extracted from the *rambling* component of the center of pressure (*COP*) signal during upright stance with eyes open and closed (*BOE* and *BCE*, respectively). Note: ***** indicates significant *Age* effect ($p < 0.02$).

	Vision (*BOE*)			No Vision (*BCE*)		
	Control	*Senior*	95% CI of the difference and *p* value	*Control*	*Senior*	95% CI of the difference and *p* value
$StabArea_{RM}$ (cm^2)	0.62	1.06	(-1.22,-0.03) .014*	0.94	1.52	(-0.95,0.98) .381
Rambling component of the *COPap*						
$Range_{RMap}$ (cm)	1.74	2.22	(-1.04,-0.07) .035	2.18	2.55	(-0.78,0.36) .352
RMS_{RMap} (cm)	0.31	0.44	(-0.18,-0.01) .006 *	0.37	0.44	(-0.12,0.04) .476
MV_{RMap} (cm/s)	0.21	0.30	(-0.16,-0.06) <.001 *	0.35	0.42	(-0.16,0.00) .067
$MJerk_{RMap}$ (cm/s^3)	2.20	3.33	(-2.00,-0.52) .003 *	3.22	4.64	(-2.42,0.06) .063
$Fmean_{RMap}$ (Hz)	0.09	0.10	(-0.05,0.01) .427	0.15	0.17	(-0.04,0.03) .805
$F80_{RMap}$ (Hz)	0.12	0.17	(-0.08,0.03) .763	0.29	0.30	(-0.06,0.05) 1.000
$SEnt_{RMap}$	0.29	0.32	(-0.11,0.01) .112	0.43	0.43	(-0.08,0.02) .412
Rambling component of the *COPml*						
$Range_{RMml}$ (cm)	0.76	1.09	(-0.59,0.04) .125	1.13	1.16	(-0.57,0.87) .642
RMS_{RMml} (cm)	0.14	0.18	(-0.12,0.00) .049	0.19	0.21	(-0.09,0.06) .642
MV_{RMml} (cm/s)	0.12	0.16	(-0.10,0.00) .112	0.19	0.19	(-0.06,0.05) .870
$MJerk_{RMml}$ (cm/s^3)	1.73	1.63	(-0.76,0.42) .743	2.10	2.00	(-0.47,1.17) .702
$Fmean_{RMml}$ (Hz)	0.13	0.13	(-0.03,0.04) .763	0.17	0.17	(-0.02,0.06) .603
$F80_{RMml}$ (Hz)	0.27	0.23	(-0.04,0.12) .365	0.37	0.33	(-0.03,0.12) .311
$SEnt_{RMml}$	0.39	0.40	(-0.05,0.05) .784	0.42	0.45	(-0.10,0.05) .722

Table 7.5. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of postural indices extracted from the *trembling* component of the center of pressure (*COP*) signal during upright stance with eyes open and closed (*BOE* and *BCE*, respectively). Note: ***** indicates significant *Age* effect ($p < 0.02$).

	Vision (*BOE*)			No Vision (*BCE*)		
	Control	*Senior*	95% CI of the difference and *p* value	*Control*	*Senior*	95% CI of the difference and *p* value
$StabArea_{TR}$ (cm²)	0.07	0.21	(-0.27,-0.05) <.001 *	0.18	0.43	(-0.33,-0.04) .012 *
Trembling component of the *COPap*						
$Range_{TRap}$ (cm)	0.71	1.48	(-1.12,-0.35) .001 *	1.47	2.01	(-1.06,0.38) .228
RMS_{TRap} (cm)	0.08	0.16	(-0.13,-0.05) <.001 *	0.16	0.23	(-0.11,0.01) .171
MV_{TRap} (cm/s)	0.23	0.35	(-0.25,-0.07) <.001 *	0.44	0.49	(-0.23,0.07) .412
$MJerk_{TRap}$ (cm/s³)	9.58	14.87	(-9.19,-2.44) .001 *	15.38	18.59	(-9.05,2.29) .352
$Fmean_{TRap}$ (Hz)	0.53	0.41	(0.03,0.18) .007 *	0.49	0.38	(-0.01,0.16) .040
$F80_{TRap}$ (Hz)	0.84	0.67	(0.04,0.29) .012 *	0.78	0.61	(-0.00,0.23) .080
$Sent_{TRap}$	0.54	0.48	(-0.01,0.15) .037	0.55	0.45	(0.01,0.15) .016 *
Trembling component of the *COPml*						
$Range_{TRml}$ (cm)	0.53	0.63	(-0.32,0.11) .208	0.67	0.81	(-0.37,0.18) .443
RMS_{TRml} (cm)	0.05	0.07	(-0.04,0.00) .031	0.07	0.08	(-0.04,0.00) .208
MV_{TRml} (cm/s)	0.11	0.16	(-0.08,-0.00) .023	0.15	0.19	(-0.08,0.03) .273
$MJerk_{TRml}$ (cm/s³)	3.78	7.06	(-3.83,-0.94) .004 *	5.15	7.30	(-3.47,0.89) .063
$Fmean_{TRml}$ (Hz)	0.46	0.41	(-0.00,0.10) .208	0.49	0.42	(0.02,0.15) .010 *
$F80_{TRml}$ (Hz)	0.74	0.68	(-0.02,0.16) .198	0.76	0.63	(0.04,0.23) .009 *
$Sent_{TRml}$	0.42	0.46	(-0.07,0.00) .080	0.41	0.47	(-0.08,0.12) .055

In general, older adults presented a larger, faster, and more irregular body oscillation during bipedal stance with eyes open (**BOE task**), compared to young adults

