Developing and Evaluating New Methods for Assessing Postural Control and Dynamics

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ABSTRACT

Falls are the leading cause of injuries among older adults (>65) and frequently result in reduced mobility, loss of independence, decreased quality of life, injury, and death. Extensive research has been conducted regarding postural coordination and control, and other mechanisms/processes involved in maintaining postural stability. However, there is relatively limited knowledge regarding the patterns of joint coordination, the underlying postural controller, and efficient methods to assess passive and active musculoskeletal properties relevant to balance. In the current work, three new methods were developed to address these limitations and also to better understand the effects of localized ankle muscle fatigue, gender, and aging on postural coordination and control.

First, two methods were used to evaluate postural coordination. A wavelet coherence approach was developed and applied to assess the level and pattern of coordination between pairs of joints (i.e., ankle-knee, ankle-trunk, and ankle-head). In addition, the uncontrolled manifold method was implemented for evaluation of potential whole-body coordination control goals. Clear patterns of intermittent wavelet coherence were evident, indicating that joint coordination is intermittently executed. Both in-phase and anti-phase coherence were detected over frequencies of 2.5 – 4.0 Hz. Shoulder and head kinematics appeared more likely than the whole-body center of mass as control goals for

whole body coordination. Both aging and ankle muscle fatigue led to a reduction of joint coordination.

Second, an intermittent sliding mode controller was developed to model quiet upright stance. In contrast to most previous postural controllers, which assume continuous control, an intermittent controller was considered more consistent with recent evidence on muscle activity and the results of the first study on postural coordination. The sliding mode controller was able to accurately track kinematics and kinetics, and generated passive and active ankle torques comparable with previous results. Ankle fatigue led to an increase in active ankle torque especially among young adults and males.

Third, a new method was developed to estimate passive and active mechanical properties at the ankle (e.g., stiffness and damping). This method was inspired from intermittent control theory, and the earlier results noted. As opposed to conventional methods, this new method is computationally efficient and does not require external mechanical or sensory perturbations. The method yielded a ratio of passive to active ankle torques consistent with earlier evidence, and larger passive and active ankle torques among males and older adults. A post-fatigue increase of active ankle torque was estimated, especially among males and young adults.

In addition to providing new analytical methods, the noted studies suggest that older adults have decreased joint coordination and increased ankle stiffness. As a practical implication of this, fall prevention training programs may benefit from seeking to develop appropriate joint coordination strategies and ankle stiffness magnitudes. To expand on the current work, future research should consider measuring muscle contraction characteristics at multiple joints and in different postures or activities.

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Chapter 1 Introduction

1.1 Background

Falls frequently lead to injuries and/or fatalities, particularly among older adults. It is estimated that one out of every three older adults (>65 years old) falls annually (Skinner et al., 1986, Tinetti, 2003, Kannus et al., 2005, Friedman and Forst, 2007, Gillespie et al., 2009a, Gillespie et al., 2009b), resulting in a total of 2.1 million older adults suffering nonfatal fall injuries every year. In 2010, 2.3 million older adults suffering from nonfatal fall injuries were treated in emergency departments, and over 662,000 of patients were subsequently hospitalized (CDC, 2012b). Falling is also the leading cause of work-related injuries and fatalities. In 2011, 40 percent of all nonfatal occupational injuries and illness cases required days away from work, and among these, falls were the leading cause, accounting for 15 percent (BLS, 2012).

Fall-related accidents also cause considerable economic loss. Between 1980 and 1997, the total cost resulting from falls was nearly $85 billion in the US. In 2000 alone, the total direct medical costs resulting from falls were greater than $19 billion. This figure includes $179 million resulting from fatal falls and $19 billion resulting from nonfatal fall injuries (CDC, 2012a). The direct cost of one disabling fall injury is about $28,000, with indirect costs totaling about $46,000, and the cost associated with work-related fall fatalities is estimated to be over $940,000 (Lehtola et al., 2010). Further, the mean cost of falls among older adults has been estimated to be $10,749 (Davis et al., 2010), and mean hospital charges for fall related injuries are in excess of $20,000 (Davis et al., 2010).

Falling can also lead to severe physical injuries. Nearly 8% of workers who experience an occupational fall require partial or total ambulatory assistance at the time of hospital discharge, and 20% - 30% of older adults suffer moderate to severe injuries when they fall (Friedman and Forst, 2007). Falls are the most common cause of nonfatal injuries as well as hospital admissions for traumatic injuries (Stevens, 2005). In 2009, U.S. emergency departments treated 2.2 million nonfatal fall injuries among older adults. Among them, more than 582,000 were hospitalized (CDC, 2011) with an average hospitalization duration of 5.5 days (Davis et al., 2010).

1.2 Factors Contributing to Fall Risk

Multiple factors contribute toward fall-related accidents, and are clustered here into three groups. The first group of risk factors are related to human physical performance and cognitive/emotional status, such as age (Bailer et al., 2003), gender (Park, 2002), muscle weakness (Sherrington et al., 2008b, Peeters et al., 2010), gait/balance/visual deficit (Rubenstein and Josephson, 2002, 2006), depressed mood, cognitive impairment and disability (Barrett-Connor et al., 2009), knowledge of injuries associated with falls, risk perception, training, mentoring (Lipscomb et al., 2003), and personal protective equipment (Hsiao and Simeonov, 2001). The second group of risk factors are associated with load handling (Leamon and Patrice, 1995), physical exertion, fatigue (Swaen et al., 2003, Lin et al., 2009), and task complexity (Hsiao and Simeonov, 2001). The third group of risk factors are related to the environment and include weather (Lipscomb et al., 2003), visibility (Hsiao and Simeonov, 2001), and physical interactions (restricted support surfaces, inclined support surfaces, material properties) (Hsiao and Simeonov, 2001). These multiple factors contributing to falls could also imply that fall interventions should be multifaceted, as discussed below.

1.3 Fall Prevention Interventions

Researchers have dedicated significant effort toward preventing falls. A variety of intervention approaches have been used, which have largely addressed the following aspects: 1) the mental-physical domain, such increasing muscle strength, physical rehabilitation, vision improvement (Day et al., 2002, Strandberg and Pitkälä, 2007, Sherrington et al., 2008a), reducing the use of psychotropic medication, and improving cognitive conditions related to diseases such as dementia (Rubenstein et al., 1996, Shaw et al., 2003, Kannus et al., 2005), 2) the environment, such as home, nursing home, and community center based hazards (Connell, 1996, Cumming et al., 1999, Cumming, 2002, Clemson et al., 2008), 3) use of supplementation, such as providing vitamin D and calcium to those at a high risk of falling (Bischoff-Ferrari et al., 2009, Batchelor et al., 2010, Handoll, 2010). Additionally, multi-factorial fall risk assessment and management programs, considering multiple factors such as exercise, environment, and other falls contributory factors, have also been developed and applied (Chang et al., 2004).

Among these interventions, general physical exercise programs such as walking appear to increase flexibility, physical strength, and agility, which are essential for balance and body functional capability (Sherrington et al., 2004). Muscle strengthening, walking, and balance retraining exercises have been shown to reduce occurrences of falls compared to a control group (Campbell et al., 1999). For example, leg muscle strengthening and body flexibility training reduced the incidence of falls by 11.1% (Day et al., 2002).

Home-based exercises conducted three days/week lasting for a year have been shown to improve physical strength and postural balance control (Gardner et al., 2001). Similarly, one-year-long home exercises designed for older adults have been found to improve balance as well as reduce the frequency of falls and fall-related injuries (Robertson et al., 2005).

Progressive and high intensity physical resistance training has been used to improve ankle and trunk strength, with supplementary evidence suggesting it can reduce the risk of falling (Liu-Ambrose et al., 2004, Sherrington et al., 2004, Sherrington et al., 2008b). Diverse types of physical training have also been used, including: seated resistance training, chair-assisted knee bends, wall squats, and heel raises. It has been suggested that these exercise modalities can help to improve balance (Lord et al., 2005). Additionally, exercises and rehabilitation programs to enhance joint coordination have been implemented for fall prevention, including Yoga (Brown et al., 2008, Schmid et al., 2010, Jones, 2011), dynamic walking (walking with upper body or lower body twisting), single limb stance, balance-ball training, and Tai Chi (Sherrington et al., 2004, Choi et al., 2005, Li et al., 2005, Sherrington et al., 2008). Several forms of mechanical or electrical-based vibration techniques have also been proposed in an attempt to increase postural stability, including the use of stochastic electrical noise (Collins et al., 1996), vibrating and facilitatory insoles (Collins et al., 2003b, Priplata et al., 2003, Priplata et al., 2006, Hijmans et al., 2007, Perry et al., 2008), and mechanical stimulation delivered to the skin surface of the lower back (Reeves et al., 2009).

1.4 Research Interests

Falls are not caused by a single factor, but are rather characterized by multidisciplinary and multi-factorial elements (Gillespie et al., 2009a), and this complexity and diversity imposes challenges to fall prevention research. Furthermore, it has also been argued that fall prevention intervention strategies do not work universally (Wu, 2002), but are dependent upon individual demographics, physical and cognitive status, required occupational tasks, and quality/safety of the work environment. All considered it is infeasible for a singular research effort to adequately investigate all these factors and aspects. In the current work, we are interested in studying physical mechanisms that include joint coordination, the central postural controller, and passive and active joint stiffness.

These mechanisms are considered important and interesting for fall prevention for several reasons. First, joint coordination is fundamental for many motor tasks such as rapid force generation, walking, and upright stance postural maintenance (Hollands et al., 2004, Barry et al., 2005, Hsu et al., 2012). Joint coordination appears to be important for fall prevention, since, for example, whole body coordinated exercise such as Tai Chi can help reduce the risks of falls (Gregory and Watson, 2009, Logghe et al., 2010). Second, postural balance maintenance is regulated and controlled through certain mechanisms, namely postural control. Ongoing efforts have been dedicated to studying different postural controllers and their roles in balance control, the results of which are helpful to interpret the mechanisms of falls (Collins and Luca, 1993, Kuo, 1995, Kiemel et al., 2008, Gawthrop et al., 2009). Third, the upright stance posture is maintained through corrective joint torques (Runge et al., 1999, Winter et al., 2001, Wojcik et al., 2011, Suzuki et al., 2012). In comparing to the postural controller, which is a "central processor" of posture, joint torque is the actual executor, or a postural actuator. A malfunction of the actuator could be related to the increased risks of falls. As such, we need a better understanding of both the central processor and the corrective joint torques for the purpose of fall prevention.

1.5 Research Needs

Several coordination mechanisms have been studied in past research, and have indicated several underlying patterns including ankle-hip coordination phase relationships (Hsu et al., 2007), muscle coordination patterns (Ting and Macpherson, 2005), and coordination control schemes (Alexandrov et al., 2005, Kiemel et al., 2008). However, some fundamental aspects of joint coordination, which may be of particular use in terms of understanding fall causes and suggesting preventive actions, have not been well clarified. For example, the joint coordination phase relationship, the coordination patterns (i.e., continuous versus intermittent), and the coordination goal remain largely unknown. Understanding such time dependent coordination patterns can help to resolve ongoing controversies, such as whether upright stance is maintained continuously or intermittently (Peterka, 2002, Loram et al., 2011). By understanding these mechanisms, it may be possible to develop more effective methods for fall prevention interventions.

Substantial effort has been put forth by many research groups to better understand how the body governs postural control (Collins and Luca, 1993, Peterka, 2000, Morasso and Sanguineti, 2002, Maurer et al., 2006, Vieira et al., 2012). Several controller models have been developed, including sensory information integration controllers (Peterka, 2002, Peterka and Loughlin, 2004), optimal controllers (Kuo, 1995, Qu and Nussbaum, 2010), and predictive controllers (Gawthrop et al., 2009). As of yet, though, there is little consensus regarding the form of the "real" controller responsible for maintaining quiet upright stance. In particular, there are ongoing debates regarding whether postural control is intermittent or continuous (Bottaro et al., 2005). Largely, past research on postural controllers has been mostly explorative, and has been done using relatively complicated mathematical formulations but with little direct empirical results. As such, a deeper and hopefully more thorough understanding of postural control mechanisms is needed, and which can be helpful in solving the debate and providing additional insights regarding fall risks and prevention.

Numerous studies have examined the passive and active components of corrective joint torques. However, different magnitudes of passive and active joint torque have been reported in existing work. For example, one study indicated that passive ankle torque accounts for the major part (>90%) of the entire ankle torque, and was argued to be generated by passive elastic properties of the joint capsule, muscle, tendon, and other soft tissues (Winter et al., 2003). In contrast, numerous other studies have also suggested that active ankle neural muscular control is an important component of the ankle corrective torque (Morasso and Sanguineti, 2002, Kiemel et al., 2008). Most existing measures of passive and active joint torques were obtained through use of mechanical or sensory perturbations (Kiemel et al., 2008). Such approaches require rather complex and expensive experimental configurations (Loram and Lakie, 2002). Methods that do not require such effort to identify passive and active joint torque are thus needed, and if successful, may provide new approaches for understanding general postural control mechanisms and the related risks of falls.

1.6 Effects of Localized Muscle Fatigue on Postural Control

Extensive evidence indicates that localized muscle fatigue (LMF) compromises the neuromuscular control system, resulting in a loss of muscle force and power output (Miura et al., 2004), decreased position sense acuity (Lee et al., 2003), reduced muscle stiffness (Kuitunen et al., 2002), impaired somatosensory feedback (Balestra et al., 1992), and compromised central neuromotor control (Corbeil et al., 2003, Todd et al., 2003). Muscle stiffness, proprioceptive/somatosensory systems, and the central nervous system are all associated with postural control (Simoneau et al., 1995, Burdet et al., 2001, Winter et al., 2001). Consequently, localized muscle fatigue can compromise postural control and increase the risks of falls (Allison and Henry, 2002, Olson, 2010).

Ankle LMF in particular can lead to impaired postural control. For example, ankle LMF compromises ankle proprioceptive input (Vuillerme et al., 2002). Yaggie and McGregor (2002) also found adverse effects on postural control following ankle LMF induced by repeated isokinetic ankle exertions. Similarly, postural sway was found to increase following localized calf muscle fatigue (Vuillerme and Nougier, 2003), and fatigue in different ankle muscle groups can result in compromised postural stability in the associated directions (Salavati et al., 2007). Evidence regarding the effects of ankle LMF on upright stance postural control has been diverse, including changes in postural displacements (Yaggie and McGregor, 2002, Vuillerme et al., 2005), mean velocity, time to boundary, and statistical mechanics variation in the center-of-pressure (Lin et al., 2009), body kinematics patterns (Madigan et al., 2006), and extended feedback time delay (Qu et al., 2009). Still lacking, though, is sufficient evidence regarding the effects of LMF on joint coordination, the postural controller, and passive and active joint torque and stiffness, each of which are critical aspects for postural stability control, and were investigated in this work.

1.7 Effects of Aging on Postural Control

As noted earlier, falls are particularly prevalent and problematic among older adults. There are several well described musculoskeletal changes due to aging, such as decreases in the synthesis of muscle proteins (i.e., myosin heavy chain) and mitochondrial proteins (Grimby and Saltin, 1983, Nair, 2005), reduced muscle mass/volume and cross-sectional area with increased fat infiltration of lean muscle tissues (Borkan et al., 1983, Demerath et al., 2012), and decreased contractile capacity (Grimby and Saltin, 1983, Bruce et al., 1989, Akagi et al., 2009, Canepari et al., 2010). Aging also causes motor neuron changes leading to decreased motor unit discharge rates and increased motor unit discharge variability, decreased lumbosacral spinal cord motor neurons, and the number of motor units (Roos et al., 1997, Doherty, 2003). Changes in muscle biomechanical and motor control parameters also occur that can contribute to fall risk, such as decreased postural coordination and slower recovery reaction time (Tinetti, 2003, Li et al., 2005, Goldberg and Neptune, 2007, Gillespie et al., 2009a). Hence, in this work, investigating the effects of aging on joint coordination, postural control, and passive and active ankle torque were additional research interests.

1.8 Research Scope

This work focused primarily on developing new methods to improve our understanding of the governing postural control mechanisms, specifically those responsible for controlling postural sway, whole body coordination, and ankle joint stiffness and torque. To achieve this, several modeling strategies were developed, applied, and evaluated. Given that aging and localized muscle fatigue cause adverse changes to postural control, the effects of these aspects were evaluated, in terms of joint coordination, postural control, and passive and active joint torque and stiffness. Both joint-level and whole body coordination were quantified using a wavelet coherence method and the uncontrolled manifold method, respectively. A sliding mode postural controller model was also developed to model the quiet upright stance motion control process. Passive and active ankle torque, stiffness, and damping were separately estimated using a new method.

Although quiet upright stance is not a particularly high risk activity, it represents a simple postural configuration, and is a very common task used to investigate postural control (Loram and Lakie, 2002, Collins et al., 2003a, Peterka and Loughlin, 2004, Qu and Nussbaum, 2010). Conclusions obtained from investigating quiet upright stance could be extended to other postures and task, though caution is certainly warranted. Even if the results of current work do not directly generalize to other postures/tasks, the results of investigating quiet upright stance were still considered useful, such as by providing a reference for subsequent investigations of other postures/tasks such as sitting, walking, or materials handling.

1.9 Innovation and Contribution

The current work investigated postural coordination, control, and joint stiffness during upright stance, topics which have been studied by several researchers in past (e.g., Collins and Luca, 1993, Winter et al., 2001, Granata et al., 2002, Peterka, 2002, Casadio et al., 2005, Creath et al., 2005, Moorhouse and Granata, 2007, Qu and Nussbaum, 2010). Based on this earlier work, several extensions (or enhancements) have been made to address inherent limitations.

The first such extension was applied for the evaluation of postural coordination. Most previous research has only focused on coordination that exists in a range of certain frequency bands of joint motion, without a full consideration of time-dependent joint coordination behaviors, and which can be a necessary component of joint coordination (Scholz and Schöner, 1999, Creath et al., 2005, Hsu et al., 2007). Time variant information is important for postural control, and also relates to the risks of falls such as through balance detection and recovery time (Rietdyk et al., 1999, Geurts et al., 2005) and gait recovery time during slip perturbations (Liu and Lockhart, 2009). As such, in the current work time-dependent postural coordination was considered and evaluated. From time-dependent changes in the postural coordination, more complete interpretations were expected, in terms of the time dependent coordination patterns and phase variations, regarding the underlying postural coordination mechanisms during upright stance. In addition, the influences of ankle localized muscle fatigue and aging were determined, to facilitate a better understanding of the impacts of such fatigue on two-joint and whole body coordination.

The second extension addressed the fact that most existing approaches have modeled upright stance using a continuous postural controller (Peterka, 2002, Maurer and Peterka, 2005, Kiemel et al., 2008, Qu and Nussbaum, 2010). Intermittent postural control, however, has been supported by experimental findings, indicating that an intermittent controller may actually underlie human control of upright stance (Loram and Lakie, 2002, Lakie et al., 2003). As such, an intermittent postural controller was developed, and this model was used to explore the effects of LMF and aging on postural control. While intermittent postural control has been modeled previously (Bottaro et al., 2008, Gawthrop et al., 2010, Loram et al., 2011), there are still some attributes of the intermittent controller that remain unknown. Examples include the effects of LMF and aging on the intermittent controller and an appropriate intermittent control model. Two new contributions were made here, related to intermittent postural control modeling. First, the new intermittent model can track experimental postural kinematics and dynamics data, which has not been extensively analyzed in prior models. Second, the model can estimate passive and active joint torques separately.