(control group). Mann Whitney U tests confirmed the effects of *Age* (control and senior) on most of the spatio-temporal variables (*StabArea, TotalSway*, $Range_{ap}$ and, $Range_{ml}$, RMS_{ap}, *TMV*, MV_{ap} and MV_{ml}, *TMJerk*, $MJerk_{ap}$ and $MJerk_{ml}$). No significant effect on variable RMS_{ml} was observed. In the frequency domain, statistical tests revealed that older adults presented only a significantly higher mean frequency of COP sway in the medio-lateral direction, compared to young adults. In the structural domain, the significant increase in the irregularity of the pattern of body sway was confirmed by statistical tests on *Age* for the variables $SEnt_{ap}$ and $SEnt_{ml}$. No significant difference existed for the asynchrony level between *COPap* and *COPml* in older adults, compared to young adults. See all *p*-values for the effects of *Age* on postural indices extracted from the COP in *Table 7.3*.

Regarding the age-related changes in the rambling and trembling components of the *COPap* and *COPml* trajectories during unperturbed stance with eyes open (**BOE task**), Mann Whitney U tests revealed higher values for both rambling and trembling areas in older adults, compared to that in young adults. In the anterior-posterior direction, older adults presented faster and less smooth rambling and trembling trajectories. Statistical tests confirmed higher rambling variability, mean velocity, and jerkiness, as well as higher trembling amplitude, variability, mean velocity, and jerkiness in older adults. In addition, the trembling component of the COPap in older adults presented lower mean frequency and lower frequency at which 80% of the spectral power is reached, confirmed by statistical tests. No significant effects of *Age* were found for the other variables extracted from the rambling component of the COP in the anterior-posterior direction. See all *p*-values for the effects of *Age* on postural indices extracted from the rambling and trembling components of the COP in *Tables 7.4* and *7.5*.

In the medio-lateral direction, there was only a significant increase in the jerkiness of the trembling component of the COP signal ($MJerk_{TRml}$) in older adults, compared to young adults. Statistical tests revealed no effects of *Age* on the remaining variables ($Range_{RMml}$, RMS_{RMml}, MV_{RMml}, $MJerk_{RMml}$, $Fmean_{RMml}$, $F80_{RMml}$, $SEnt_{RMml}$, $Range_{TRml}$, RMS_{TRml}, MV_{TRml}, $Fmean_{TRml}$, $F80_{TRml}$, and $SEnt_{TRml}$). See all *p*-values for the effects of *Age* on postural indices extracted from the rambling and trembling components of the COP in *Tables 7.4* and *7.5*.

Additional statistical tests were performed to investigate the effects of aging on postural behavior when visual input was not allowed by asking the participants to close their eyes while standing upright (**BCE task**). Regarding spatio-temporal variables, a faster and less smooth COP displacement in older adults was confirmed by Mann Whitney U tests on *Age* for *TMV, MV$_{ap}$, MV$_{ml}$, TMJerk*, *MJerk$_{ap}$*, and *MJerk$_{ml}$*. No significant differences between older and young adults were found for the area, amplitude, and variability of the COP displacement. In the frequency domain, statistical tests revealed no effects of *Age* on *Fmean$_{ap}$*, *Fmean$_{ml}$*, *F80$_{ap}$*, and *F80$_{ml}$* when participants closed their eyes. In the structural domain, older adults presented a more irregular pattern of COP displacement in time for both directions and an increased asynchrony between COPap and COPml, compared to that for young adults. This result was confirmed by significant effects of *Age* on *SEnt$_{ap}$, SEnt$_{ml}$*, and *CrossSEnt*. See all *p*-values for the effects of *Age* on postural indices extracted from the COP in *Table 7.3*.

In addition, the effects of age on the rambling and trembling components of the postural sway during bipedal stance with eyes closed (**BCE task**) differed from that with eyes open. Mann Whitney U tests revealed only a significantly larger trembling (*StabArea$_{TR}$*), a significant increase in the irregularity of the anterior-posterior trembling (*SEnt$_{TRap}$*), and a significant decrease in the frequency of the medio-lateral trembling (*Fmean$_{TRml}$* and *F80$_{TRml}$*) in older adults with eyes closed, compared to that for young adults. No significant difference was found for the other variables extracted from the trembling component and for any variable extracted from the rambling component of the COP. See all *p*-values for the effects of *Age* on postural indices extracted from the rambling and trembling components of the COP in *Tables 7.4* and *7.5*.

The effects of visual input on balance control.

Tables 7.6, 7.7 and *7.8* present the median across participants (*Control* and *Senior*) of postural indices extracted from the COP signal and from both rambling and trembling components of the COP during upright stance with eyes open and closed (*BOE* and *BCE*, respectively), along with *p*-values from Wilcoxon signed-rank tests on factor *Vision* (BOE and BCE conditions) for these response variables.

Table 7.6. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of postural indices extracted from the center of pressure signal during upright stance with eyes open and closed (*Vision* and *No Vision* conditions, respectively). Note: [+] represents significant *Vision* effect ($p < 0.02$).

	Control Group			Senior Group		
	Vision	*No Vision*	95% CI of the difference and *p* value	*Vision*	*No Vision*	95% CI of the difference and *p* value
StabArea (cm^2)	0.80	1.25	(-1.44,-0.12) .003 [+]	1.53	2.07	(-0.89,0.46) .233
TotalSway (cm)	344	472	(-171,12) .041	499	550	(-94,76) .510
Range_{ap} (cm)	1.90	2.41	(-1.18,0.03) .062	2.96	3.05	(-0.60,0.32) .551
Range_{ml} (cm)	0.89	1.40	(-1.37,0.12) .016 [+]	1.25	1.47	(-0.69,0.24) .198
RMS_{ap} (cm)	0.32	0.40	(-0.16,0.03) .062	0.46	0.45	(-0.06,0.07) .730
RMS_{ml} (cm)	0.15	0.19	(-0.10,-0.02) .016 [+]	0.19	0.22	(-0.09,0.06) .414
TMV (cm/s)	0.63	1.00	(-0.64,-0.22) .003 [+]	1.47	1.53	(-0.53,-0.00) .006 [+]
MV_{ap} (cm/s)	0.53	0.83	(-0.58,-0.23) .003 [+]	1.05	1.22	(-0.53,-0.04) .002 [+]
MV_{ml} (cm/s)	0.24	0.33	(-0.21,0.00) .004 [+]	0.73	0.78	(-0.13,0.10) .551
TMJerk (cm/s^3)	89	133	(-86.-16) .003 [+]	375	404	(-102,35) .074
MJerk_{ap} (cm/s^3)	74	111	(-80.-17) .003 [+]	261	282	(-96,15) .041
MJerk_{ml} (cm/s^3)	36	46	(-24,-0.05) .003 [+]	199	205	(-33,-39) .826
Fmean_{ap} (Hz)	0.18	0.29	(-0.24,-0.06) .006 [+]	0.24	0.37	(-0.22,-0.05) .002 [+]
Fmean_{ml} (Hz)	0.23	0.28	(-0.16,0.07) .286	0.42	0.41	(-0.26,0.27) .510
F80_{ap} (Hz)	0.05	0.13	(-0.24,0.05) .216	0.11	0.21	(-0.20,0.14) .286
F80_{ml} (Hz)	0.42	0.44	(-0.27,0.15) .386	0.46	0.48	(-0.60,0.48) .311
SEnt_{ap}	0.66	0.78	(-0.38,-0.03) .004 [+]	0.96	1.15	(-0.32,-0.02) .006 [+]
SEnt_{ml}	0.68	0.68	(-0.10,0.20) .182	1.28	1.25	(-0.21,0.28) .638
CrossSEnt	1.40	1.37	(-0.14,0.56) .131	1.66	1.68	(-0.31,0.34) .730

Table 7.7. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of postural indices extracted from the *rambling* component of the center of pressure (*COP*) signal during upright stance with eyes open and closed (*Vision* and *No Vision*, respectively). Note: [+] represents significant *Vision* effect ($p < 0.02$).

	Control Group			Senior Group		
	Vision	*No Vision*	95% CI of the difference and p value	*Vision*	*No Vision*	95% CI of the difference and p value
$StabArea_{RM}$ (cm^2)	0.62	0.94	(-1.80,0.17) .004 [+]	1.06	1.52	(-0.82,0.48) .272
Rambling component of the *COPap*						
$Range_{RMap}$ (cm)	1.74	2.18	(-1.00,0.15) .131	2.22	2.55	(-0.57,0.40) .551
RMS_{RMap} (cm)	0.31	0.37	(-0.14,0.05) .168	0.44	0.44	(-0.06,0.08) .826
MV_{RMap} (cm/s)	0.21	0.35	(-0.21,-0.10) .003 [+]	0.30	0.42	(-0.19,-0.04) .001 [+]
$MJerk_{RMap}$ (cm/s^3)	2.20	3.22	(-2.40,-0.47) .003 [+]	3.33	4.64	(-2.40,-0.36) .002 [+]
$Fmean_{RMap}$ (Hz)	0.09	0.15	(-0.11,-0.04) .003 [+]	0.10	0.17	(-0.09,-0.03) .004 [+]
$F80_{RMap}$ (Hz)	0.12	0.29	(-0.19,-0.09) .003 [+]	0.17	0.30	(-0.18,-0.06) .002 [+]
$SEnt_{RMap}$	0.29	0.43	(-0.17,-0.05) .003 [+]	0.32	0.43	(-0.14,-0.04) .003 [+]
Rambling component of the *COPml*						
$Range_{RMml}$ (cm)	0.76	1.13	(-1.29,0.20) .050	1.09	1.16	(-0.50,0.27) .470
RMS_{RMml} (cm)	0.14	0.19	(-0.10,-0.01) .016 [+]	0.18	0.21	(-0.09,0.06) .330
MV_{RMml} (cm/s)	0.12	0.19	(-0.11,-0.02) .003 [+]	0.16	0.19	(-0.08,0.03) .074
$MJerk_{RMml}$ (cm/s^3)	1.73	2.10	(-1.58,0.25) .033	1.63	2.00	(-0.69,0.40) .300
$Fmean_{RMml}$ (Hz)	0.13	0.17	(-0.08,0.01) .248	0.13	0.17	(-0.06,0.01) .177
$F80_{RMml}$ (Hz)	0.27	0.37	(-0.17,0.01) .155	0.23	0.33	(-0.14,-0.01) .035
$SEnt_{RMml}$	0.39	0.42	(-0.08,0.07) .859	0.40	0.45	(-0.08,0.03) .414

Table 7.8. Median and 95% confidence interval of the difference across participants (*Control* and *Senior*) of postural indices extracted from the *trembling* component of the center of pressure (*COP*) signal during upright stance with eyes open and closed (*Vision* and *No Vision*, respectively). Note: [+] represents significant *Vision* effect ($p < 0.02$).