The third extension was motivated by the evidence summarized above, that joint stiffness is critical for postural control. Most existing methods have measured joint stiffness using complex and costly laboratory instruments, most of which are used to generate motor sensory or mechanical perturbations aimed to separate passive from active joint stiffness. Additionally, during quiet upright stance, available biomechanical models of joint stiffness are typically limited for calculating ankle stiffness (Loram and Lakie, 2002, Casadio et al., 2005) or at most including trunk stiffness (Kiemel et al., 2008). However, it remains extremely challenging for those models to be easily extended to calculate passive and active stiffness of two or more joints without using perturbation-based experimental methodologies. Here, an approach for estimating joint stiffness was developed based on underlying intermittent control theory and experimental evidence (Loram and Lakie, 2002). This new method approach does not require any perturbation methods as noted for earlier studies. Further, the model has high computational efficiency and low modeling effort, and may be extensible to calculate passive and active stiffness of other joints such as trunk and neck. Along with developing the model, it was also applied to estimate the effects of ankle LMF and aging on passive and active ankle stiffness.

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Chapter 2 An Introduction to and Demonstration of the Wavelet Coherence and Uncontrolled Manifold Methods

2.1 Wavelet Coherence

2.1.1 Background

2.1.1.1 Coherence

Coherence analysis, also known as cross spectrum analysis, is used to analyze the extent to which any frequency in one signal is correlated with the same frequency in another signal (Timmer et al., 1998). It thus can be used to identify the coupling between two signals at one or more frequencies. It can also be used to identify the concentration of power at given frequencies as well as phase lags. Coherence can be determined through the following steps.

First, we transfer signals to cross covariance

$$C_{xy}(k) = \frac{1}{N}\sum_{i=0}^{N-k} X(i)\, Y(i+k) \tag{1}$$

where X and Y are two different input signals and N is the sample length.

Second, a Fourier transform is conducted to transform the covariance to cross spectrum, which is a function in the frequency domain, using

$$f_{xy}(\omega) = \frac{1}{N}\sum_{k=1}^{N} C_{xy}(i)\, e^{-2\pi i\omega k} \tag{2}$$

where ω is angular frequency.

The cross spectrum also can be written in complex form, with a real part $\underline{Real\_spectrum}$ and an imaginary part $\underline{Image\_spectrum.}$ (Brillinger, 2001).

$$f_{xy}(\omega) = Realpectrum_{xy}(\omega) - \ Image\_spectrum_{xy}(\omega)\mathrm{I} \tag{3}$$

With this complex form, the phase angle can be derived as

$$\text{Phase Angle}(\omega) = Tan^{-1}\left(\frac{Image\_spectrum_{xy}(\omega)}{Real\_spectrum_{xy}(\omega)}\right) \tag{4}$$

The square of the <u>cross spectrum power</u> is calculated using the $Real\_spectrum$ and $Image\_spectrum$, as

$$P_{xy}(\omega)^2 = \ Real\_spectrum_{xy}(\omega)\ ^2 + Image\_spectrum_{xy}(\omega)\ ^2 \tag{5}$$

Subsequently, the <u>square of coherence</u>

$$K_{xy}(\omega)^2 = \frac{\left|P_{xy}(\omega)\right|^2}{|P_{xx}(\omega)|^2\left|P_{yy}(\omega)\right|^2} \tag{6}$$

indicates that coherence is a normalization of cross power spectrum $P_{xy}(m)$ versus two autospectra $P_{xx}(\omega)$ and $P_{yy}(\omega)$. The normalization process can reduce bias caused by coupling effects of power spectra between two signals (Brillinger, 2001).

<u>2.1.1.2 Wavelet Coherence</u>

Wavelet coherence is an enhanced version of coherence analysis. Wavelet coherence is a localized correlation function in the frequency domain, as opposed to coherence analysis that is a correlation function in the time domain. Both approaches can identify the level of coherence between two signals. Relative to coherence analysis, however, wavelet coherence does not assume stationary characteristics of the data. The joint probability distribution of wavelet coherence does not change when data are shifted in time or space.

White noise is a typical stationary process, whereas human movement is often a non-stationary process. For example, any jerk in motion is characterized by non-stationary properties, and responses to sudden perturbations are also often non-stationary processes. Another important advantage of the wavelet coherence approach is that it can yield richer information, since output is generated in both the frequency and time domains. These two benefits suggest that wavelet coherence is a promising tool for processing non-stationary signals, the characteristics of which changes over time.

Wavelet coherence as a theory has not been extensively used until it was proposed for processing geophysical time series data (Grinsted et al., 2004). These authors sought to quantify the coherence between Arctic Oscillation index and the Baltic maximum sea ice extent record. Using the wavelet coherence method, they found that there is a significant coherence between the two phenomena, with an anti-phase relationship.

The first step of wavelet coherence is to perform the continuous wavelet transform and obtained wavelets, using

$$W_n^x(s) = \sqrt{\frac{\delta T}{s}} \sum_{n'=1}^{N} x_n \, \varphi\left[(n' - n)\frac{\delta T}{s}\right] \tag{7}$$

where x_n is signal with length N. The $\varphi\left[(n' - n)\frac{\delta T}{s}\right]$ term is a wavelet function, and herein the Morlet function is used. S is scale, and $\sqrt{\frac{\delta T}{s}}$ is used for normalizing the wavelet function.

The second step is to calculate the <u>cross wavelet transform</u> of the wavelets, which yields the wavelet spectrum power and phase angles.

$$W_n^{xy} = W_n^x \left(W_n^y\right)^* \tag{8}$$

where * represents complex conjugation

Similar to the cross spectrum in coherence analysis, the cross wavelet transform is used to calculate the common power of two signals written in a complex form.

$$W_n^{xy} = \mathrm{W}_1 + \mathrm{W}_2\mathrm{i} \tag{9}$$

From this, the phase angle = $Tan^{-1}\left(\frac{\mathrm{W}_2}{\mathrm{W}_1}\right)$ and the cross wavelet power $\mathrm{W}^{\mathrm{xy}^2} = W_1^2 + W_2^2$ can be calculated.

The third step is to calculate <u>wavelet coherence</u>, using

$$\mathrm{C}_\mathrm{n}^2(\mathrm{s}) = \frac{\left|\mathrm{S}\left(\mathrm{s}^{-1}\mathrm{W}_\mathrm{n}^{\mathrm{xy}}(\mathrm{s})\right)\right|^2}{\mathrm{S}(\mathrm{s}^{-1}|\mathrm{W}_\mathrm{n}^\mathrm{x}(\mathrm{s})|^2)\,.\;\mathrm{S}\left(\mathrm{s}^{-1}\left|\mathrm{W}_\mathrm{n}^\mathrm{y}(\mathrm{s})\right|^2\right)} \tag{10}$$

where S is a smoothing operator function (Liu, 1994). Usually, an exponential smoothing function is used as the smoothing operator function (Torrence and Compo, 1998, Grinsted et al., 2004).

As an important concept, the cone of influence (COI) can be used to detect edge effects of the wavelet coherence. Coherence outside of COI is considered as computational artifacts. This is due to the fact that the wavelet function itself has edge effects, as such also resulting in edge effects on wavelet coherence once convolved with input signals.

The significance test of wavelet coherence can be estimated through generating AR1 noise. Red noise is one type of AR1 noise, and which is often used for testing significance of wavelet coherence. Red noise is characterized by increasing power with decreasing frequency, and can be obtained through Monte Carlo simulation methods. The generated AR1 noise has similar power probability distribution as the investigated signals, such that a significance test against the AR1 noise data can also yield an approximate significance level for the investigated signals (Torrence and Compo, 1998).

2.1.2 Demonstration Examples

To demonstrate what the wavelet coherence method can achieve, two simulated signals are used. Both are based on a simple sinusoid or cosine wave function with 0.25Hz frequency, magnitude of 1, and which can be written as

$$Y = A \times Sin(\omega \times t) \tag{11}$$

where A = 1, $\omega = 2 \times \pi \times f = 0.5\pi$

For demonstration, different magnitudes of sine or cosine functions are also used. For example, the magnitude could be increased to 5.

$$Y_1 = 5 \times Sin(0.125\pi \times t) \tag{12}$$

The change of wavelet coherence with different magnitude or phase angle can be observed when the magnitude of the sine function changes.

Phase-shifted sine or cosine wave functions are also used, such as shifting the phase $\pm\pi/4$ which gives $Y_2 = 1\times Sin(0.125\pi\times t \pm \pi/4)$. By shifting the phase of a sine or cosine wave function, a change in phase angles of the wavelet coherence is expected.

Normally distributed random numbers are generated with mean 0 and standard deviation 1. Random numbers can be generated through Matlab function ***randn***, which generates uniformly distributed random noise distributed as *N* (0,1). The aim of using random number inputs is to identify how random noise can affect the wavelet coherence and phase angles.

Brownian motion, also known as a diffusion equation, is also generated. It can be described by:

$$x^2 = 2Dt \tag{13}$$

where D is the diffusion coefficient, *t* is time, and *x* is displacement.

Brownian motion data are correlated random numbers, which can be used to test how input signal correlation affects wavelet coherence.

For all plots below, the level of coherence is denoted through color coding (see legend bar), and with levels in the range of (0, 1). The phase angle is represented by arrow direction, with right = in-phase, left = anti-phase, up = leading 90 degrees, and down = lagging 90 degrees. The cone of influence (COI) is also demonstrated, which in all cases shows no wavelet coherence

Scenario one uses two identical input signals with different magnitudes (Figure 2.1). Varying the magnitude of the sine function does not induce any change in the wavelet coherence output. The phase angle is zero, which means the two input signals retain ain-phase relationship. The coherence is significant across all frequency bands.

$$Y_1 = 1 \times \mathrm{Sin}(0.125\pi \times t) \quad Y_1 = 5 \times \mathrm{Sin}(0.125\pi \times t) \quad Y_1 = 15 \times \mathrm{Sin}(0.125\pi \times t)$$
$$Y_2 = 1 \times \mathrm{Sin}(0.125\pi \times t) \quad Y_2 = 5 \times \mathrm{Sin}(0.125\pi \times t) \quad Y_2 = 5 \times \mathrm{Sin}(0.125\pi \times t)$$

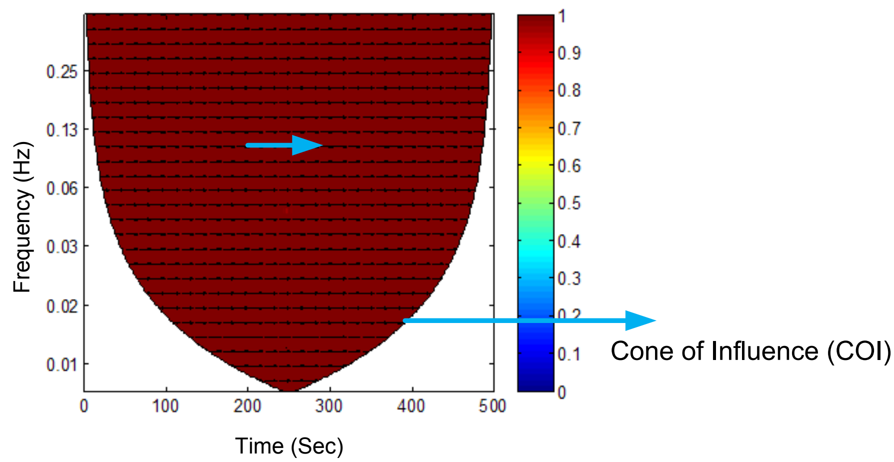

Figure 2.1: Wavelet coherence of different magnitude of Sine functions.

Scenario two uses two sine functions with different phases (Figure 2.2). Both advanced and delayed phases $\pm\pi/8$, $\pm\pi/4$, $\pm\pi/2$ are used. With phase-shifted signals, the wavelet coherence phase angles can be observed. The coherence power mainly exists around 0.125Hz with phase angle of $\pm\pi/8$, $\pm\pi/4$, $\pm\pi/2$. An in-phase angle also appears in the low frequency bands, which seems to be not reflecting the input signal phase relationship.

It indicates that the coherence phase angle around the main frequency band (such as 0.125 Hz) is more accurate for presenting the input signal phase relationship.

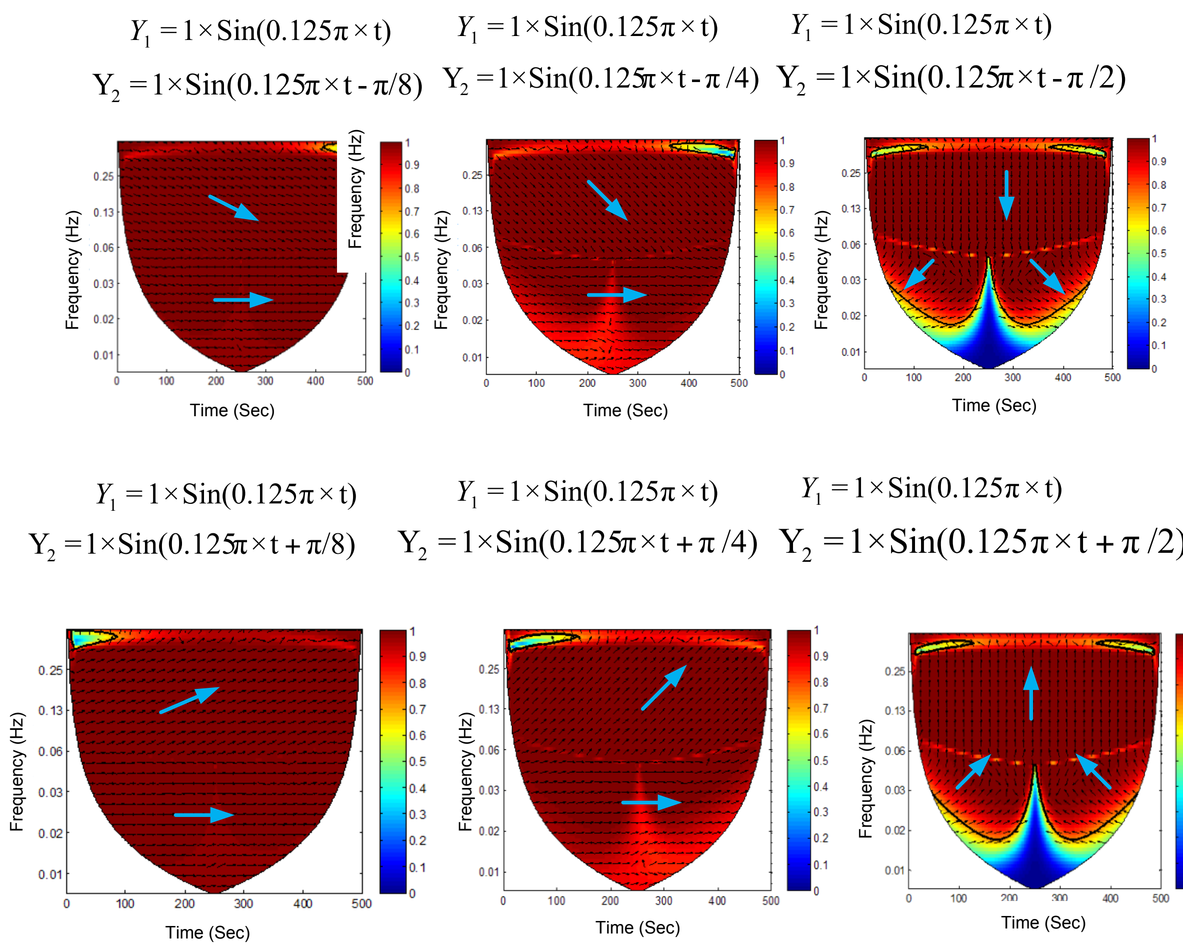

Figure 2.2: Wavelet coherence of different phase of Sine functions.

Scenario three uses a simple sine function with a fixed magnitude 1 and phase $-\frac{\pi}{8}$ along with one containing random noise (magnitude varying over 0.1 – 5) (Figure 2.3). With a small magnitude of random noise (0.1), there is still phase lock behavior. With an increase in random noise magnitude, though, the phase lock disappears, as does the significant wavelet coherence. Some non-phase-lock significant coherence still exists, however, even with random noise of high magnitude (15).

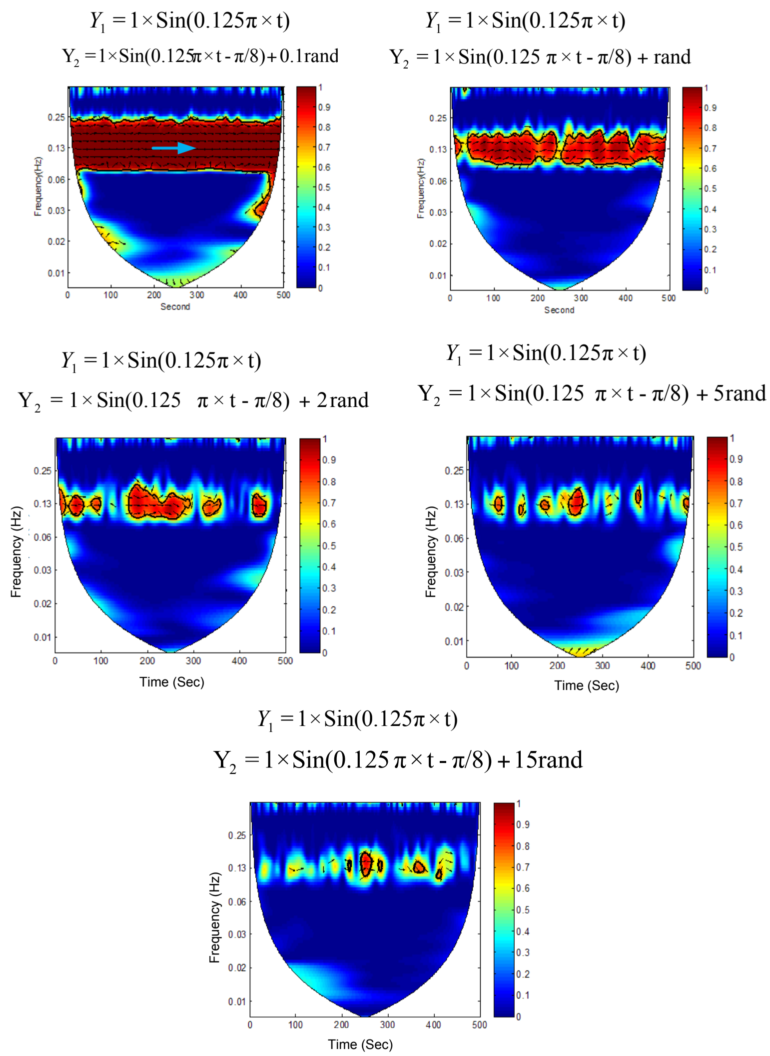

Figure 2.3: Wavelet coherence of two sine functions, with and without random noise.

Scenario four adds random noise to both simple sine functions (Figure 2.4). With low magnitude noise (0.1), the phase lock phenomenon exists. Once the noise magnitude increases, however, the phase lock gradually disappears. With a magnitude of 15, the phase lock is minimal, and the wavelet coherence then appears random.

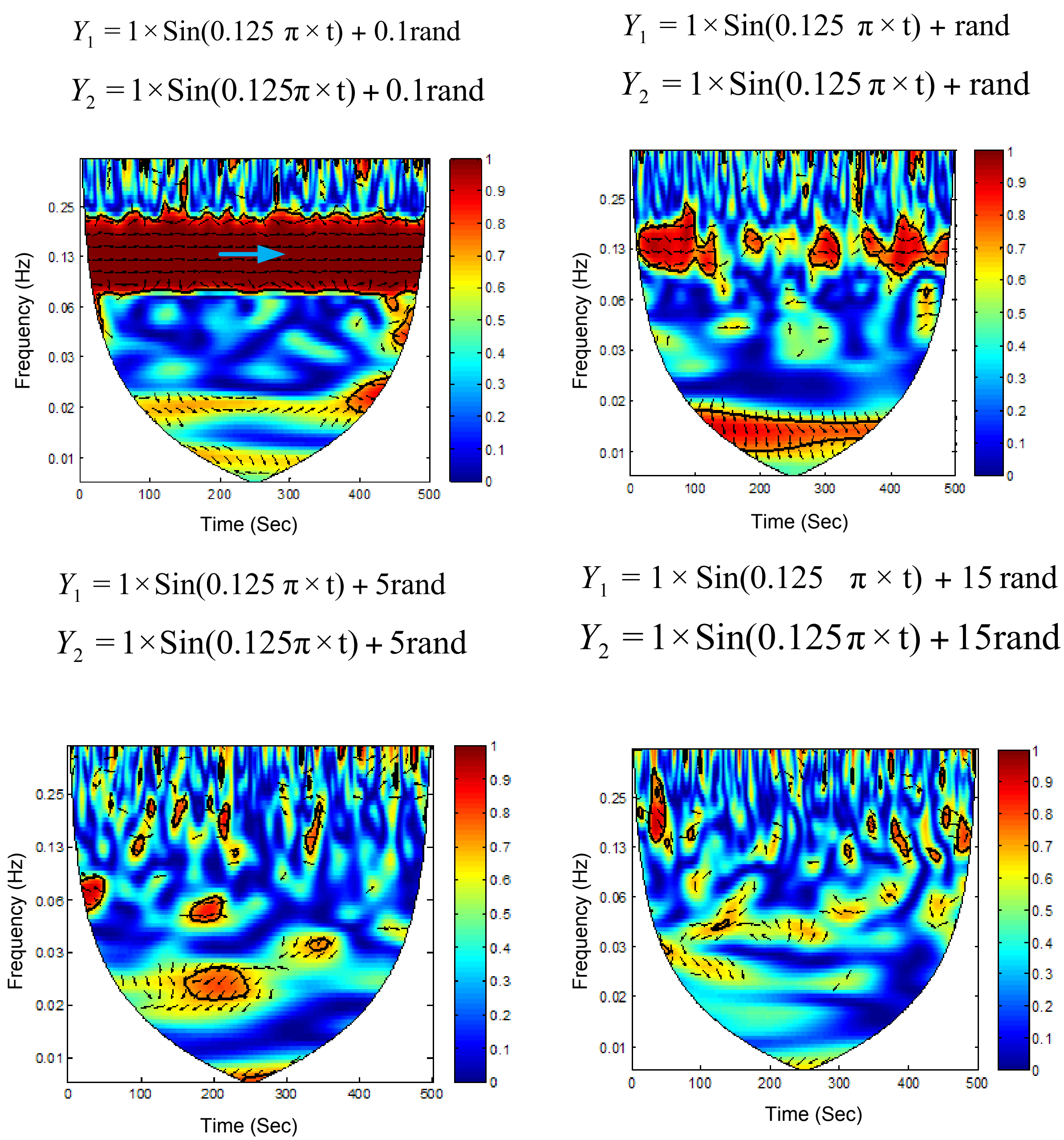

Figure 2.4: Wavelet coherence of two Sine functions with random noise.

Scenario five uses two signals consisting of random noise N(0, 1) with coefficient of 0.1 (Figure 2.5). In this case all wavelet coherence is just random and there is no phase lock phenomenon

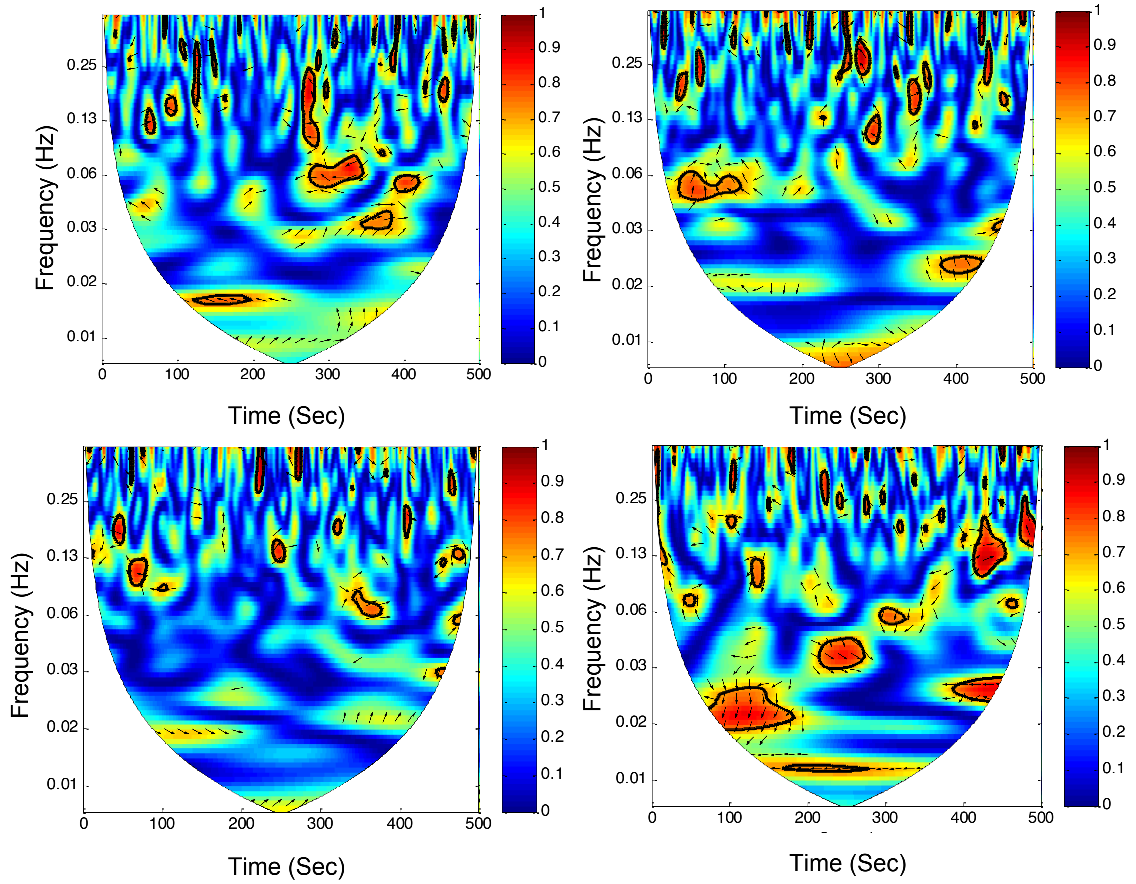

Figure 2.5: Wavelet coherence of two random noise signals.

Scenario six uses two Brownian motion signals (Figure 2.6). The Brownian motion is a correlated random noise signal. The wavelet coherence appears random without phase lock. This result is similar to the wavelet coherence from two random noise signals (above), indicating that correlated random noise does not differ from random noise in terms of coherence and phase angles.

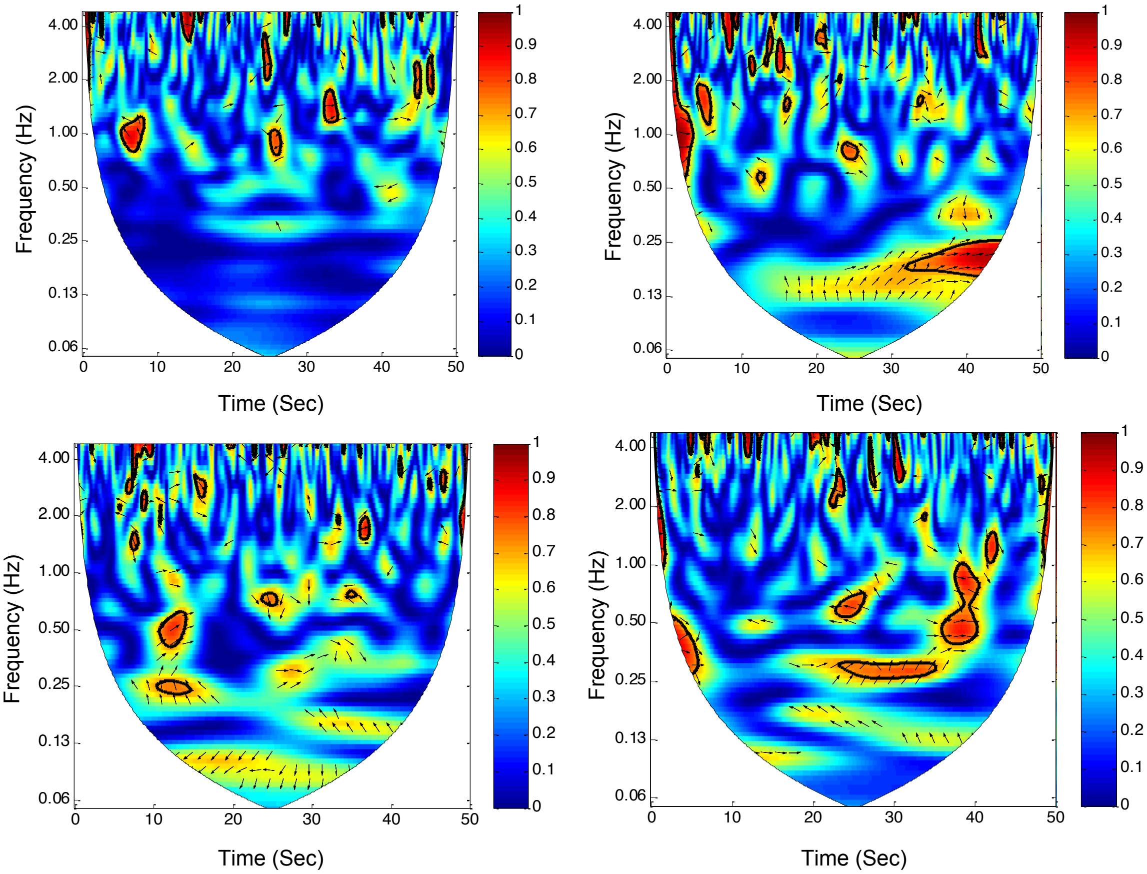

Figure 2.6: Wavelet coherence of two Brownian motion signals.

2.1.3 Summary and Conclusions of Wavelet Coherence Analysis

The examples above illustrate that wavelet coherence is able to detect phase lock behavior. For two identical signals, there are in-phase lock phenomena across all frequency bands. This is similar to Fourier transform based coherence, where coherence values are related to a range of frequencies instead of just one particular frequency. As such, care should be taken when choosing a specific frequency band for coherence analysis. A power spectrum plot of the investigated signal is helpful to understand the power distribution of the signal. By knowing the frequency - power relationship, the coherence related to significant power frequency band should be chosen for analysis. However, in many cases, a frequency band that does not contain significant power can still be interesting for coherence analysis, since it might have some information that is not necessarily connected with much power.

With wavelet coherence, phase shifts can be detected. For example, with input signal phase angle shifted by $\pm\pi/8$, $\pm\pi/4$, $\pm\pi/2$, the wavelet coherence phase angle is also shifted by $\pm\pi/8$, $\pm\pi/4$, $\pm\pi/2$. For two random signals and correlated random noise (Brownian motion), the phase relationship does not seem to be locked and appears random. It should be noted that the phase angles around the main power spectrum of the input signals represent the input signal phase angles at best.

Wavelet coherence is also able to detect the common power of two signals. With two identical input signals, the common power is time invariant. In terms of two random signals, the common power of two signals is distributed randomly across frequency bands through time. It is worth noting that the input signal frequency band(s) should be known in advance, so that focus is maintained on investigating the common power frequency bands, which is related to the input signal frequency band(s). Wavelet coherence can possibly bring some artifacts of common power between two signals due to cone of influence (COI) effects and such effect should be considered during analysis of wavelet coherence.

2.2 Uncontrolled Manifold

A manifold is a higher-dimensional analog of smooth curves and surfaces (Tu, 2010). The idea of a manifold was not originated from a single person, but rather was a collective contribution from many people. Bernhard Riemann is recognized as the main contributor, who initially proposed the idea of differential geometry on a higher dimensional space, the original concept of the manifold (Tu, 2010). There are several active research fields involving manifolds, topological manifolds, analytic manifolds, and complex manifolds among others.

The idea of an uncontrolled manifold was inspired from studies of robotic control. In such control, the robot hand is controlled through a series of links (namely the lower arm, the upper arm, and the body). Depending on the complexity of the robotic system there could be from a few to many links used to control the hand and manipulator. The hand and manipulator is also called an end-effector, and its motion is dependent on the robot link kinematics, as well as their interactions. The link system is often redundant, which means a certain end-effector position can be generated through different combinations of link motions. A specific name is given by robotic researchers and engineers when dealing with the problem of kinematic redundancy: *internal motion* also known as a self-motion manifold (Murray et al., 1994).

The end-effector position can be given as:

$$\boldsymbol{g}(\theta_{1\ldots n}) = \{\boldsymbol{c}_d\} \tag{14}$$

where g is the link kinematic function with multiple angles from 1 to n (the number of links), and c_d is the end-effector desired position. The link kinematic solution that satisfies this equation is also formulated as a self-motion manifold.

With the derivative of joint angle for each term in $\boldsymbol{g}$, the Jacobian matrix can be obtained. The right part of the end-effector position equation should be zero. As a result, the Jacobin matrix satisfies the following equation.

$$\boldsymbol{J}(\theta_{1\dots n})\dot{\boldsymbol{\theta}} = 0 \tag{15}$$

From equation 15, the joint motion falling into the self-motion manifold must have joint velocities embedded in the null space of the manipulator Jacobian matrix.

The self-motion manifold was termed the uncontrolled manifold by Scholz and Schoner (1999), who applied the uncontrolled manifold to study body coordination during the sit to stand motion (Scholz and Schöner, 1999). In their study, it was hypothesized (and found) that the center of mass position is one possible end-effector during the sit to stand motion. The self-motion was renamed as uncontrolled, to reflect the fact that motion within the uncontrolled subspace leaves the end-effector position unaffected. Conversely, motion orthogonal to this subspace can affect the end-effector position, leading to a deviation from its normal positions.

As from robotic studies, any motion that does not alter the end-effector position falls into the null space of the Jacobian matrix.

$$\boldsymbol{J}(\theta_{1\ldots n})\boldsymbol{\varepsilon} = 0 \qquad (16)$$

Where ε is the null space of the Jacobian matrix**.** By comparison with Equation 15, ε is actually equal to $\dot{\theta}$.

Scholz and Schöner (1999) also extended the original self-motion manifold in the following way. They proposed the parallel to null space component, which can be calculated as:

$$\boldsymbol{P}_{null} = \sum_{i=1}^{n} \boldsymbol{\varepsilon_i}\,{}_{\boldsymbol{\theta_{dev-mean}}} \qquad (17)$$

where $\boldsymbol{P}_{null}$ is the parallel component to the null space of Jacobian matrix, and $\theta_{dev-mean}$ is the joint angle that is deviated from the mean joint angle, i.e. $\boldsymbol{\theta_{dev-mean}} = \boldsymbol{\theta} - \overline{\boldsymbol{\theta}}$.

Similarly, the orthogonal component can be obtained:

$$\boldsymbol{O}_{null} = \boldsymbol{\theta_{dev-mean}} - \boldsymbol{P}_{null} \qquad (18)$$

According to Scholz and Schöner (1999), the final uncontrolled manifold is equal to the normalized parallel component divided by the orthogonal component:

$$U = \frac{Norm(P_{null})}{Norm(\boldsymbol{O}_{null})} \qquad (19)$$

where U is the uncontrolled manifold ratio and Norm is the normalization process of the parallel and orthogonal components to the null space of the Jacobian matrix.

From this process, it is clear that Scholz and Schöner (1999) did not make significant modifications to the original idea of the self-motion manifold. Rather, they introduced two new terms, the parallel component to the null space and the orthogonal component to the null space, using relatively straightforward mathematical forms. Subsequently, they calculated the ratio between the parallel and orthogonal components and called it the uncontrolled manifold ratio. This seemingly simple process has actually transformed the null space, a vector of values, into a single value (a ratio). Such a transformation makes it more straightforward to interpret the self-motion manifold and uncontrolled manifold, by just using one single ratio value. If the ratio is > 1, it means the parallel component is larger than the orthogonal component and indicates that the end-effector is the control variable of the joint coordination. A control variable is a mathematical and control terminology, which in physical terms can be interpreted as the goal of a certain motion. However, if the ratio is < 1, the orthogonal component is bigger than the parallel component. In this case, joint motion then tries to push the end-effector position out of its desired trajectory, and we thus can conclude that the end-effector is not the control variable of the joint coordination.

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Chapter 3 Effects of Localized Muscle Fatigue and Aging on Two-Joint and Whole-body coordination during Upright Stance

Abstract

In this study, two-joint and whole-body kinematic coordination were examined during quiet upright stance, the effects of aging, gender, and localized muscle fatigue (LMF) of the ankle were also contrasted. Wavelet coherence and uncontrolled manifold (UCM) methods were used to quantify two-joint coordination and whole-body coordination, respectively. Results confirmed that ankle-knee, ankle-trunk, and ankle-head wavelet coherence were intermittent rather than continuous. In terms of age and gender differences, older adults had larger intermittent time intervals, which were attributed to reduced joint coordination, while females showed larger intermittent in-phase to anti-phase ratios, which was associated with their more erect posture. Following ankle LMF, the ankle-head coordination time intervals among older adults decreased, in contrast to an increase observed among the younger group, and indicating more adverse effects of ankle LMF among the latter. In contrast to the whole-body center of mass (COM) as a potential control variable, the head and shoulder appeared to be more important for maintaining upright stance, as evidenced by larger UCM values. In conclusion, this study further suggests that kinematic co-ordination among multiple body joints is achieved intermittently rather than continuously during quiet upright stance. In contrast to prior evidence, the current results suggest that control whole-body COM may not be a primary goal during quiet upright stance.