	Control Group			Senior Group		
	Vision	*No Vision*	95% CI of the difference and *p* value	*Vision*	*No Vision*	95% CI of the difference and *p* value
$StabArea_{TR}$ (cm²)	0.07	0.18	(-0.23,-0.06) .004 [+]	0.21	0.43	(-0.32,-0.02) .005 [+]
Trembling component of the *COPap*						
$Range_{TRap}$ (cm)	0.71	1.47	(-1.57,-0.40) .006 [+]	1.48	2.01	(-1.16,-0.04) .008 [+]
RMS_{TRap} (cm)	0.08	0.16	(-0.12,-0.05) .003 [+]	0.16	0.23	(-0.10,0.00) .021
MV_{TRap} (cm/s)	0.23	0.44	(-0.26,-0.13) .003 [+]	0.35	0.49	(-0.26,0.02) .012 [+]
$MJerk_{TRap}$ (cm/s³)	9.58	15.38	(-10.5,-4.2) .003 [+]	14.87	18.59	(-10.2,0.3) .002 [+]
$Fmean_{TRap}$ (Hz)	0.53	0.49	(-0.03,0.11) .131	0.41	0.38	(-0.08,0.10) .470
$F80_{TRap}$ (Hz)	0.84	0.78	(-0.03,0.18) .182	0.67	0.61	(-0.11,0.15) .363
$Sent_{TRap}$	0.54	0.55	(-0.07,0.07) .790	0.48	0.45	(-0.06,0.09) .245
Trembling component of the *COPml*						
$Range_{TRml}$ (cm)	0.53	0.67	(-0.40,0.15) .424	0.63	0.81	(-0.34,0.10) .158
RMS_{TRml} (cm)	0.05	0.07	(-0.04,0.01) .308	0.07	0.08	(-0.03,0.01) .046
MV_{TRml} (cm/s)	0.11	0.15	(-0.10,0.02) .155	0.16	0.19	(-0.06,0.02) .152
$MJerk_{TRml}$ (cm/s³)	3.78	5.15	(-3.96,0.70) .026	7.06	7.30	(-1.97,0.90) .158
$Fmean_{TRml}$ (Hz)	0.46	0.49	(-0.10,0.03) .110	0.41	0.42	(-0.04,0.06) .950
$F80_{TRml}$ (Hz)	0.74	0.76	(-0.13,0.07) .131	0.68	0.63	(-0.06,0.12) .382
$Sent_{TRml}$	0.42	0.41	(-0.05,0.03) .790	0.46	0.47	(-0.05,0.03) .826

When visual input was not available, **older adults** kept fairly similar body sway characteristics as they presented when standing with eyes open, except for a faster and more irregular anterior-posterior sway. Wilcoxon signed-rank tests revealed significant higher *TMV*, MV_{ap}, $Fmean_{ap}$, and $SEnt_{ap}$ during bipedal stance with eyes closed compared to eyes open. No effects of visual input in older adults were found for the other spatio-temporal, frequency, and structural variables. See all *p*-values for the effects of *Vision* on postural indices extracted from the COP in *Table 7.6*.

The effects of visual input on the rambling and trembling components of the postural sway in older adults revealed a faster, more irregular, and less smooth anterior-posterior rambling; and a larger, faster, more variable, and more irregular anterior-posterior trembling. Wilcoxon signed-rank tests revealed a significant increase on $MVel_{RMap}$, $MJerk_{RMap}$, $Fmean_{RMap}$, $F80_{RMap}$, and $SEnt_{RMap}$ when older adults closed their eyes. No effects of *Vision* were found for the other rambling variables. Regarding the variables extracted from the trembling component of the COP, statistical tests revealed a significant increase on *StabArea*, $Range_{TRap}$, MV_{TRap}, and $MJerk_{TRap}$. No effects of *Vision* were found for the other trembling variables. See all *p*-values for the effects of *Vision* on postural indices extracted from the rambling and trembling components of the COP in *Tables 7.7* and *7.8*.

Finally, additional statistical tests revealed different effects of visual input on postural behavior of young adults compared to the findings in seniors. In general, **young adults** presented a larger, faster, less smooth, and more irregular sway when visual input was not available. Wilcoxon signed-rank tests confirmed significant effects of *Vision* on *StabArea*, $Range_{ml}$, RMS_{ml}, *TMV*, MV_{ap}, MV_{ml}, *TMJerk*, $MJerk_{ap}$, $MJerk_{ml}$, $Fmean_{ap}$, and $SEnt_{ap}$. No effects of *Vision* were shown for the other postural indices in control participants. See all *p*-values for the effects of *Vision* on postural indices extracted from the COP in *Table 7.6*.

The effects of temporary removal of visual input in young adults were also observed in changes in both rambling and trembling components of postural sway. Statistical tests revealed that young adults significantly increased the rambling and trembling areas when they performed bipedal stance with eyes closed, compared to open eyes. In the anterior-posterior direction, young adults presented a faster, less smooth, and

more irregular rambling and a larger, faster, more variable and less smooth trembling when they closed their eyes. These results were confirmed by Wilcoxon signed-rank tests on factor *Vision* for MV_{RMap}, $MJerk_{RMap}$, $Fmean_{RMap}$, $F80_{RMap}$, $SEnt_{RMap}$, $Range_{TRap}$, RMS_{TRap}, MV_{TRap}, and $MJerk_{TRap}$. In the medio-lateral direction, young adults significantly increased their rambling variability (RMS_{RMml}) and velocity (MV_{RMml}). No significant effects of *Vision* in control participants were found for the remaining variables extracted from the rambling and trembling components of the COP signal. See all *p*-values for the effects of *Vision* on postural indices extracted from the rambling and trembling components of the COP in *Tables 7.7* and *7.8*.