3.1 Introduction

Controlling the various motions associated with human upright stance is essential for maintaining balance. However, upright stance is an inherently unstable position due to gravitational torque and internal disturbances such as hemodynamic and neuromuscular noise. Moreover, complex sensorimotor controls are involved with postural sway, which rely on visual, vestibular and proprioceptive feedback (Peterka, 2000, Maurer et al., 2006), and feed forward mechanisms (Fitzpatrick et al., 1996). Several joints are involved in this process, and contribute to balance maintenance during quiet upright stance (Kiemel et al., 2008, Pinter et al., 2008), and these joints may be controlled synergistically. For example, Hsu and coworkers (2007) studied the interactions of the ankle and hip joints and identified a non-randomized pattern of phase coordination between them. Specifically, they proposed that both ankle- and hip-centered motor control pathways exist for maintaining upright stance (Hsu et al., 2007). An earlier study similarly suggested that motor variability is channeled through both the ankle and hip joints (Krishnamoorthy et al., 2005). However, unlike the mechanisms associated with the ankle, hip-centered motor control may utilize a task-level feedback controller that varies depending on the undertaken task (Welch and Ting, 2008).

In addition to ankle and hip coordination, other joints such as the trunk and neck also appear to be involved in the coordination process (Keshner, 2003, Patel et al., 2010). With respect to ankle-hip and truck-neck coordination, muscle synergies are activated, thereby controlling a network of muscles simultaneously, as opposed to controlling individual muscles separately, which could involve more control effort (Torres-Oviedo and Ting, 2010). However, this recent report contradicts an earlier study that suggested joint coordination occurs though independent control mechanisms, where each coordinated joint is independently controlled. Specifically, Alexandrov et al. (2005) demonstrated that ankle, knee, and hip movements are independently controlled through the same eigen-movements as in feed-forward control (which is governed by central control mechanisms).

In the context of upright stance, ankle and hip motion coordination appears to be in-phase for frequencies below 1.0Hz, and anti-phase for frequencies above 1.0Hz (Creath et al., 2005, Zhang et al., 2007). Moreover, in-phase ankle-hip coordination is thought to be regulated through neural control, whereas anti-phase coordination may occur as a direct result of the biomechanics of upright stance (Kiemel et al., 2008). Several researchers have shown that in-phase and anti-phase ankle-hip coordination is regulated through numerous ankle, leg, and trunk muscle groups (e.g., the soleus, rectus femoris, and erector spinae), and that there is strong coherence among these muscle groups (Saffer et al., 2008, Sasagawa et al., 2009). Saffer et al. (2008) also showed that the phase lag between muscle activity and joint angle increased from low frequency (<1 Hz) to higher frequency (>1.6 Hz), as a result of both feedback and feed forward control influences.

Emphasizing the importance of whole body coordination, Bernstein (1967) described how whole-body coordination was in fact essential for carrying out motor tasks. Later, studies investigating axial synergy—a phenomenon involving upper-body bending movements accompanied by an opposing motion of the lower body—also commented on the required contribution of multiple joints such as the ankle-hip and head-trunk, which can therefore be classified as whole-body coordination (Mouchnino et al., 1992, Massion et al., 1997, Alexandrov et al., 1998). Despite these essential findings, the precise mechanisms responsible for whole-body coordination are still under exploration. Several possible motor control goals have consequently been identified, including the following for quiet upright stance: 1) maintaining the projection of the center of mass position within the support surface (Reisman et al., 2002, Hsu et al., 2007), 2) limiting the center of pressure position within the support surface (Ferry et al., 2004), and 3) maintaining head and gaze stability (Ledebt and WienerVacher, 1996, DiFabio and Emasithi, 1997). Although this earlier work presents some clues as to how whole-body coordination is achieved, as yet there is no consensus. This divergence, therefore, provided one rationale for the current study.

Localized muscle fatigue (LMF) affects joint motion and postural coordination. At a basic level, LMF affects the local, peripheral, and central nervous systems, leading to compromised sensory input and motor output, and thereby affecting postural coordination (Paillard, 2012). For example, participants in one study displayed increased flexed knee motions following a prolonged hopping task (Bonnard et al., 1994). Similarly, jumping height, peak joint moment, and joint angular velocity were also reduced (Rodacki et al., 2001, Rodacki et al., 2002) as a result of prolonged hopping. Upper-extremity muscle fatigue can also compromise postural coordination. For example, following a repeated ball-throwing task, elbow and hand coordination was reduced compared to pre-fatigue levels (Forestier and Nougier, 1998). An adaptation in lifting postural coordination was also observed following a repetitive lifting task, wherein knee and hip motions decreased and lumbar spine flexion increased (Sparto et al., 1997). The effects of LMF on postural control also vary across different locations of the LMF. For example, lumbar LMF has been shown to lead to greater compromises in postural control as compared ankle LMF (Lin et al., 2009). Based on this evidence, another goal of this study was to investigate how joint coordination is affected by LMF.

With aging, human muscles exhibit decreases in mass, volume, and the number of fibers and motor units. These physiological changes lead to aging being typically associated with weaker muscles, more restricted mobility, and coordination decrements (Porter et al., 1995, Vandervoort, 2002). Aging is also associated with neuron loss in the brainstem and cerebellum, which usually results in a loss of neural-controlled joint coordination (Sjöbeck et al., 1999). These factors directly contribute to slower target acquisition times (Barry et al., 2005), compromised whole-body coordination (Hsu et al., 2012), over control of inter-joint coordination (Vernazza-Martin et al., 2008), and slower force generation capability (Barry et al., 2005). As such, it can be postulated that joint coordination among older adults is likely to be different (i.e., diminished) in comparison to younger adults, and which we therefore investigated in this study.

Despite extensive existing investigations of upright stance, several aspects related to joint coordination still remain unclear. For example, the time-dependent characteristics of two-joint coordination remain generally unknown, and there is disagreement on whether two-joint coordination is continuous or intermittent. Two important studies have demonstrated that upright stance control is continuous (Peterka, 2000, Peterka and Loughlin, 2004), wherein a continuous feedback of afferent sensory information triggers essential muscle contractions and generates correct joint torques to maintain balance. In contrast, researchers have recently argued that intermittent control is the underlying control model for maintaining upright stance (Loram and Lakie, 2002, Loram et al., 2004, Bottaro et al., 2005, Gawthrop et al., 2009, Vieira et al., 2012). The latter suggests that predicative control signals burst intermittently for controlling upright stance. Therefore, identifying coordination patterns is essential for clarifying how coordination is executed. An additional issue of importance is that, as indicated above, the likelihood of whole-body coordination and related control goal both still remain insufficiently understood. Therefore, this study was designed to explore the whole-body coordination control goal, with an aim of better understanding the relationship between joint coordination and whole-body coordination.

For this study, we evaluated several aspects of kinematic coordination, both before and after localized muscle fatigue, and among both young and older adults: 1) ankle and knee coordination, 2) ankle and trunk coordination, 3) ankle and head coordination, and 4) whole-body coordination. For quantifying two-joint coordination, we utilized wavelet coherence (Chapter 2). This method has been used in some studies for investigating spatiotemporal coordination characteristics involving two input signals (Kumar and FoufoulaGeorgiou, 1997, Grinsted et al., 2004). Similar to other studies evaluating whole-body coordination (Scholz and Schöner, 1999, Krishnamoorthy et al., 2003, Krishnamoorthy et al., 2005, Hsu et al., 2007), the uncontrolled manifold (UCM) was also used here, since it can verify whether the variables of interest represent essential control variables for whole-body coordination. Specifically, UCM was employed to ascertain whether the shoulder, head, and whole-body COM kinematics represent essential underlying control variables for maintaining upright stance.

3.2 Methods

3.2.1 Experimental Setup

Data from an earlier experiment (Lin et al., 2009) were used herein. All participants read and signed an informed consent form approved by the Virginia Tech Institutional Review Board. Participants had no self-reported injuries, illnesses, musculoskeletal disorders, or falls in the year prior to the experiment. A total of 32 participants (gender balanced) completed the study, half of whom were young adults (18-25 years) and half older adults (55-65) years. Repeated trials of quiet upright stance were done both before and following ankle plantar-flexor muscle fatigue, and involved standing without shoes as still as possible, with their feet together, arms by their sides, head upright, and eyes closed. Each trial lasted 75 seconds, with at least one minute between two consecutive trials. Participants completed 3 trials under pre-fatigue conditions, and 11 trials under post-fatigue conditions; here, the first three post-fatigue trials were used.

Whole-body kinematics were estimated using surface markers, using a 6-camera system (Vicon 460, Lake Forest, CA, USA), with a sample rate of 20 Hz. In total, 18 reflective markers were attached on the body, distributed bilaterally (as relevant) at the temple, chin, acromion, sternum, elbow joint center, wrist joint center, anterior superior Iliac spine, knee joint center, lateral malleolus, and 5th metatarsal. Raw kinematics signals were low-pass filtered (Butterworth, 8 Hz cut-off frequency, 4th order, zero lag, bi-directional) and transformed to obtain body position time series; the first 10s and the last 5s of data were removed to eliminate any potential transition effects.

3.2.2 Kinematics

Following filtering and windowing of marker data, three dimensional marker locations were available (+X = front; +Y = right; and +Z = up). Among these, the X-Z and Y-Z axes defined the sagittal plane (SP) and frontal plane (FP), respectively. Joint angles were defined as between a given body segment and the horizontal (Figure 3.1), and were determined using an approach similar to a previous report (Black et al., 2007a). Specifically, foot, ankle, knee, hip, trunk, shoulder, and head, upper arm, and lower arm joint angles were calculated. For each joint, the distal d1(x.y,z) and proximal positions d2(x,y,z) of the respective body segment were used, along with the equations below, to estimate angles in both the sagittal and frontal planes. Angles for bilateral segments (i.e., the extremities) were averaged.

$$\theta_{FPA} = \tan2^{-1}\left(\frac{d1[z]-d2[z]}{d1[x]-d2[x]}\right) \qquad (1)$$

$$\theta_{SPA} = \tan2^{-1}\left(\frac{d1[z]-d2[z]}{d1[y]-d2[y]}\right) \qquad (2)$$

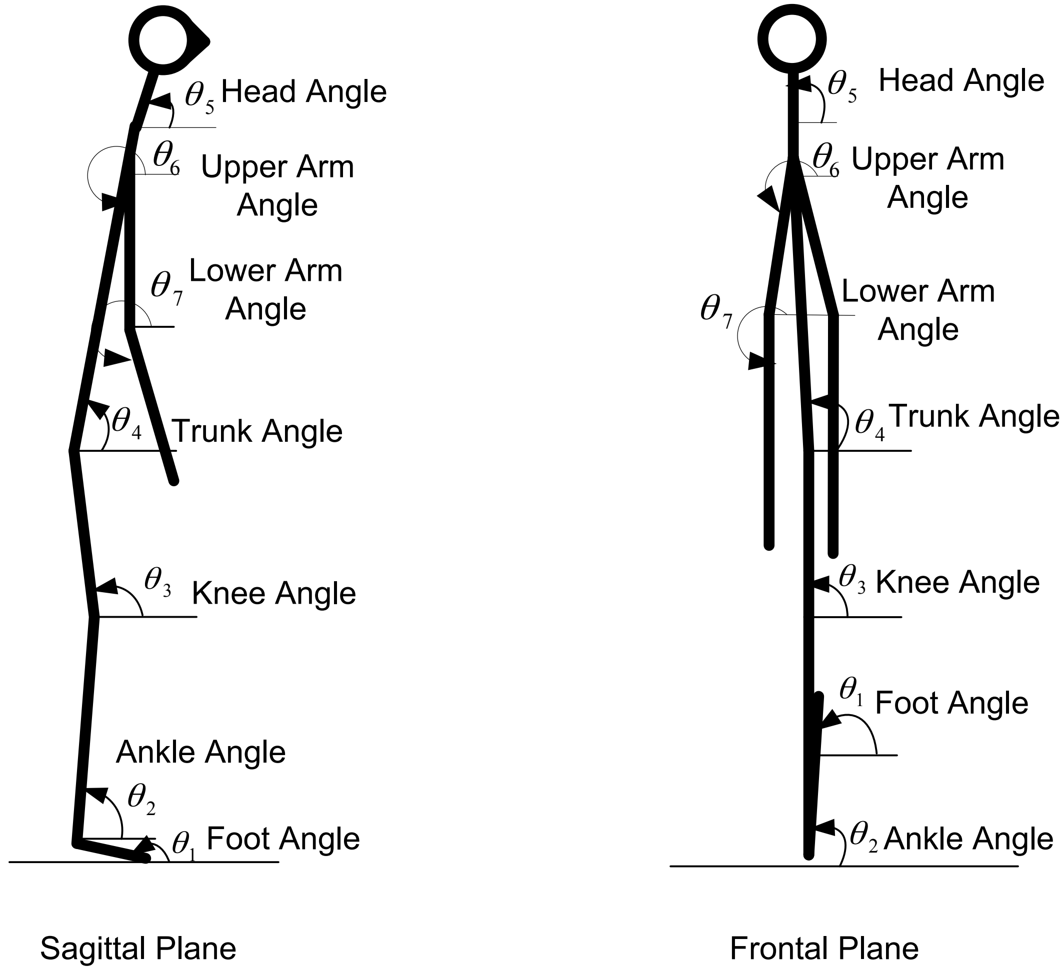

Figure 3.1: Joint angles in sagittal and frontal planes during upright stance.

3.2.3 Wavelet Coherence

A continuous wavelet transform was applied to the ankle, trunk, and head joint angle time series, and subsequently, a cross wavelet transform was used for selected pairs of angles. Specifically, this was done for the following pairs of angle: ankle-knee (AK), ankle-trunk (AT), and ankle-head (AH). The cross wavelet transform was:

$$W^{xy} = W_n^{\theta_1} \left(W_n^{\theta_2}\right)^* \tag{3}$$

where θ_1 and θ_2 represent two joint angles, $W_n^{\theta_1}$ and $W_n^{\theta_2}$ are the continuous wavelet transforms of these joint angles at time n, , and * is the complex conjugate. Based on the cross wavelet transform, wavelet coherence was calculated using the method introduced by Grinsted et al. (2004a). Because prior research has indicated that the ankle plays an essential role in upright stance (Winter et al., 2001), the ankle angle was chosen as one of the two input signals when calculating wavelet coherence. For this study, coherence was determined between sagittal plane and frontal plane ankle joint angles and those of the trunk, knee, and head. These latter angles were selected because the trunk, knee, and head all are involved in maintaining postural stability (Keshner, 2003, Creath et al., 2005).

A representative example of the wavelet coherence derived between AK, AT, and AH angles is presented in Figure 3.2 for one trial. As shown in this figure, the wavelet coherence was intermittent from 1 to 4 Hz, which is represented by the intermittent incidences of wavelet coherence from 1 to 4 Hz. Below 2.5 Hz, there exists a relatively large cone of influence (COI), indicating the edge — artificial effects of wavelet coherence (Torrence and Compo, 1998), which were also seen in all trials. As such, the frequency band >2.5 Hz was chosen for all subsequent analyses.

Intervals between these intermittent instances of wavelet coherence were estimated, and denoted as a mean time interval (sec), which represents the typical time interval between the “islands” of intermittent wavelet coherence (Figure 3.2). Specifically, for each frequency f, the number of occurrences of non-significant wavelet coherence was identified, and denoted as $N_n(f)$, and the number of occurrences of significant wavelet coherence was denoted as $N_s(f)$. Thus, N(*f*) = $N_n(f) + N_s(f)$ represents the total length of the signal (number of samples). The percentage of time that non-significant wavelet coherence exists, relative to the total sample time (T) for the frequency *f*, is thus described as:

$$T_{non-significant}(f) = \frac{N_n(f)}{N(f)} \times T \tag{4}$$

By dividing the $T_{non-significant}(f)$ by the numbers of intervals between coherent islands, $N\_Interval(f)$, the mean time interval between these islands is equal to:

$$MTI(f) = \frac{T_{non-significant}(f)}{N\_Interval(f)} \tag{5}$$

Finally, the mean time interval across the frequency band of 2.5 – 4 Hz was obtained as the mean value of MTI at each discrete frequency in this range.

Values for the number of intervals between coherent islands, $N\_Interval(f)$, was calculated based on a simple empirical 5-point method, and which is illustrated in Figure 3.3. Consider a certain frequency *f* (in the range 2.5 ～ 4.0 Hz), and assume the wavelet coherence at time *t* (0 ～ 60 sec) is significant. Then, if the wavelet coherence at earlier times (*t* - 0.05 and *t* - 0.1 sec) are significant, but at subsequent times (*t* + 0.05 and *t* + 0.1 sec) are not significant, we than conclude that time *t* represents an occurrence of one time interval. This 5-point method assumes that the length of the significant coherence island is generally larger than 0.1 sec and that the length of the insignificant coherence island is also generally larger than 0.1 sec. We tested different island sizes (0.05, 0.1, and 0.15 sec, which correspond to 3, 5, and 7 points respectively) to compare their impacts on the value of $N\_Interval(f)$. No substantial differences in derived $N\_Interval(f)$ values were found using the three zone sizes in the frequency range of 2.5 – 4 Hz (Figure 3.4).

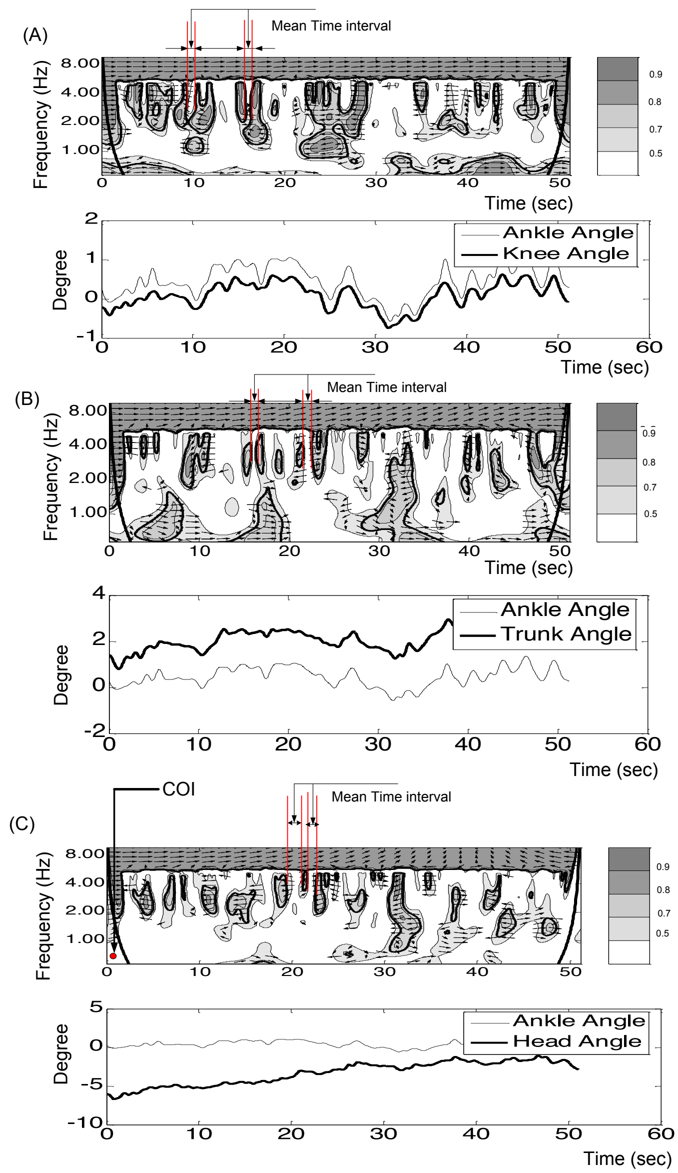

Figure 3.2: Illustrate of methods used to assess the intervals between “islands” of significant wavelet coherence. Samples are shown for coherence between (A) ankle-knee, (B) ankle-trunk, and (C) ankle-head frontal plane angles for a representative trial. For clarity, the illustrated joint angles are the actual joint angles minus $\frac{\pi}{2}$.