7.5. DISCUSSION

The present study investigated the age-related changes in features of human body sway underlying postural control of unperturbed stance. The postural indices selected to assess body sway represented multiple dimensions of analysis and were able to detect subtle changes in postural control of upright stance. In general, older adults tend to oscillate more and faster in both directions, increase their medio-lateral sway frequency, present a more irregular and random body sway pattern in both directions, and modify both rambling and trembling components of their postural sway.

Age-related changes in postural sway during unperturbed stance.

As the individual grows older, changes in the physiological systems and sensorimotor integration have a negative impact on postural control. Gross motor changes can be evaluated clinically by observing movement strategies when a challenging situation is presented to the individual. For example, a balance deficit while standing on one leg can be visually observed by a large sway of the trunk and a high guard of the arms to maintain equilibrium. In this study, biomechanical tools were used to examine aspects of postural control that are not detected during clinical examination. Traditional and novel postural indices in spatio-temporal, frequency, and structural domains were

extracted from the COP signal recorded from healthy young and older adults performing unperturbed stance.

Regarding the functional base of support, King et al (1994) reported a significant decrease in the anterior-posterior direction in women after the age of 60 years, with a reduction of 16% per decade thereafter. Duncan et al (1990) also reported an age-related decline in the maximum anterior-posterior body oscillation. However, the results from the present study found no evidence of a significant difference between the maximum anterior-posterior and medio-lateral limits of stability between young and older adults. These discrepant findings may be due to the fact that the senior group in the present study included only healthy nonfaller older adults; whereas other studies may have had mixed nonfallers and fallers, older (65–74 years), older old (75–84 years) and oldest old (over 85 years) adults; or healthy individuals with health comorbidities. Therefore, it is suggested that the functional base support tends to decline as the individual ages, but it seems to still be preserved in healthy nonfaller older adults.

Most of the spatio-temporal variables were sensitive to detecting the effects of age-related physiological changes on postural sway. The larger, faster, and more variable body sway in older adults has been previously described in the literature (Maki et al 1990, Prieto et al 1996, Amiridis et al 2003, Choy et al 2003, Benjuya et al 2004, Freitas et al 2005, Seigle et al 2009, Vieira et al 2009, Sarabon et al 2013, Silva et al 2013, Wiesmeier et al 2015). The natural age-related decline in sensory function may contribute to older adults oscillating more before the system recognizes that the COP is getting closer to the limits of stability. The system, then, seems to correct the unstable position of the COP by moving it back faster to restore equilibrium. Overshooting corrections may also explain the larger and more variable body sway in older adults. Interestingly, the sway jerkiness, suggested by Mancini et al (2011,2012) as an alternative measure of body sway, was able to detect age-related changes on postural control. The higher sway jerkiness in older adults revealed a less smooth and shakier body sway compared to young adults. This pattern of broken and shaky postural sway in older adults could also be observed during voluntary movements, such as the performance of the *BOUNDARIES* task (see *Figure 7.3*).

Despite common agreement that older adults sway faster in both directions, different findings about the effects of aging on the mean frequency of the COP sway have been reported. In this study, only the sway frequency in the medio-lateral direction increased with aging. This finding agrees with other studies (McClenagham et al 1996, Winter et al 1996), but it is contradictory to the decreased medio-lateral frequency in older adults reported by Sarabon et al (2013) and the increased sway frequency in the anterior-posterior direction reported by Maki et al (1990), Freitas et al (2005), and Wiesmeier et al (2015). Divergences in the literature may be explained by the age of the participants, their health status, and the duration of the bipedal stance in each study. Based on the findings from the present study, it is suggested that, in the first stage of natural aging (65 to 74 years old), postural instability in the medio-lateral direction is more affected than in the anterior-posterior direction. This idea is supported by Wiesmeier et al (2015), who reported a significant higher age-related increase in the medio-lateral sway frequency compared to that for the anterior-posterior direction.

Another major finding in the present study was the effect of the natural process of aging on the dynamics of the postural sway. The sample entropy estimate was extremely sensitive to changes in the pattern of the COP displacement in time for both directions ($p < 0.001$). This novel biomechanical tool revealed a more irregular and random postural sway in older adults, compared to young adults. Previous studies using Multiscale Entropy and Sample Entropy (Duarte and Sternard 2008 and Borg and Laxaback 2010, respectively) reported an increased irregularity of the anterior-posterior postural sway in older adults, but no significant difference in the medio-lateral sway. However, the short length of the COP data analyzed (60 and 30 seconds, respectively) could explain their different findings.

A few other studies have reported other applications of entropy analysis in a different population. *Rényi* entropy revealed an increased irregularity of the COP sway in the first day of a concussion episode (Gao et al 2011). Conversely, individuals with Ehlers-Danlos Syndrome (hypermobility type) presented a less random COP sway, measured by smaller approximate and sample entropies (Rigoldi et al 2013). In addition, contradictory findings using sample entropy estimates were reported for the irregularity and randomness of the postural sway in dancers. Stins et al (2009) reported increased

irregularity of body sway in dancers compared to nondancers, whereas Perez et al (2014) reported no significant difference in the regularity and randomness of the body sway between dancers and nondancers. Once more, different findings among studies may be due to the computational approach used and the duration of the task analyzed.

In summary, despite the promising usefulness of entropy estimates to measure the dynamics of the postural sway, more studies are needed to investigate the structural mechanisms behind postural control in different populations and under different tasks and conditions. We speculate that higher COP entropies showing increased body sway irregularity may be related to the use of more feedback mechanisms of postural control, whereas lower entropies showing a less random and more regular body sway may be related to not only feedback mechanisms but also feedforward (anticipatory adjustments) mechanisms of postural control.

Finally, the use of rambling and trembling decomposition to investigate postural control in older adults revealed increased rambling and trembling oscillations. In general, the rambling-trembling hypothesis suggests a superposition of two processes of upright stance control. Zatsiorsky and Duarte (2000) suggested that the rambling mechanism reflected neural mechanisms of supraspinal origin, while the trembling component reflected spinal reflexes and changes in the intrinsic mechanical properties of muscles and joints. Most of the age-related changes in the spatio-temporal domains happened for the rambling and trembling components in the anterior-posterior direction. Sarabon et al (2013) also found age-related changes on both components, such as increased rambling and trembling mean velocity in the anterior-posterior direction. Despite the lack of studies using this innovative approach to investigate postural control in older adults, some studies reported changes in rambling and trembling signals under different experimental conditions. Danna-dos-Santos et al (2008) asked healthy adults to maintain their upright stance while an online feedback of the COP displacement was provided. In order to maintain their COP inside different targets in the screen, participants decreased their rambling sway area and increased their trembling area as the target difficulty increased. In another study, the manipulation of the base of support also imposed changes in the behavior of rambling and trembling trajectories (Mochizuki et al 2006). In addition, Shin et al (2011) reported a higher ratio of trembling to COP in individuals with

multiple sclerosis compared to age and gender matched controls, but no differences in the ratio of rambling to COP. These studies support the hypothesis that the rambling mechanism is activated by central processes searching for postural stability, whereas the trembling mechanism is activated by peripheral processes and reflexes. Therefore, it is suggested that the increased anterior-posterior rambling in older adults may reflect impaired or declined sensorimotor integration. It seems that older adults have some difficulty in integrating information of the body deviation from the vertical with motor commands for corrective actions. As a result, there is an increase in the trajectory of consecutive reference points, which oscillate faster, less smoothly, and closer to the limits of the base of support. The findings also support the idea of the trembling component of the COP signal reflecting the peripheral mechanism of upright stance control. It is suggested that the increased trembling trajectory may be due to the natural age-related changes in the modulation of spinal reflex gains, in the peripheral mechanical properties, and in the pattern of multiple postural muscles contraction.

The effects of a sensory challenging task (temporary visual removal) on postural sway in older adults.

Considering that the simplicity of the unperturbed stance may not be able to detect some age-related changes on postural control mechanisms, a sensory challenging task was included by asking the participants to stand still with their eyes closed. Several studies had reported increased body sway under conditions of reduced or conflicting sensory information in older adults compared to young adults (Woollacott et al 1986, Horak et al 1989, Teasdale et al 1991). When visual input was manipulated in older adults, they increased the velocity, frequency, and randomness of their postural sway only in the anterior-posterior direction. Interestingly, the magnitude of the anterior-posterior sway did not significantly change. The increased postural sway frequency and randomness in older adults when visual input was not available were also identified by Weismeier et al (2015) and Fino et al (2015), respectively. Conversely, Benjuya et al (2004) reported increased anterior-posterior sway amplitude and medio-lateral sway velocity when older adults performed stance with eyes closed, compared to eyes open,

but they analyzed only 20 seconds of unperturbed stance. This greater impact of visual input on the anterior-posterior sway may be explained by the larger range of motion of movements around ankles, knees, and hips in the sagittal plane compared to that in the frontal plane. Furthermore, the rambling-trembling decomposition revealed that the absence of visual input in older adults had more effect on the rambling component of the *COPap*. The findings reinforce the idea of the rambling trajectory representing a central mechanism of postural control, since sensorimotor integration was challenged during upright stance with eyes closed.

In contrast, young adults presented similar effects of visual input on the frequency and structural variables as older adults: increased sway frequency and randomness in the anterior-posterior direction. However, young adults presented more changes in the spatio-temporal domain compared to that for older adults. Under temporary removal of visual information, young adults increased not only their anterior-posterior sway velocity and jerkiness like the older adults, but also their medio-lateral sway magnitude, variability, and jerkiness. Likewise, Benjuya et al (2004) pointed out more spatio-temporal changes in postural sway in young adults compared to older adults when visual input was not allowed.

This greater effect of visual input in young adults compared to older adults suggests changes in sensory weighting in older adults. Taken together, these findings agree with the increase in the reliance on visual input to control balance up to 65 years old, followed by a decrease in the contribution of visual input on balance control reported by Lord and Ward (1994). In another study, a greater contribution of vision in nonfallers compared to fallers was reported (Turano et al 1994), which may add to the hypothesis that individuals rely less on visual input as they age and present signs of balance deficits, such as fall episodes. An alternative explanation for this greater effect of visual input in young adults might be due to the pattern of cocontraction of muscles acting around the ankle joints in older adults. The agonist-antagonist cocontraction, previously reported during bipedal stance with eyes either open or closed (Benjuya et al 2004), might be the mechanism used by older adults to avoid bringing their COP closer to the limits of stability. In addition to this, the fear of reaching the limits of stability may also represent another factor to explain why older adults do not increase the magnitude of postural sway

when they close their eyes. Contrary to older adults, young adults seemed to be comfortable to explore their base of support by increasing their postural sway when standing with eyes closed.