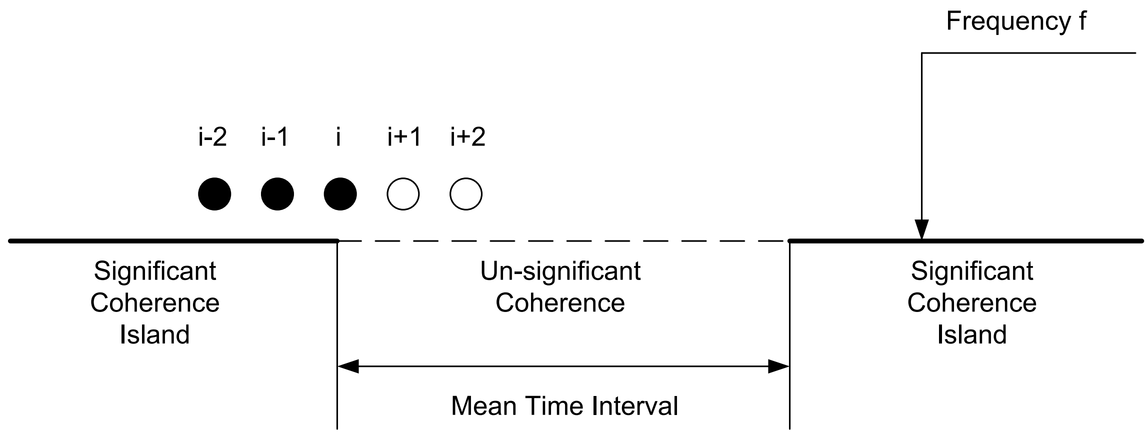

Figure 3.3: The 5-point method used to identify the occurrence of one time interval between two islands of significant coherence (at a given frequency f).

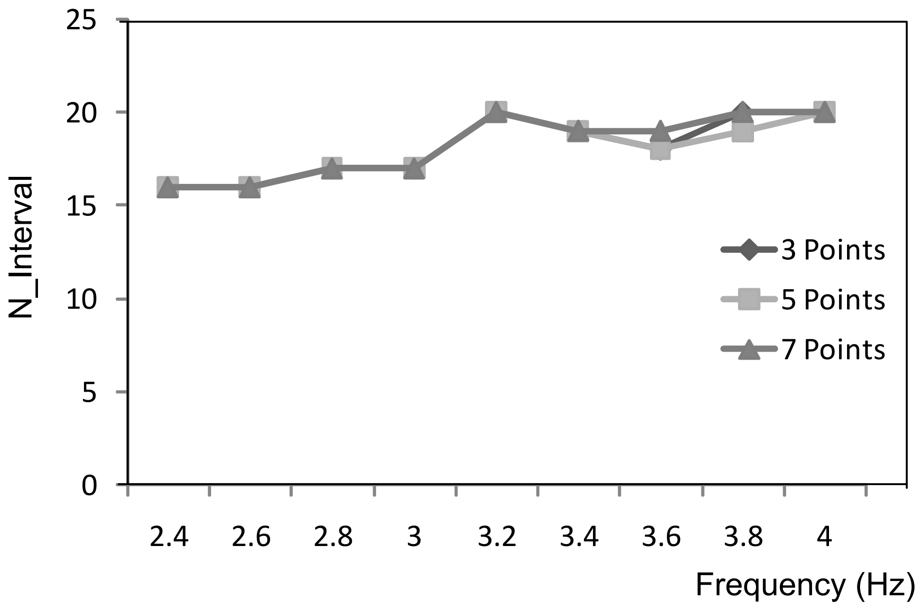

Figure 3.4: Derived values of N-interval based on 3, 5, and 7 point methods.

A sample of the wavelet coherence phase is shown in Figure 3.5. For AH and AT coordination in the frequency range of 1 – 4 Hz, there was a mix of in-phase and anti-phase relationships, i.e., in-phase and anti-phase relationships appear intermittently between 1 – 4 Hz. Due to the relatively large COI below 2.5 Hz, which was also identified for all trials, the frequency band of >2.5 Hz was chosen for phase analysis. To characterize the relationship between in-phase and anti-phase, the number of in-phase and anti-phase instances of significant wavelet coherence per frequency were computed over 2.5 – 4 Hz. For each frequency *f* between 2.5 – 4 Hz, the in-phase significant wavelet coherence was identified and is denoted as $IN(f)$, while the corresponding anti-phase significant wavelet coherence is $ANTI(f)$. The phase ratio then can be described as:

$$PH\_R(x) = \frac{IN(f)}{ANTI(f)} \tag{6}$$

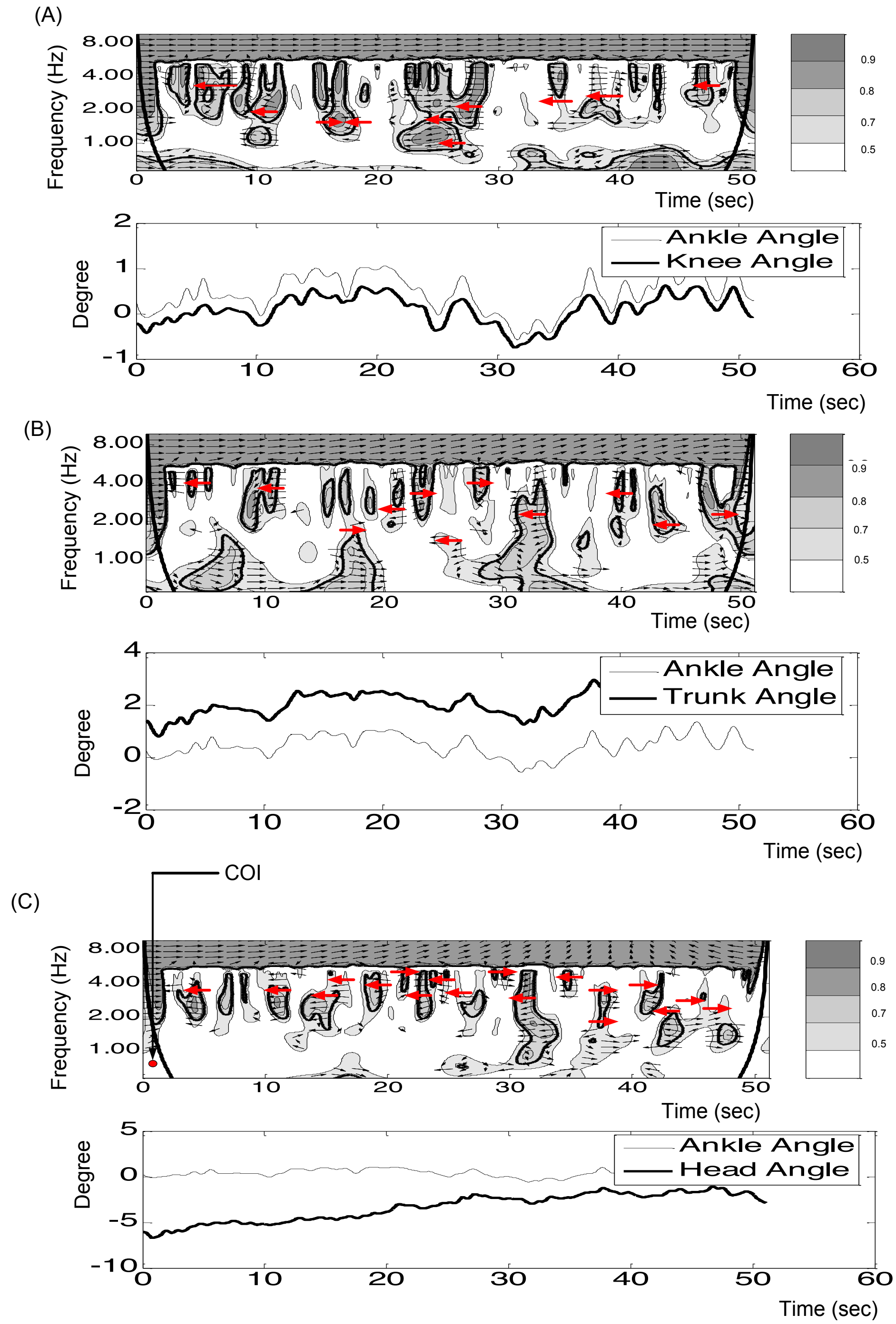

Figure 3.5: Phase results from wavelet coherence for (A) ankle-knee, (B) ankle-trunk, and (C) ankle-head frontal plane angles, including in-phase and anti-phase portions, are presented for a representative trial. For clarity, the illustrated joint angles are the actual joint angle minus $\frac{\pi}{2}$.

3.2.4 Uncontrolled Manifold (UCM) Analysis

As demonstrated by early research (e.g., DiFabio and Emasithi, 1997 and Hsu et al., 2007), the whole-body center of mass (COM) or the head could be primary control values used in whole body coordination. Thus, task variables selected for UCM analysis here were the whole body COM, shoulder, and head locations. Despite the fact that the shoulder has not been explicitly regarded as a control goal for whole body coordination, a strong biomechanical constraint is evident between the head and trunk/shoulder, and which have been also been shown moved in a coordinated fashion (Mouchnino et al., 1992, Bloomberg et al., 1997, Keshner, 2003). Additionally, the small relative movement between the shoulder and trunk implies that the shoulder could play an important role for ankle-trunk coordination, which is essential for quiet upright stance control (Hsu et al., 2007, Zhang et al., 2007). The COM, shoulder, and head locations were calculated based on the seven-segment model (Figure 3.1). Specific approaches used to calculate whole-body COM, shoulder, and head locations, in the ML and AP directions, are provided in Appendix A.

The processes for calculating the projection of variance on the UCM ($UCM_{||}$) and the projection of variance orthogonal to the UCM (UCM_T) are also shown in Appendix A. The projection of variance determines how much deviation occurs without altering the value of the task-level variable, while the ratio of $UCM_{||}/UCM_T$represents whether the goal of a specific task can be achieved (Black et al., 2007b). The uncontrolled manifold ratio is thus equal to the ratio: $UCM_{||}$ / UCM_T.

3.2.5 Statistical Analysis

From wavelet coherence analysis, dependent variables were the phase ratio and mean time interval for AK, AT, and AH angles in the sagittal and frontal planes. From UCM analysis, the head, shoulder, and COM uncontrolled manifold ratios were estimated separately in the AP and ML directions. The means of these dependent variables were obtained separately in the three pre-fatigue and three post-fatigue trials, and these means were used subsequently in statistical analyses. Effects of age and gender were evaluated for the pre-fatigue trials, using separate two-way ANOVAs. The differential effects of fatigue associated with age and gender were then analyzed using separate two-way ANCOVAs on change scores (post-fatigue – pre-fatigue) for each dependent variable, with independent variables of age and gender, and the pre-fatigue measure as a covariate. The level of significance for all tests was set at $p < 0.05$, and all statistical analyses were performed using JMP 9.02 (SAS Inc., Cary, NC, USA). Summary statistics are presented as means (SD).

3.3 Results

3.3.1 Statistical Results

A summary of statistical results is presented in Table 3.1, which shows the AK, AT, and AH wavelet coherence results, and mean values in the sagittal and frontal planes (SP and FP) for both pre- and post-fatigue conditions. The head, shoulder, and COM uncontrolled manifold results and mean values in the AP and ML directions are also shown for both pre- and post-fatigue conditions.

Table 3.1: Summary of statistical results. Significant effects are bolded (i.e., *p* values <0.05). Note that a significant effect in the final column indicates that the mean change score was ≠ 0. A=Age, G=Gender, SP = Sagittal Plane, FP = Frontal Plane, ML= Medio-Lateral, AP=Anterior-Posterior, AK= Ankle-Knee, AT=Ankle-Trunk, and AH=Ankle-Head.

		Pre Fatigue				Fatigue			
		Mean(std)	A	G	AXG	A	G	AXG	Mean(std)
Wavelet Coherence	AK Phase - SP	1.08(0.36)	0.13	0.45	0.49	0.37	0.47	0.99	1.09(0.39)
	AK Time - SP	1.00(0.18)	**0.043***	0.15	0.31	0.23	0.79	0.69	1.04(0.18)
	AT Phase - SP	1.01(0.42)	0.59	0.29	**0.029***	0.55	0.21	0.15	1.08(0.46)
	AT Time - SP	1.35(0.24)	0.94	0.37	0.63	0.69	0.40	0.45	1.39(0.21)
	AH Phase - SP	0.93(0.32)	0.37	0.83	0.48	0.54	0.81	0.91	0.86(0.29)
	AH Time - SP	1.45(0.26)	0.25	0.31	0.94	**0.017***	0.85	0.075	1.43(0.24)
	AK Phase - FP	0.89(0.27)	0.14	0.91	0.37	0.24	0.054	0.39	0.90(0.33)
	AK Time - FP	1.14(0.21)	0.92	0.18	0.29	0.18	0.42	0.42	1.13(0.23)
	AT Phase - FP	1.44(0.61)	0.13	**0.025***	0.19	0.39	0.94	0.14	1.37(0.64)
	AT Time - FP	1.27(0.19)	**0.026***	0.37	0.99	0.33	0.81	0.42	1.30(0.23)
	AH Phase - FP	0.96(0.38)	0.70	**0.012***	0.62	0.17	0.91	0.55	0.99(0.41)
	AH Time - FP	1.41(0.22)	0.96	0.38	0.65	0.15	0.39	0.77	1.39(0.25)
Untronctrolled Manifold	Head UCM - AP	1.05(0.13)	0.35	0.17	0.41	0.39	0.082	0.27	1.05(0.15)
	Shoulder UCM - AP	1.10(0.10)	**0.0097***	0.65	0.97	0.77	0.45	0.79	1.12(0.09)
	COM UCM - AP	0.634(0.0009)	0.28	0.09	0.95	0.60	0.61	0.69	**0.633(0.0009)***
	Head UCM - ML	1.00(0.10)	0.95	0.95	0.69	0.72	0.83	0.15	1.02(0.09)
	Shoulder UCM - ML	1.07(0.15)	0.36	**0.010***	0.92	0.86	0.10	0.066	1.10(0.10)
	COM UCM - ML	0.633(0.0009)	0.33	**0.047***	0.95	0.74	0.45	0.79	**0.634(0.0008)***

3.3.2 Pre-Fatigue

In comparing between groups, the AK mean time interval SP was larger among older vs. young adults (1.05(0.15) and 0.96(0.10)) sec, respectively). Similarly, the AT mean time interval FP was larger among older adults (1.32(0.14) sec) versus the young group (1.21(0.11) sec). In terms of gender differences, the AT phase ratio FP was larger among females than males (1.67(0.69) and 1.21(0.41), respectively) as was the AH phase ratio FP (1.09(0.27) and 0.84(0.24), respectively). There was an age x gender interaction effect on ankle-trunk phase. The phase ratio was smaller among older females in comparison to young females, but this ratio among males was similar across age groups. Additionally, the young female ankle-trunk phase ratio was larger than for young males, however the older female ankle-trunk phase ratio was smaller than that for older males. As such, there were divergent results with age between genders. Shoulder UCM AP was larger among young vs. older adults (1.15(0.09) and 1.08(0.09), respectively). In comparing genders, shoulder UCM ML was larger among females (1.13(0.09)) than males (1.03(0.14)), and COM UCM ML was smaller among females (0.634(0.001)) versus males (0.635(0.0004)).

3.3.3 Post-Fatigue

Post-fatigue changes were significant for both COM UCM ML and AP, and which increased and decreased, respectively, due to fatigue. Shoulder UCM ML also increased significantly post-fatigue. The effects of fatigue on ankle-head mean time interval differed between age groups (Figure 3.6). Specifically, this ratio increased post-fatigue in the young group, but decreased in the older group.

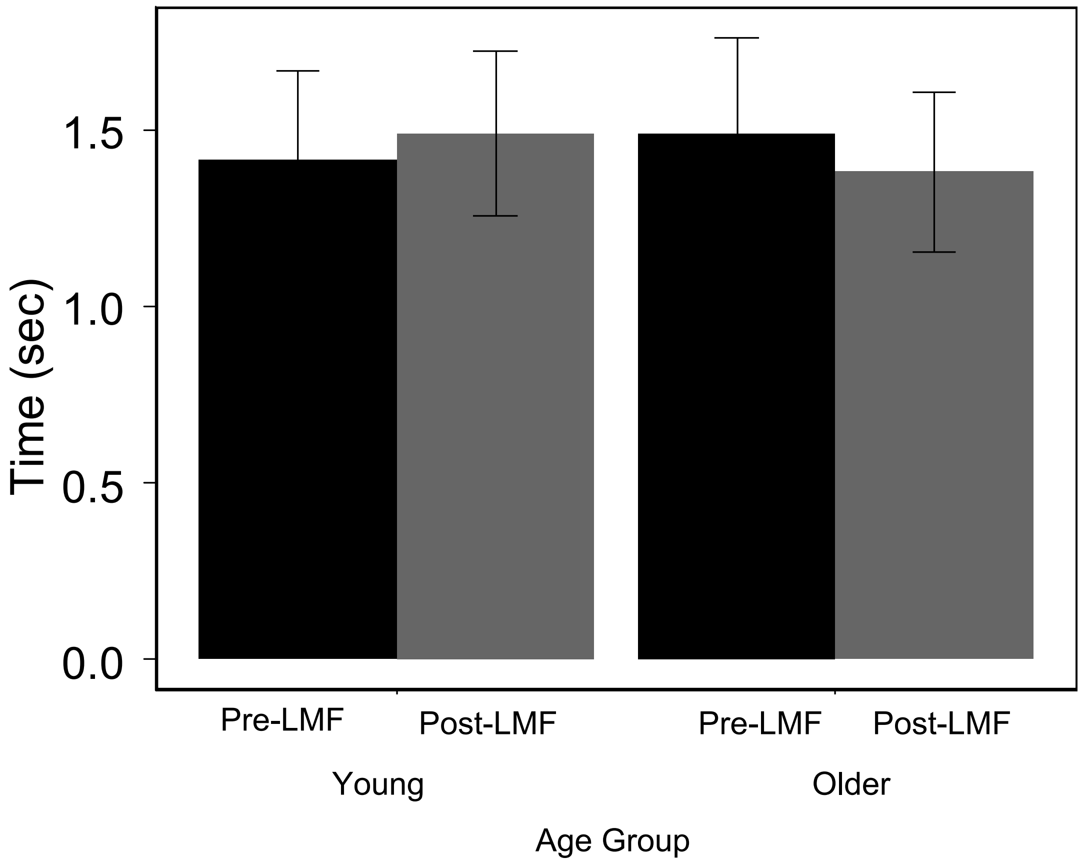

Figure 3.6: Ankle-head mean time interval in the sagittal plane for both pre and post ankle LMF.