In addition to different effects of visual information in young and older adults, age-related changes in postural control of unperturbed stance were found even with eyes closed. Regarding spatio-temporal variables, only the mean velocity and mean jerkiness responses were sensitive enough to detect change in postural sway between young and older adults. The faster postural sway when eyes were either open or closed has been reported before (Baloh et al 1994, Prieto et al 1996, Benjuya et al 2004, Seigle et al 2009). Interestingly, the effects of age on sway area and ranges differ among studies. Benjuya et al (2004) found an increased length of the COP path and sway amplitude in older adults with eyes closed compared to young adults. Conversely, Seigle et al (2009), Vieira et al (2009), and the present study found no significant effect of age when performing unperturbed stance with eyes closed. Such disparity in results may be due to different methods used to compute the sway area. In addition, the findings reported here regarding the increased irregularity of the anterior-posterior COP pattern as the individual ages is consistent with the higher sample entropy of the COPap displacement in older adults reported by Borg and Laxaback (2010). They also reported that fallers have a more irregular sway than nonfallers when standing with eyes closed.

7.6. CONCLUSIONS

The study reinforces the role of the COP sway as a crucial tool to assess postural control in older adults. Not only traditional measures, but also innovative postural indices were included to cover different aspects underlying the mechanisms of balance control. The effects of the natural process of aging on balance control can be detected even in the first stages of aging (65 to 74 years old) using spatio-temporal, frequency, and structural variables. As the individual gets older, they tend to increase their body sway amplitude, irregularity, velocity, jerkiness, and frequency to control unperturbed stance. They also

seem to move faster and less smoothly among more distant reference points, in addition to an increased movement around this moving reference point.

Interestingly, the challenging condition of removing visual input in older adults led to an increased velocity, irregularity, and frequency of postural sway in the anterior-posterior direction. This very noticeable effect of visual input on the anterior-posterior sway also was observed in young adults, indicating an increased reliance on visual input compared to older adults. In addition, a few effects of the natural process of aging on postural control were still present when young and older adults were standing without visual input.

In conclusion, balance assessment using force platforms are of the utmost importance to identifying characteristics of postural sway affected by natural aging. The objective postural indices extracted from COP signals in this study provided useful knowledge regarding postural markers for potential balance instability and risk of falls. Moreover, the findings also advance knowledge for health professionals to direct interventional protocols focusing on balance control in older adults.

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CHAPTER 8
SUMMARY OF CONCLUSIONS

The dissertation focused on the investigation of the effects of aging on the mechanisms of multi-muscle control and on postural sway behavior during upright stance. A series of studies were performed and each chapter presented a discussion of its results. In general, the findings corroborate the plasticity of the aging CNS in using compensatory mechanisms to control upright stance and avoid falls. The purpose of this chapter was to present the main conclusions from the dissertation.

8.1. Intermuscular Coherence Analysis as a novel approach to investigate the formation of postural muscle synergies

Based on the *Motor Redundancy (Abundance) Hypothesis*, the CNS unites motor components into functional groups to overcome the large number of degrees of freedom of the system. The intermuscular coherence approach used in the dissertation provided new information on the strategies used by the CNS of both young and older adults to generate and distribute neural commands to multiple postural muscles.

Intermuscular coherence analysis at lower frequency bands revealed to be a promising tool to advance studies in the mechanisms underlying human postural control. Findings from the dissertation opened a broad horizon of possibilities to study the formation of postural muscle synergies. Further studies are needed to advance knowledge on how the aging CNS reorganizes correlated neural inputs to multiple postural muscles under a variety of challenging tasks. This novel approach should also be expanded to study the generation of synergistic muscle groups in individuals with different levels of balance deficits.

8.2. Aging is associated with a reorganization of correlated neural inputs to postural muscles forming synergistic muscle groups

Healthy nonfaller older adults present signs of correlated neural drive to postural muscles.

Studies presented in *Chapters 5* and *6* suggested that the formation of synergistic muscle groups during unperturbed stance in older adults was driven by correlated neural inputs to different postural muscles, as happened in young adults (see *Chapters 3* and *4*). Older adults showed signs of synchronization at lower frequency bands for three distinct muscle groups (push-forward M-mode, push-back M-mode, and antagonist group). No correlated neural drive was found for muscles pairs formed by one anterior and one posterior, non-antagonist, muscles. These findings suggested that the aging CNS was able to control a large number of degrees of freedom by forming synergistic muscle groups, as young adults did.

The frequency distribution of correlated neural inputs to postural muscles changes with age.

A few age-related changes in the frequency distribution of neural drive to postural muscles were observed. Results presented in *Chapter 6* revealed synchronization at lower frequency bands of posterior muscles in a larger frequency band, as well as synchronization of antagonist muscles in a smaller frequency band, compared to that in young adults. In contrast, the frequency band of the synchronization at lower frequency bands of anterior muscles in older adults was similar to that observed in young adults, and no significant synchronization at lower frequency bands of mixed muscles was observed for either older or young adults. These findings suggested that the aging CNS reorganized the neural drive to postural muscles during upright stance.

Healthy nonfaller older adults present stronger intermuscular coherence at lower frequency bands, compared to young adults.

In general, intermuscular coherence at lower frequency bands was stronger in healthy nonfaller older adults, compared to that in healthy young adults (see *results* in *Chapter 6*). Age-related increase in the magnitude of correlated neural inputs within the frequency band of 1–10 Hz was reported for three synergistic muscle groups (push-forward, push-back, and antagonist muscle groups). Once more, such increased synchronization at lower frequency bands may reflect changes in the control strategy and compensatory postural adjustments in older adults. These findings corroborate the plasticity of the aging CNS in reorganizing the formation of synergistic postural muscle groups in order to control upright stance and avoid falls.