3.4 Discussion

In this study two-joint coordination and whole-body coordination during upright stance were investigated, and the effects of ankle LMF, age, and gender on such coordination were examined. The work was based on several prior reports confirming that both two-joint and whole-body coordination are involved in upright stance maintenance (Krishnamoorthy et al., 2003, Creath et al., 2005, Krishnamoorthy et al., 2005, Hsu et al., 2007). Specifically, joint coordination patterns in the space-time domain and whole-body coordination goals were evaluated. Wavelet coherence and uncontrolled manifold methods were used, respectively, for assessing two-joint coordination and whole body coordination. Unlike conventional coherence methods, that are typically limited to the time domain only (Creath et al., 2005), wavelet coherence, which is a space-time analysis method, can provide richer time and frequency information for evaluating two-joint coordination, and essentially enabling two-joint coordination patterns to be readily identified. Whole-body coordination and related control goals were investigated using uncontrolled manifold techniques, which have been adopted in prior studies for analyzing whole-body coordination during quiet upright stance (Hsu et al., 2007, Wu et al., 2009). Significant wavelet coherence between ankle-knee, ankle-trunk, and ankle-head angles were identified in a wide range of frequency bands (<1.0 Hz to 8Hz). This finding is consistent with previous work showing substantial power content for upright stance motions from <1.0Hz (Mezzarane and Kohn, 2007) up to 8.0Hz (Krafczyk et al., 1999). For frequencies above 4.0Hz, although significant wavelet power spectra were identified, the magnitude of wavelet coherence was very small. Based on this limited coherence magnitude, the upper frequency limit for conducting mean time interval and phase ratio analyses was set to 4.0 Hz. Below 2.5 Hz, the cone of influence (COI) became larger, which means that only partial wavelet coherence could be used for calculating mean time interval and phase ratio. Consequently, 2.5 Hz was chosen as the lower limit for conducting wavelet coherence analyses.

Ankle-knee, ankle-trunk, and ankle-shoulder wavelet coherence results were intermittent rather than continuous (Figure 3.2). Specifically, the intermittent mean time interval was larger than 1 sec, ranging from 1.01 (0.42) sec for ankle-knee to 1.45(0.26) sec for ankle-head. Additionally, the mean time interval value decreased as the frequency increased from <1.0 Hz to 4.0 Hz. The magnitude of mean time interval also differed between ankle-knee, ankle-trunk, and ankle-head wavelet coherence. Among these, the ankle-knee mean time interval at 4.0 Hz was about 600 – 700 ms in the sagittal plane, and 500 – 600 ms in the frontal plane. Ankle-trunk and ankle-head mean time interval were about 900 – 1300ms in the sagittal and frontal planes. Such joint coordination mean time interval can be related back to intermittent muscle burst time-intervals, as pointed out by earlier research. Loram and Lakie (2002b) demonstrated that the intermittent soleus muscle burst time-interval was 300ms - 420ms during quiet upright stance. In light of the time delay between muscle burst and joint motion of 150ms indicated by Peterka and Loughlin (2004), this suggests that every muscle burst is associated with ankle-knee coordination, and every 2 or 3 muscle bursts correspond to ankle-trunk and ankle–head coordination at 4Hz. However, at even higher frequency, the association between muscle bursts and joint coordination changed, since the mean time interval decreases with increasing frequency, making it possible that one muscle burst corresponds to joint coordination for ankle-knee, ankle-trunk, and ankle-head.

The wavelet coherence results (Figure 3.3) are also consistent with previous research showing that both in-phase and anti-phase co-exist during upright stance (Creath et al., 2005, Zhang et al., 2007). Different from these earlier findings, however, which showed that in-phase coherence was typically below 1.0Hz while anti-phase coherence was above 1.0Hz, our results indicate that in-phase and anti-phase coherence may be distributed both below and above 1.0 Hz. This discrepancy could be due to the fact that wavelet coherence is able to deliver richer phase information, in contrast to Fourier transform techniques used in the prior studies noted. In other words, wavelet coherence was able to identify both frequency- and time-dependent phase variations, in contrast to Fourier transform methods, which only produce frequency-dependent phase variation information.

We also found that the phase ratio was close to one, and ranged from 0.89(0.27) for ankle-knee FP, to 1.44(0.73) for ankle-head phase FP. This magnitude was consistent for ankle-knee, ankle-trunk, and ankle-coordination from <1.0 Hz to 4.0 Hz. This result may not be entirely attributed to mechanical constraints between two joints, as indicated by Kiemel et al. (2008). Even though the ankle and trunk interact with mechanical constraints provided by the hip and the L5/S1 joints, there is minimum mechanical constraint between the ankle and head, which also has a phase ratio close to one. As such, some central control mechanisms such as eigen-movement control (Trivedi et al., 2010) and/or multiple muscle synergy (Ting and Macpherson, 2005) might also underlie the phase coherence patterns we identified here.

With respect to the effects of aging, older adults displayed a larger ankle-knee mean time interval in the sagittal plane, as well as a larger ankle-trunk mean time interval in the frontal plane. This indicates that adult ankle-trunk and ankle-head coordination among young individuals occurred more frequently, and suggests that young adults are more adept at compensating for the loss of postural stability through more frequent ankle-knee and ankle-trunk coordination activities. This conclusion is supported by prior research showing that older adults have larger time delays in the control loop, and demonstrate less upright stance stability (Qu et al., 2009, Davidson et al., 2011, Nishihori et al., 2011). In terms of gender-related outcomes, females had a larger ankle-trunk phase ratio in the frontal plane, and a larger ankle-head phase ratio in the frontal plane. The larger phase ratio among females indicates that more in-phase rather than anti-phase movement is embedded into ankle-trunk and ankle-head coordination. This outcome could be associated with postural differences between genders, wherein the females in this study tended to display more erect postures in comparison to males. Again, this finding also supports a prior study showing that female athletes preferred a more erect landing posture than their male counterparts by utilizing more hip and ankle range of motions (Decker et al., 2003). The more upright posture adopted by females might also be correlated to a larger UCM value, since shoulder UCM among females here was larger than among males.

Following ankle LMF, COM UCM increased in the ML direction and decreased in the AP direction, which indicates a directional dependency of fatigue effects. An increased shoulder UCM was also observed in the ML direction following fatigue, though this differences only approached significance ($p=0.075$). As a whole, these results suggest that the whole body control variable in the ML direction increased, but decreased in the AP direction, because of fatigue of the ankle plantar-flexors. Based on the definition of

UCM, the increased UCM value means that the projection of variance $UCM_{||}$ accounts for a larger percentage of the projection of variance orthogonal to the UCM_T, which also indicates that the goal of a specific task can be more likely achieved (Black et al., 2007b). As such, the decreased UCM in the AP direction is likely a direct detrimental effect from ankle fatigue, whereas the increased UCM in ML direction may be an adaptive effect possibly aimed to compensate for adverse effects of plantar-flexor fatigue.

Results also varied between age groups following ankle LMF, in that the mean time interval among older adults decreased, whereas it increased for young adults. Since the increase in mean time interval is related to a reduced level of joint coordination, the mean time interval increase for the young participants indicates that they are less able to maintain postural stability through joint coordination. This finding also indicates that ankle LMF created a less substantial impairment of coordination for older adults. One underlying factor related to this outcome could be the physiological changes in muscle fibers (in particular, the *increase* in fatigue-resistant Type I muscles) that occur with aging (Chan et al., 2000). A similar subsequent study also reported that muscular metabolic anaerobic pathways change with aging, which results in more fatigue resistance among older adults (Kent-Braun et al., 2002). Because localized ankle muscle fatigue affected rather distal joint coordination here (i.e., ankle-head coordination), it would appear that joint coordination involves *simultaneous* rather than *independent* muscular control. In other words, compromised performance in one joint due to LMF can affect two-joint coordination. It is important to note that ankle LMF here was only induced unilaterally. This might help to explain why there were no significant differences between pre-fatigue and post-fatigue ankle-trunk and ankle-knee wavelet coherence values, and these may be compromised by bilateral ankle LMF.

Another result of this investigation was that shoulder and head UCM values were >1 in both the AP and ML directions. In contrast, the COM UCM values were smaller <1 in both directions. As indicated by a prior study, if a task variable of UCM is <1, the task variability is structured in such a way as not to minimize end-effector variability (Scholz and Schöner, 1999). Therefore, our results indicate that whole body COM is unlikely to be a principal control variable during upright stance, supporting evidence from by previous research (Hsu et al., 2007, Zhang et al., 2007). Instead, our findings suggest that the shoulder and/or head are more likely to be relevant control variables in both the sagittal and frontal planes. This might be because the arms (both left and right), which contribute substantially to the center of mass position, play only a limited role in maintaining quiet upright stance. In contrast, the ankle and trunk, which are involved in manipulating the shoulder and head positions, play an important role in maintaining upright stance (Creath et al., 2005).

There are several limitations associated with this research that should be noted. First, and as noted above, only unilateral ankle LMF was involved. This protocol likely led to bilateral differences in ankle motion control and possible consequent changes in ankle-knee, ankle-trunk, and ankle-head coordination. Since the magnitude of ankle motion in quiet upright stance is small, however, it was assumed that the influences of unequal fatigue on joint coordination were limited. Second, and regarding the mean time interval equation (Eq. 5), the number of intermittent intervals was computed using an empirical local search method. It is possible that this method did not yield reliable results for some higher frequency intermittent intervals, due to the fact that rapid changes of the intervals are smaller than the limit of the local search area. However, only frequencies up to 4Hz were used for calculating mean time interval, and as such any computational errors associated with the empirical local search method should be small. Third, UCM magnitudes reported here differ from previous studies, which have given UCM values of 2 ～ 3 (Qu, 2012) , 1.5 ～ 5 (Wu et al., 2009), and 3 ～ >20 (Hsu et al., 2007). While care was taken to ensure the accuracy of COM, shoulder, and head positions and UCM calculations, it is still possible that the differences in UCM results are related to some inherent kinematics errors associated with the small magnitudes of sway motion.

3.5 Conclusions

In this research, ankle-knee, ankle-trunk, and ankle-head coordination, as well as whole-body coordination, were investigated during upright stance. The effects of ankle LMF, aging, and gender on coordination were also evaluated. Wavelet coherence and UCM methods were applied in this analysis of two-joint coordination and whole-body coordination. Results confirmed that ankle-knee, ankle-trunk, and ankle-head wavelet coherence are intermittent rather than continuous. The larger mean time interval associated with older adults suggests that this group was less able to use joint coordination for maintaining upright stance. It was also evident that (perhaps counter-intuitively) ankle LMF had more adverse effects on young adults, suggesting that they had a reduced ability to use joint coordination for balance maintenance. One gender-related finding was the larger phase ratio identified among females, and which was possibly related to their more erect posture. Based on UCM analysis, the head and shoulder appeared more likely to be significant control variables for maintaining upright stance, rather than whole body COM. It should be stressed, however, that this study focused exclusively on ankle-centered (ankle-knee, ankle-trunk, and ankle-head) body coordination. Future studies should investigate the role and/or influence of other two-joint coordination strategies. Additionally, inducing bilateral ankle LMF would also contribute to a more comprehensive understanding of the effects of ankle LMF on coordination.

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Appendix A

Wavelet Transform

$$W_n^x(s) = \sqrt{\frac{\delta T}{s}} \sum_{n'=1}^{N} x_{k=1:4}^{n'} \varphi\left[(n'-n)\frac{\delta T}{s}\right]$$

Where $x_{k=1:4}^{n}$ is the joint angle k at time n, where x_1^n is ankle angle, x_2^n is knee angle, x_3^n is trunk angle, and θ_4^n is neck angle. $\varphi\left[(n'-n)\frac{\delta T}{s}\right]$ is wavelet function, n is localized time index, $(n'-n)$ is time coefficient of the wavelet function φ, which is a Morlet function in our study, and s (0.1 – 17) is the scale of the wavelet transform. $\sqrt{\frac{\delta T}{s}}$ is used to normalize the wavelet function.

Center of Mass Determination

Center of Mass (COM) equation is:

$$\text{COM} = \sum_{i=1}^{7} \frac{m_i}{M} com_i$$

Where M is the whole body mass, and m_i and com_i are the mass and center of mass of each body segment.

The geometric model of whole body COM in the A-P direction, which is in sagittal plane is

$$COM_{A-P}(\theta_{foot-AP}, \theta_{shank-AP}, \ldots, \theta_{lower\ arm-AP}) = x_{MT5} + M_{foot} C_{foot} L_{foot} \cos(\theta_{foot-AP}) + M_{shank} (L_{foot} \cos(\theta_{foot-AP}) + C_{shank} L_{shank} \cos(\theta_{shank-AP})) + M_{thigh} (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + C_{thigh} L_{thigh} \cos(\theta_{thigh-AP})) + M_{trunk} (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + L_{thigh} \cos(\theta_{thigh-AP}) + C_{trunk} L_{trunk} \cos(\theta_{trunk-AP})) + M_{head}$$

$\left(L_{foot}\cos\left(\theta_{foot-AP}\right) + L_{shank}\cos(\theta_{shank-AP}) + \right.$

$\left. L_{thigh}\cos\left(\theta_{thigh-AP}\right) + L_{trunk}\cos(\theta_{trunk-AP}) + C_{head}L_{head}\cos(\theta_{head-AP}) \right) \quad +$

$M_{upper\ arm} \quad \left(L_{foot}\cos\left(\theta_{foot-AP}\right) + L_{shank}\cos(\theta_{shank-AP}) + L_{thigh}\cos\left(\theta_{thigh-AP}\right) + \right.$

$\left. L_{trunk}\cos(\theta_{trunk-AP}) + C_{upper\ arm}L_{upper\ arm}\cos\left(\theta_{upper\ arm-AP}\right) \right) \quad + \quad M_{lower\ arm}$

$\left(L_{foot}\cos\left(\theta_{foot-AP}\right) + L_{shank}\cos(\theta_{shank-AP}) + \right.$

$L_{thigh}\cos\left(\theta_{thigh-AP}\right) + L_{trunk}\cos(\theta_{trunk-AP}) + L_{upper\ arm}\cos\left(\theta_{upper\ arm-AP}\right) +$

$\left. C_{lower\ arm}L_{lower\ arm}\cos(\theta_{lower\ arm-AP}) \right)$

The geometric model of COM in the M-L direction, which is in frontal plane is:

$COM_{M-L}\left(\theta_{foot-ML}, \theta_{shank-ML}, \ldots, \theta_{lower\ arm-ML}\right) = y_{MT5} + M_{foot} \quad C_{foot}$

$L_{foot}\cos\left(\theta_{foot-ML}\right) + M_{shank} \left(L_{foot}\cos\left(\theta_{foot-ML}\right) + C_{shank}L_{shank}\cos(\theta_{shank-ML}) \right)$

$+ \quad M_{thigh}$

$\left(L_{foot}\cos\left(\theta_{foot-ML}\right) + L_{shank}\cos(\theta_{shank-ML}) + C_{thigh}L_{thigh}\cos\left(\theta_{thigh-ML}\right) \right) \quad +$

$M_{trunk} \quad \left(L_{foot}\cos\left(\theta_{foot-ML}\right) + L_{shank}\cos(\theta_{shank-ML}) + L_{thigh}\cos\left(\theta_{thigh-ML}\right) + \right.$

$\left. C_{trunk}L_{trunk}\cos(\theta_{trunk-ML}) \right) \quad + \quad M_{head}$

$\left(L_{foot}\cos\left(\theta_{foot-ML}\right) + L_{shank}\cos(\theta_{shank-ML}) + \right.$

$\left. L_{thigh}\cos\left(\theta_{thigh-ML}\right) + L_{trunk}\cos(\theta_{trunk-ML}) + C_{head}L_{head}\cos(\theta_{head-ML}) \right) \quad +$

$M_{upper\ arm} \quad \left(L_{foot}\cos\left(\theta_{foot-ML}\right) + L_{shank}\cos(\theta_{shank-ML}) + L_{thigh}\cos\left(\theta_{thigh-ML}\right) + \right.$

$\left. L_{trunk}\cos(\theta_{trunk-ML}) + C_{upper\ arm}L_{upper\ arm}\cos\left(\theta_{upper\ arm-ML}\right) \right) \quad + \quad M_{lower\ arm}$

$\left(L_{foot}\cos\left(\theta_{foot-ML}\right) + L_{shank}\cos(\theta_{shank-ML}) + \right.$

$L_{thigh}\cos\left(\theta_{thigh-ML}\right) + L_{trunk}\cos(\theta_{trunk-ML}) + L_{upper\ arm}\cos\left(\theta_{upper\ arm-ML}\right) -$

$C_{lower\ arm}L_{lower\ arm}\cos(\theta_{lower\ arm-ML})\ )$

The geometric model of whole body vertical COM position is:

$COM_Z\left(\theta_{foot-ML}, \theta_{shank-ML}, \dots, \theta_{lower\ arm-ML}\right) = Z_{MT5} + M_{foot}\ C_{foot}$

$L_{foot}\sin\left(\theta_{foot-ML}\right) + M_{shank}\left(\ L_{foot}\sin\left(\theta_{foot-ML}\right) + C_{shank}L_{shank}\sin(\theta_{shank-ML})\ \right) +$

$M_{thigh}\ \left(\ L_{foot}\sin\left(\theta_{foot-ML}\right) + L_{shank}\sin(\theta_{shank-ML}) + \ C_{thigh}L_{thigh}\sin\left(\theta_{thigh-ML}\right)\right)$

$+ \ M_{trunk}\ \ \left(\ L_{foot}\sin\left(\theta_{foot-ML}\right) + L_{shank}\sin(\theta_{shank-ML}) + \ L_{thigh}\sin\left(\theta_{thigh-ML}\right) +\right.$

$\left.C_{trunk}L_{trunk}\sin(\theta_{trunk-ML})\right) + M_{head}$

$\left(\ L_{foot}\sin\left(\theta_{foot-ML}\right) + L_{shank}\sin(\theta_{shank-ML}) +\right.$

$\left.L_{thigh}\sin\left(\theta_{thigh-ML}\right) + L_{trunk}\sin(\theta_{trunk-ML}) + C_{head}L_{head}\sin(\theta_{head-ML})\ \right) +$

$M_{upper\ arm}\ \ \left(\ L_{foot}\sin\left(\theta_{foot-ML}\right) + L_{shank}\sin(\theta_{shank-ML}) + \ L_{thigh}\sin\left(\theta_{thigh-ML}\right) +\right.$

$\left.L_{trunk}\sin(\theta_{trunk-ML}) + C_{upper\ arm}L_{upper\ arm}\sin\left(\theta_{upper\ arm-ML}\right)\ \right) + M_{lower\ arm}$

$\left(\ L_{foot}\sin\left(\theta_{foot-ML}\right) + L_{shank}\sin(\theta_{shank-ML}) +\right.$

$L_{thigh}\sin\left(\theta_{thigh-ML}\right) + L_{trunk}\sin(\theta_{trunk-ML}) + L_{upper\ arm}\sin\left(\theta_{upper\ arm-ML}\right) -$

$\left.C_{lower\ arm}L_{lower\ arm}\sin(\theta_{lower\ arm-ML})\ \right)$

The head positions in AP and ML directions:

$HEAD_{A-P}\left(\theta_{foot-AP}, \theta_{shank-AP}, \dots, \theta_{head\ -AP}\right) = y_{MT5} + L_{foot}\sin\left(\theta_{foot-AP}\right) +$

$L_{shank}\sin(\theta_{shank-AP}) + L_{thigh}\sin(\theta_{head-AP}) + L_{trunk}\sin(\theta_{trunk-AP}) +$

$L_{head}\sin(\theta_{head-AP})$

$HEAD_{M-L}(\theta_{foot-ML}, \theta_{shank-ML}, \dots, \theta_{head-ML}) = y_{MT5} + L_{foot}\cos(\theta_{foot-ML}) + L_{shank}\cos(\theta_{shank-ML}) + L_{thigh}\cos(\theta_{thigh-ML}) + L_{trunk}\cos(\theta_{trunk-ML}) + L_{head}\cos(\theta_{head-ML})$

The shoulder positions in AP and ML directions:

$SHOULDER_{A-P}(\theta_{foot-AP}, \theta_{shank-AP}, \dots, \theta_{trunk-AP}) = y_{MT5} + L_{foot}\sin(\theta_{foot-AP}) + L_{shank}\sin(\theta_{shank-AP}) + L_{thigh}\sin(\theta_{thigh-AP}) + L_{trunk}\sin(\theta_{trunk-AP})$

$SHOULDER_{M-L}(\theta_{foot-ML}, \theta_{shank-ML}, \dots, \theta_{trunk-ML}) = y_{MT5} + L_{foot}\cos(\theta_{foot-ML}) + L_{shank}\cos(\theta_{shank-ML}) + L_{thigh}\cos(\theta_{thigh-ML}) + L_{trunk}\cos(\theta_{trunk-ML})$

Uncontrolled Manifold Determination

An approximation of task variable with respect to the changes of joint angle is

$T - T^0 = J(\theta^0)\,(\theta - \theta^0)$

where T is task variable matrix – center of mass, head, and shoulder positions in sagittal and frontal planes. $J(\theta^0)$ is the Jacobian matrix, which is the first-order derivative of the center of mass, head, and shoulder positions with respect to joint angles. After obtaining the Jacobian matrix $J(\theta^0)$, the null space of the Jacobian matrix ϵ_i should satisfy the following equation.

$J(\theta^0)\,\epsilon_i = 0$

Any motion restricted to the null space of the manipulator Jacobian is called a self-motion or internal motion because it leads to no displacement of the end-erector (Martin, 2005).

Projection of the joint angle variance on the null space gives the parallel component of the uncontrolled manifold.

$\theta_{||}= \sum_{i=1}^{n} \varepsilon_i^T (\theta - \bar{\theta})\, \epsilon_i$

Where ε_i is Jacobian matrix, n is the total number of degrees of freedom in the system, θ is the joint angle, $\bar{\theta}$ is the average of joint angle, $\theta - \bar{\theta}$ constructs the deviation matrix, and ϵ_i is the basis vector of the null space. θ can be found in Figure 3.1.

The perpendicular component of the uncontrolled manifold is:

$\theta_T= (\theta - \bar{\theta}) - \theta_{||}$

The variability per degree of freedom within the uncontrolled manifold is

$UCM_{||}= \sqrt{(n-d)^{-1} N^{-1} \sum \theta_{||}^2}$

Where d is the task variable dimension and N is the number of trials.

The variability per degree of freedom perpendicular to the uncontrolled manifold is

$UCM_T= \sqrt{d^{-1} N^{-1} \sum \theta_T^2}$

Chapter 4 Development of an Intermittent Control Model for Evaluating Aging and Muscle Fatigue Effects on Human Upright Stance

Abstract

Previous experimental evidence has suggested that human upright stance is maintained or controlled intermittently, rather than continuously. Several intermittent postural control models have been proposed, however most of models are explorative, many of these are rather complex and do not provide estimates of physically-relevant parameters (e.g., joint kinematics/kinetics). Here, a sliding mode control model was developed for assessing quiet upright stance, and which was evaluated using data from both young and older participants and pre/post unilateral ankle plantar-flexor fatigue. The model was able to track, with reasonable accuracy, important experimental measures, including center of mass (COM) kinematics and ankle torque. However, COM angular acceleration and ankle torque tracking errors were larger among older adults. The model also enabled separate estimates of passive and active ankle torques, with respective contributions estimated to be 97% and 3% of the total ankle torque. While both age groups had similar passive ankle torques, older adults and males displayed larger active ankle torque. With respect to ankle fatigue, larger increases in active ankle torque were observed among males and older adults. In summary, a sliding mode intermittent controller was able to track upright stance kinematics and kinetics with reasonable accuracy, though tracking results did vary with age, gender, and fatigue. Such a model may have future utility in understanding human postural control.

4.1 Introduction

Human (bipedal) upright stance is known to be inherently unstable, and muscle force (joint torque) has to be generated to maintain this posture (Winter et al., 2003). Several recent studies have investigated the generation of muscle torque, which has been described hypothetically as a continuous process requiring feedback (Peterka, 2002, Maurer and Peterka, 2005, Van Der Kooij and De Vlugt, 2007, Vette et al., 2007). Diverse afferent inputs are involved in the control of upright stance—including those from foot plantar pressure sensors, semicircular canals, and ankle proprioceptors—with the central nervous system relying on a fusion of these sensory channels for upright stance control (Mergner et al., 2003). As such, it has been argued that an optimal and multisensory feedback mechanism exists to eliminate the inaccuracy caused by neural time delays that are on the order of ~100ms (van der Kooij et al., 2001). The reliance on feedback control signals may also be varied and re-weighted depending on any available sensory or spatial-orientation information (e.g., different visual and proprioceptive input) and environmental conditions (Peterka, 2002). A later study by Peterka and Loughlin (2004) described that closed-loop integration of sensory information and reweighting of diverse sensory input were both important for determining corrective torques.

Some researchers, however, have challenged the notion that feedback control is solely responsible for maintaining upright stance. For example, Winter et al. (2001) suggested that upright stance can be sufficiently maintained through passive ankle stiffness only, which essentially is a form of open-loop control. In other work, both open-loop and closed-loop control mechanisms have been inferred during upright stance (Collins and

Luca, 1993, Collins et al., 1996). (Fitzpatrick et al., 1996) also questioned the "feedback-only" control mechanism for maintaining upright stance and balance, suggesting instead that a feed forward mechanism must be involved to compensate for insufficient control torque and to maintain a bounded postural sway. Finally, contributions of the vestibular system and the proprioceptive system have been documented as likely sources of feed forward control signals (van der Kooij et al., 2001).

With advancements in measurement equipment, muscle activities can now be measured with higher resolution and precision. For example, researchers have reported that calf muscles are actively adjusted 2.6 times per second and 2.8 times per unidirectional sway of the body center of mass (COM) (Loram et al., 2005b). It is these small (30–300 μm) calf muscle movements that provide impulsive, ballistic regulation of COM movement (Loram et al., 2006, Onambele et al., 2006). Given that short-range passive ankle stiffness is larger than gravitational torque, these findings suggest that it is unnecessary for the ankle muscle-tendon complex to continuously activate to control human upright stance (Loram et al., 2007). A later study by Loram et al. (2009) observed that the ankle muscles contract during sway away from the equilibrium position, but lengthen when the body sways back to the equilibrium position. This intermittent muscle activation yields a "drop-catch" and "throw-catch" pattern of sway motion during upright stance, also known as a ballistic-like intermittent motion (Loram and Lakie, 2002b, Casadio et al., 2005).

Based on these experimental findings, intermittent and predictive postural control models have been developed (Bottaro et al., 2008, Asai et al., 2009, Gawthrop et al., 2009). However, a number of these intermittent control methods have been problematic, since they involve the use of complex techniques while yielding few results that are physically relevant, such as balance corrective torques and joint angles (Asai et al., 2009, Gawthrop et al., 2009). Typically, the core of an intermittent control model is a phasic controller, which has previously been implemented as a "bang-bang" controller (Bottaro et al., 2008). The feed forward control signal is an efferent signal used to activate muscles in the bang-bang controller. In their controller, Bottaro et al. (2008) used a heuristic method to generate the phasic feed forward control signals, which, however, could not subsequently be generalized. For example, the control signal burst frequency was pre-determined at around 250ms, and the signal amplitude was empirically dependent on sway angle and ankle torque. Although this approach was suitable for certain upright stance scenarios, it was unclear whether it could be generalized to a wider range of control conditions. In general, an intermittent control model that does not rely on a limited intermittent control signal generator is preferred.

Localized muscle fatigue (LMF) has been reported to compromise postural control during quiet upright stance, as evidenced by decreased single limb stance time and increased variability of stability during walking (Helbostad et al., 2010, Paillard, 2012). Additionally, it has been demonstrated that center of pressure (COP) velocity increases following localized ankle muscle fatigue (Gribble and Hertel, 2004, Pline et al., 2006). Several other balance measures such as COP, COM, and time-to-boundary are also influenced by ankle LMF (Lin et al., 2009). Moreover, a model-based approach has suggested that the control feedback signal time delay is longer due to ankle LMF (Qu et al., 2009). In contrast, relatively few studies have used the intermittent control modeling approach to clarify changes in biomechanical-related parameters following ankle LMF. For instance, potential changes in the relative contributions of the passive and active joint torque associated with ankle LMF have not been described.

A substantial body of research has investigated the compromising effect of aging on postural control. Relatively poorer postural control among older adults is associated with numerous factors, including the following: impaired cognitive resources, a stronger reliance on visual information to maintain balance (Poulain and Giraudet, 2008), reduced concurrent cognitive performances (Lacour et al., 2008), lowered capability to process complex cognitive tasks (Bernard-Demanze et al., 2009), and a decline in joint range of motion, stimulus reaction time, joint proprioception, and overall physical strength (Schultz, 1992, Wojcik et al., 2001, Perry et al., 2007). These age-related physical and motor control changes have also been associated with adverse health outcomes. Lihavainen et al. (2010) described reduced physical capacity due to musculoskeletal illness and musculoskeletal pain, while Shaffer and Harrison (2007) reported on the functional declines of the somatosensory system, which are closely associated with postural control. Despite numerous prior studies, the effects of aging on a postural controller, and in particular, an intermittent postural controller, remain largely unknown.

Sliding mode control is a method developed for controlling a nonlinear system, and which is proficient at dealing with unknown or inaccurate system dynamics, thereby ensuring control robustness (Hernandez et al., 2011). Sliding mode control has been used in a variety of studies designed to model humanoid robot postural control, such as biped robot gait stability control (Wang et al., 2011), human motion synthesis (Spiers et al., 2010), knee joint actuator (Bae et al., 2010), and cooperative humanoid robots (Moosavian et al., 2011). Sliding mode control is particularly useful in that it is able to track both linear and nonlinear system dynamics (Aguiar and Hespanha, 2003, Guan et al., 2005). Additionally, sliding mode may be the underlying controller for both simple and complex human movements (Lim et al., 2003, Mohammed et al., 2005).

The purpose of the current research was to develop a sliding mode control model to simulate upright stance and postural sway, with a particular focus on whether such a model could do so with reasonable accuracy, as well as account for the effects of age and LMF. Through implementation of this modeling strategy, the tracking performances of body kinematics (e.g., COM angle, angular velocity, and angular acceleration) and kinetics (e.g., ankle torque) were evaluated. The sliding mode control model was also used to examine both passive and active contributions to ankle torque and their roles in intermittent control.

4.2 Methods

4.2.1 Experiment Procedures

Data from an earlier experiment (Lin et al., 2009) were used here, and which were obtained from a gender-balanced pool of 16 young and 16 older adults. As described in the noted study, participants completed repeated trials of quiet upright stance on a force platform (AMTI OR6-7-1000, Watertown, MA, USA) before any experimental interventions (i.e., “pre-fatigue,”) and then following a protocol designed to induce unilateral muscle fatigue in the ankle plantar flexors (i.e., “post-fatigue”). During these trials, participants were instructed to stand (without shoes) as still as possible with their feet together, arms by their sides, head upright, and eyes closed. Each trial lasted 75 seconds, with at least one minute between two consecutive trials. A total of 14 trials (3 pre-fatigue and 11 post-fatigue) were completed. Only data from the first three post-fatigue trials were used here, which were obtained 0.75, 2, and 4 min after the fatigue protocol.

Joint positions were estimated using 18 spherical reflective markers attached over bony landmarks. Marker locations were sampled (at 20 Hz) using a 6-camera motion capture system (Vicon 460, Lake Forest, CA, USA). COP time series were obtained from triaxial forces and moments sampled (at 100 Hz) from the force platform. Both COP and joint kinematics were low-pass filtered (Butterworth, 5 Hz cut-off frequency, 4th order, zero lag) and the first 10 s and the last 5 s were removed. Mean values of extremity joint centers were obtained across the bilateral values, which yielded a 2D representation. The obtained surface marker kinematics needed to be adjusted to locate joint centers, and this process is described in Appendix A.

Using the estimated joint kinematics, the whole body COM angle (from vertical) was calculated (see Appendix A). For simplicity, a single-segment inverted pendulum model of the body was employed (Figure 4.1), the plant dynamics for which are:

$$I\ddot{\theta} - Mg\hbar sin(\theta) = T \tag{1}$$

where θ is the whole body COM angle (or, ankle angle from vertical), $\ddot{\theta}$ represents whole body COM angular acceleration, M indicates body mass, g is the gravitational constant, h indicates the distance between the COM and ankle joint, and T represents the torque generated by the ankle.

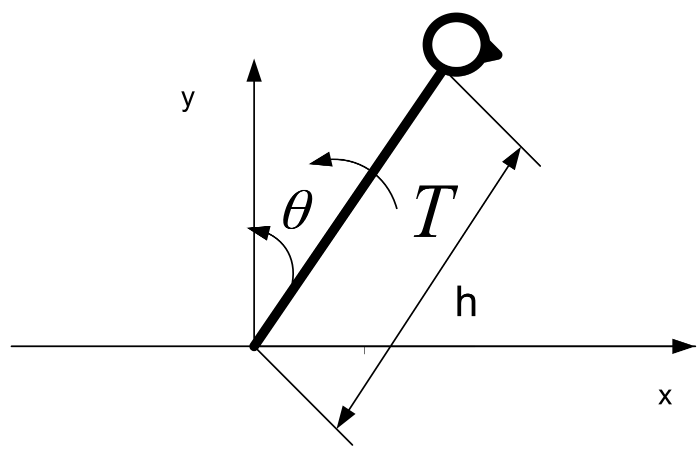

Figure 4.1: Single-segment inverted pendulum model of human upright stance.

4.2.2 Sliding Mode Controller

The goal of sliding mode control is to reduce an Nth order system to a 1st order system, thereby reducing the nonlinearity and uncertainty of the underlying control dynamics (Slotine and Li, 1991). Sliding mode control encapsulates multiple control dimensions to a single dimension, by creating a control error, as evidenced in the following equation:

$$\tilde{q} = q - q_d = [\tilde{q} \quad \dot{\tilde{q}} \quad \cdots \quad \tilde{q}^{(n-1)} \quad]^T \tag{2}$$

where q is the state variable (here, ankle angle θ), and q_d is the desired state. In our case, we chose n = 2. We constructed the sliding surface as follows:

$$s = \dot{\tilde{q}} + \lambda\tilde{q} \quad (3)$$

where λ is a positive numeric value that could affect the rate of estimate state convergence to desired state (Slotine and Li, 1991).

This sliding surface is of first order, where the state variable is also of first order. The sliding surface needs to satisfy the following equation:

$$\frac{1}{2}\frac{d}{dt}s^2 \le -\eta|s| \quad (4)$$

where η is strictly positive. This equation specifies that the control trajectory must point towards the sliding surface. Once the trajectory converges onto the sliding surface, it remains on the surface. The objective of sliding mode control is to force the initial trajectory to converge from an arbitrary position to the invariant sliding surface, which is typically also the desired trajectory. Convergence of the trajectory to the invariant sliding surface (s = 0) is demonstrated in Figure 4.2. As shown, there is a strong chattering motion around the invariant sliding surface, which is caused by intermittent switching of the control signals. This chattering is due to the intermittent control inputs that force the diverged trajectory back to the invariant sliding space.

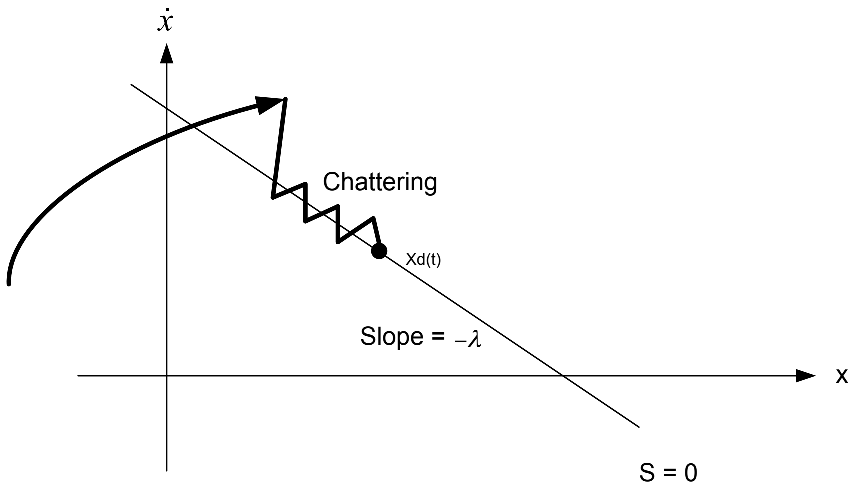

Figure 4.2 Sliding control surface, in which the chattering is a result of the intermittent switch of the control signals.

The first-order derivative of the squared sliding surface is:

$$\frac{1}{2}\frac{d}{dt}s^2 = \ s\,\dot{s} \tag{5}$$

Given the sliding surface $s = \dot{\tilde{q}} + \lambda\tilde{q}$, the first-order derivative of the sliding surface is:

$$\dot{s} = \ \ddot{q} - \ \ddot{q}_d + \lambda\dot{\tilde{q}} \tag{6}$$

From Equation 1, the plant dynamics of the inverted pendulum model are:

$$I\ddot{q} - \ Mg\hbar sin(q) = T \tag{7}$$

By substitution for $\ddot{q}$, the first-order derivative of the sliding surface becomes:

$$\dot{s} = \left(\frac{mg\hbar sin(q)}{I} \ + \ \frac{T}{I}\right) - \ddot{q}_d + \lambda\dot{\tilde{q}} \tag{8}$$

The best estimated ankle torque $\hat{T}$, which makes $\dot{s} = 0$ is thus:

$$\hat{T} = \ I\ddot{q}_d \ - \ \lambda I\dot{\tilde{q}} - \ Mg\hbar sin(q) \tag{9}$$

To account for both the imprecision of the system dynamics and the chattering behavior, the control law T has to be discontinuous (Slotine and Li, 1991), where a sign torque function of s is introduced as:

$$T = \hat{T} - k\eta sgn(s) \tag{10}$$

Where

$sgn(s) = +1$ if s > 0

$sgn(s) = -1$ if s<0

K is the control gain.