Healthy nonfaller older adults keep similar control of multiple muscles under the sensory challenging task of upright stance without visual input.

Despite reduced frequency distribution and strength of the synchronization at lower frequency bands of antagonist postural muscles observed in older adults when they closed their eyes, no significant change in correlated neural inputs to the other synergistic muscle groups suggested minimal effects of visual input on the control strategy in older adults (see *results* in *Chapter 6*). Interestingly, young adults presented a reorganization of neural drive to postural muscles when visual input was temporarily absent (see *results* in *Chapters 4* and *6*). When healthy young adults closed their eyes, their intermuscular coherence at lower frequency bands decreased and was no longer significant. Taken together, these findings corroborate age-related difficulty in sensory reweighting and suggest that healthy nonfaller older adults rely less in visual information than healthy young adults.

8.3. The importance of multiple postural indices to understand the mechanisms underlying age-related changes in postural sway

Balance assessment using force platform to record COP displacement is crucial to characterize postural sway in different populations. Several studies have shown reliable

variables extracted from COP signals to measure postural control. However, one variable alone may describe only part of the behavior. For example, how does one explain the balance of dancers and older adults based only on their larger body sway compared to healthy young non-dancer adults?

The study presented in *Chapter 7* used not only traditional variables to quantify postural sway, but also novel measurements. The use of a force platform and the *BalanceLab* software to extract postural indices from multiple domains enabled a more comprehensive insight of the effects of aging on postural sway behavior. The results suggested the use of postural indices from multiple domains to cover different aspects underlying the mechanisms of postural control of individuals with different levels of balance control.

This new knowledge is crucial to direct efforts of health professionals to optimize treatment and rehabilitation of age-related balance disorders. The use of such postural indices characterizing body sway behavior of older adults during unperturbed upright stance should also be expanded to different functional and challenging situations, such as unipedal stance, tandem stance, and reaching or releasing an object. Moreover, this approach should be explored to advance knowledge of the mechanisms used by other populations with different levels of balance deficits, such as individuals with a history of traumatic brain injury, Parkinson, cerebral vascular accident, diabetes, and other orthopedic and neurological disorders.

8.4. Aging is associated with changes in body sway behavior

Healthy nonfaller older adults present a larger, faster, less smooth, and more irregular body sway, compared to healthy young adults.

The effects of aging on postural control can be observed by changes in the behavior of body sway. Postural indices extracted from COP displacements in spatio-temporal, frequency, and structural domains are able to detect such effects. Although healthy nonfaller older adults (65 to 74 years old) preserved their functional base of

support during unperturbed bipedal stance, they increased their sway variability, amplitude, velocity, and frequency. In addition, body sway in older participants was characterized by jerky movements and a more irregular pattern of body oscillation in time compared to that in young adults. Both declines in sensorial and motor functions seem to impact the control of the position of the center of pressure (COP) within the base of support. It seems that older individuals tend to oscillate closer to the limits of stability and, when trying to move the COP back to a safer position, they do it fast. In addition, it was suggested that overshooting postural corrections may be one reasons for increased sway variability, shaky sway pattern, and decreased predictability of the COP position in time observed in older individuals.

Postural control adjustments in older adults were also detected by changes in both rambling and trembling components of the COP displacement in the anterior-posterior direction. This finding corroborates the decline in both central and peripheral processes involved in postural control as the individual grows older.

Interestingly, the aforementioned changes in the mechanisms of postural control can be detected even in the early stage of aging. They may be interpreted as compensatory adjustments to the progressive and non-homogeneous decline of physiological functions. One can speculate that the aging system may be over-activating postural reactions to avoid falls, resulting in ongoing overshooting responses.

Healthy nonfaller older adults present a faster and more irregular body sway under the sensory challenging task of upright stance without visual input.

Under temporary visual removal, older adults kept similar magnitude of body sway, but their body oscillated faster and in a more irregular pattern compared to upright stance with eyes open. The temporary absence of visual input also seems to affect the sensorimotor integration in older adults. The resulting changes in the rambling component of the body sway when older adults closed their eyes corroborate the idea of the link between rambling and central processes of control.

Interestingly, young adults presented more changes in postural indices when they closed their eyes than older adults did. It was suggested that such differences may be due

to age-related changes in multisensory inputs and sensorimotor integration. The natural decline in visual function and the difficulty on sensory reweighting experienced by older adults may explain some of the reasons why they did not rely in visual inputs as much as young adults.

8.5. Contributions to science and clinical relevance

A variety of traditional and novel objective techniques to quantify different features of the human postural control, such as the center of pressure and center of mass behavior, forces at the surface, torques around body joints, muscle activity, joint displacement, body segment position, and brain (electrical) activity, are available. However, two gaps hinder full application of research findings to clinical settings. The first gap regards the interpretation of data. Simple descriptions of recorded data have no value if they are not translated into functional meanings. The second gap regards the application of research findings by health professionals. Physical therapists, physicians, and other health professionals should benefit from posturography and other techniques in order to customize their interventions according to their client's needs.

In general, the methodological approaches used in the dissertation to quantitatively analyze muscle synergy and body sway behavior provided a step forward to understanding the mechanisms underlying postural control in older adults. The application of postural control research findings to clinical assessment, prevention, and rehabilitation of balance disorders are fundamental to improve the individual's quality of life. In conjunction with clinical functional and systems assessment, findings reported in the dissertation regarding quantitative posturography may be used not only to identify balance problems and predict fall risk. They also provide valuable information regarding the causes of age-related balance problems and the mechanisms adopted by the CNS to compensate for an individual's balance deficits.