T can be further decomposed into passive torque and active torques:

$$T_{passive} = -Mg\hbar sin(q) \tag{11}$$

$$T_{active} = I\ddot{q}_d - \lambda I\dot{\tilde{q}} - k\eta sgn(s) \tag{12}$$

To achieve stability, the second-order derivative of the sliding surface has to satisfy the sliding mode condition as shown in Equation 4.

$$\frac{1}{2}\frac{d}{dt}s^2 \leq -\eta|s| \tag{13}$$

The left part of the equation is equal to:

$$\frac{1}{2}\frac{d}{dt}s^2 = s\dot{s} \tag{14}$$

By considering Equation 8, we can obtain the following equation:

$$s\dot{s} = s\left[\left(\frac{mg\hbar sin(q)}{I} + \frac{T}{I} - \ddot{q} + \lambda\dot{\tilde{q}}\right)\right] = s\left[\left(\frac{mg\hbar sin(q)}{I} + \frac{I\ddot{q} - \lambda I\dot{\tilde{q}} - Mg\hbar sin(q) - k\eta sgn(s)}{I} - \ddot{q} + \lambda\dot{\tilde{q}}\right)\right]$$

$$= s\left(\frac{-k\eta sgn(s)}{I}\right) = -\frac{K}{I}|s| \leq -\eta|s| \tag{15}$$

Compare the left and right part of the $-\frac{K}{I}|s| \leq -\eta|s|$ segment. In order to make the formula hold, K must be larger than zero. As such, the control gain K must be positive for achieving stable postural control. The following control gain was chosen to reduce chattering and make the trajectory s smoother.

$$K = |\lambda I \dot{\tilde{q}}| + \eta \qquad (16)$$

where $\eta > 0$

Total ankle torques, in the sagittal plane, were calculated based on force plate data, including COP, the vertical and horizontal (AP) forces, and foot moment (see Figure 4.3). For this, the vertical force Fy and horizontal force Fz were available from a force plate. Similarly, COPx and COPy represent the center of pressure coordinates, which are also obtained from the force plate measures. The foot mass Mfoot and footCM (distance between foot center of mass to ankle) were obtained from anthropometry tables (Enoka, 2008). Ankle height (AH) was obtained from marker data.

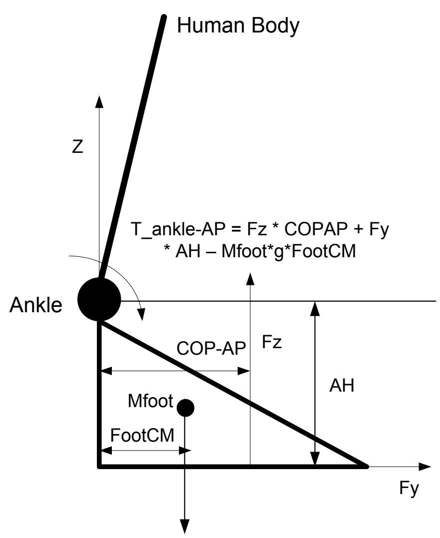

Figure 4.3: Illustration of methods used to determine experimental ankle torque (T_ankle-AP).

The steps used to calculate COM angle, COM angular velocity, COM angular acceleration, and ankle torque are shown in Figure 4.4, along with the steps involved in applying the sliding mode control method to model upright stance. These same steps were used to process each trial. The Matlab ODE4 solver was used to formulate the upright stance dynamics model, which also produced an estimated COM angle, COM angular velocity, COM angular acceleration, and passive and active ankle torque.

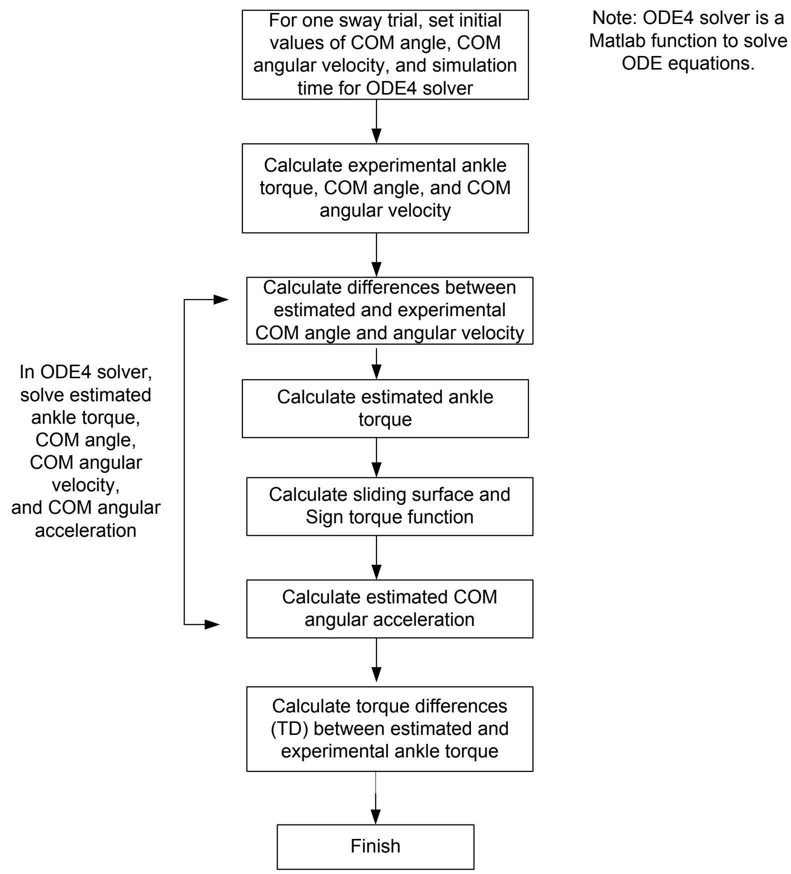

Figure 4.4: Flow chart for using sliding mode control to model and simulate upright stance.

To evaluate the tracking performance of the sliding mode controller, root mean square errors (RMSEs) between experimental and modeled ankle kinematics and torques were determined for each trial. Specifically, RMSEs for COM angle, angular velocity, angular acceleration, and total ankle torque were obtained. In addition, Pearson product-moment correlation coefficients were calculated to compare experimental and modeled ankle torques. A correlation coefficient that was greater than 0.9 was interpreted to be strong. Within each trial, mean results for both passive and active predicted ankle torque were determined, along with the associated mean passive/active torque ratio. Phase relationships between passive ankle torque and COM angle, was well as active ankle torque and COM angular acceleration were also obtained. The main purpose of the latter was to determine if there was phase coherence between the pairs of variables.

4.2.3 Statistical Analysis

Respective within-subject means for the dependent variables (i.e., COM angle, angular velocity, and angular acceleration, total ankle torque tracking errors, modeled and experimental ankle torque correlation, passive ankle torque, active ankle torque, and the passive/active ankle torque ratio) were obtained for the three pre-fatigue and three post-fatigue trials. These mean values were subsequently used in statistical analyses. The effects of age and gender were evaluated for pre-fatigue trials with separate two-way ANOVAs for each of the dependent variables. The differential effects of fatigue associated with age and gender were analyzed using separate two-way ANCOVAs on change scores (post-fatigue – pre-fatigue) for each dependent variable, with independent variables of age and gender, and the pre-fatigue measure as a covariate. The level of significance for all tests was set at $p < 0.05$, and all statistical analyses were performed using JMP 9.02 (SAS Inc., Cary, NC, USA).

4.3 Results

4.3.1 Representative Trial

The sliding surface and first-order derivative of the sliding surface for a representative trial are shown in Figure 4.5. It shows despite a large overshoot to about 0.15 for both sliding surface and its first order derivative at the initial starting point (T=0) due to the large difference between initial desired and estimated ankle kinematics, the close to zero values are evident for both of these values through the entire simulation. The phase plot of the sliding surface also quickly converged to zero with small oscillations. It is also apparent that the phase starts from 0.15 for both sliding surface and its first order derivative; it then quickly converges to zero and also maintains its position around zero during the simulation period.

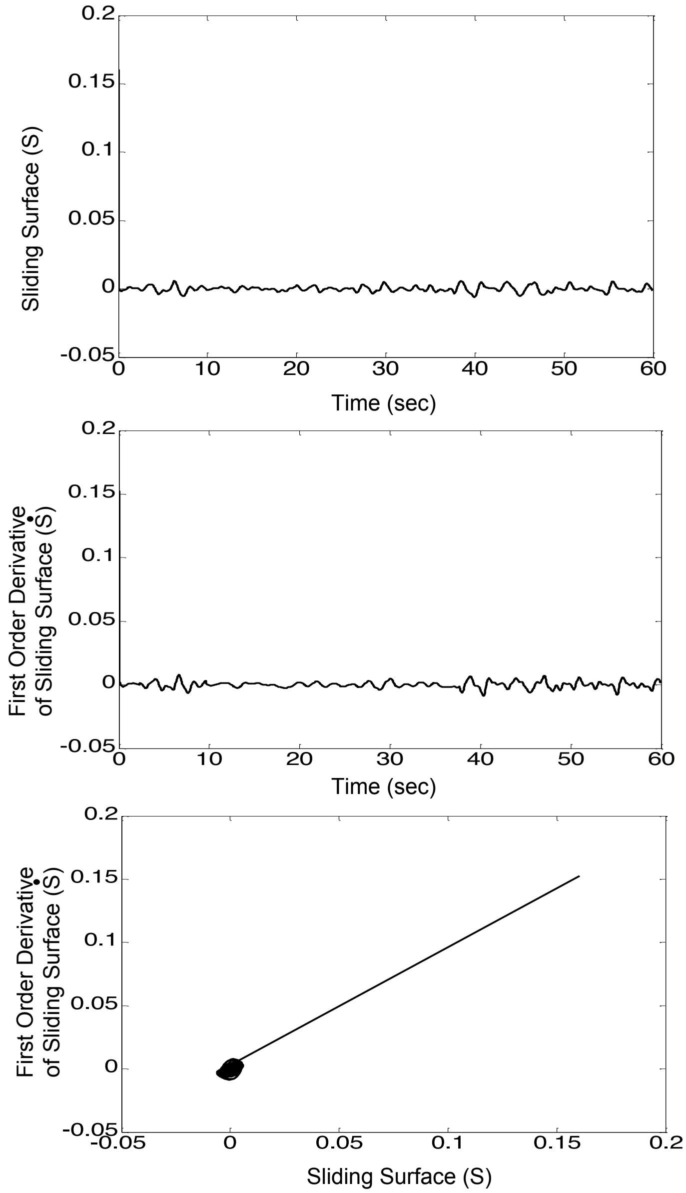

Figure 4.5: Sliding surface (top), first order derivative of sliding surface (middle), and phase plot of the sliding surface (bottom) for a representative trial.

The tracking performance of the sliding mode for this same trial is illustrated in Figures 4.6 and 4.7, which indicate relatively small errors between the modeled and experimental

COM kinematics. Both modeled and experimental ankle torque results for this trial are shown in Figure 4.8, which indicates that modeled ankle torque results followed experimental values with only a small tracking error.

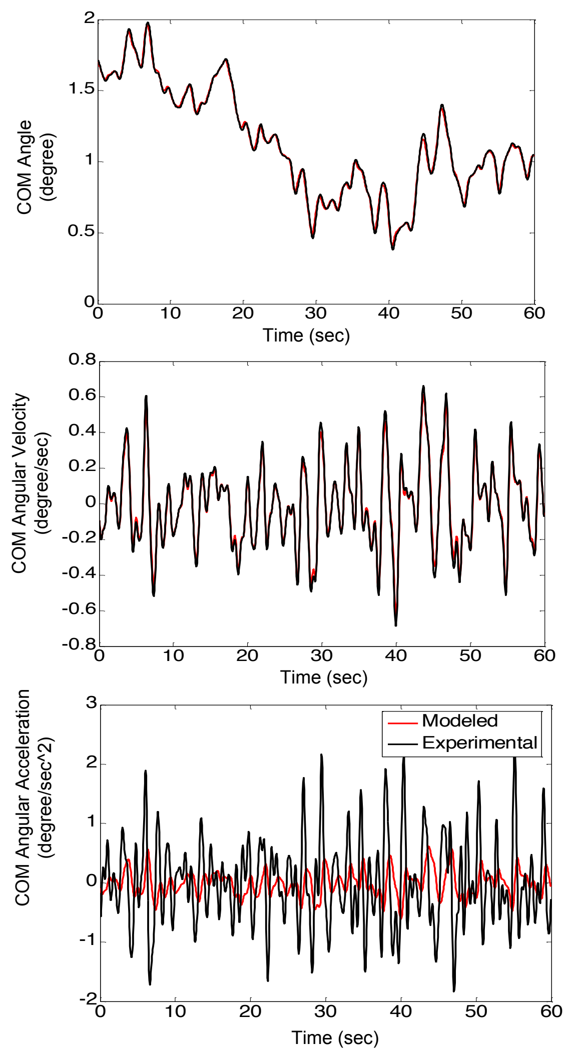

Figure 4.6: Modeled and experimental COM angle (top), angular velocity (middle), and angular acceleration (bottom) for a representative trial.

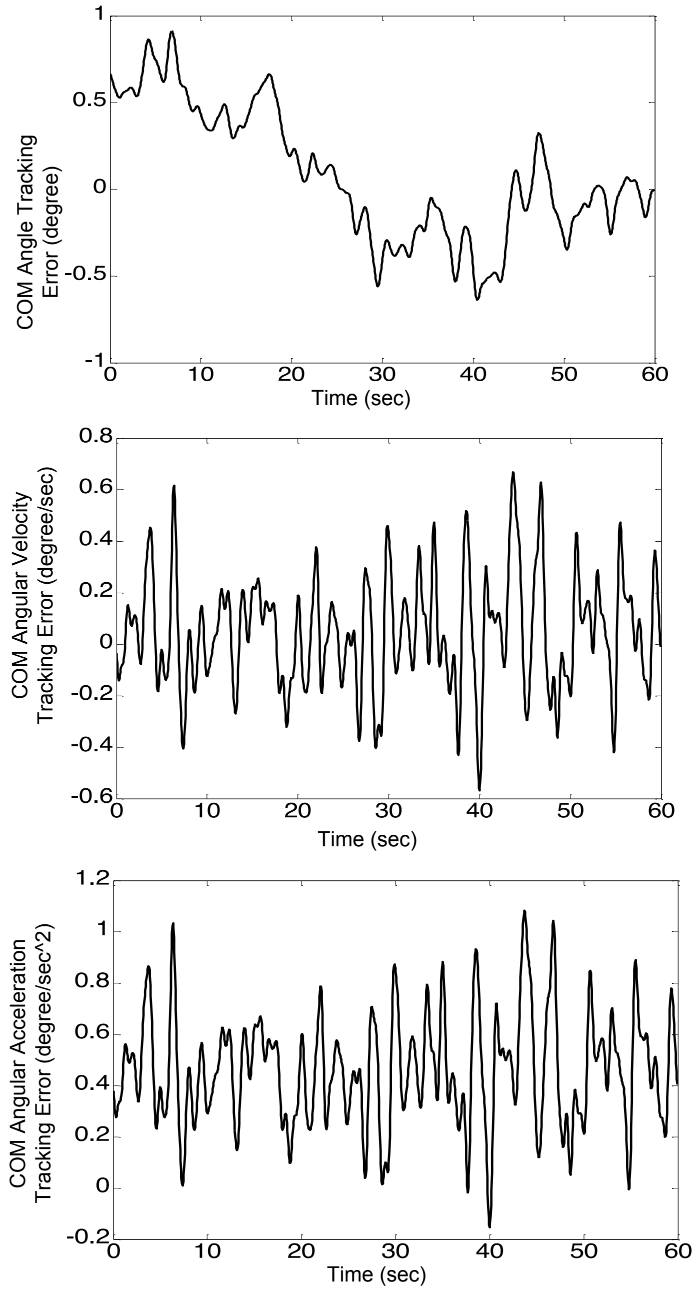

Figure 4.7: COM angle (top), angular velocity (middle), and angular acceleration (bottom) tracking errors for a representative trial.

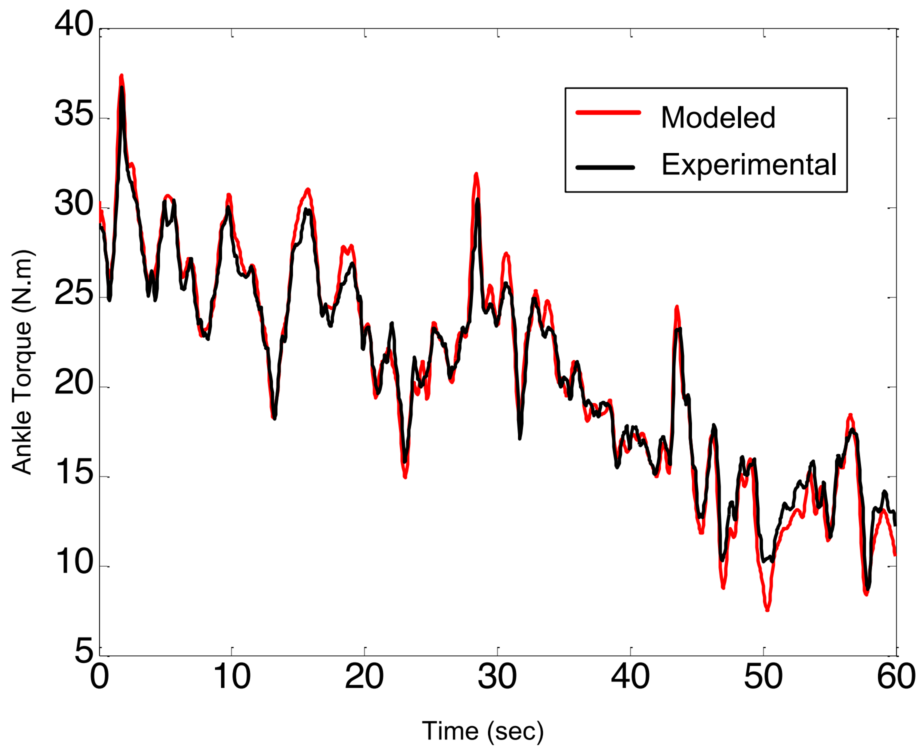

Figure 4.8: Modeled and experimental ankle torque for a representative trial.

Results for passive ankle torque and COM angle for the trial are shown in Figure 4.9, and active ankle torque and COM angular acceleration are shown in Figure 4.10. These figures confirm that passive ankle torque and COM angle, as well as active ankle torque and COM angular acceleration, were both strongly correlated and in-phase.

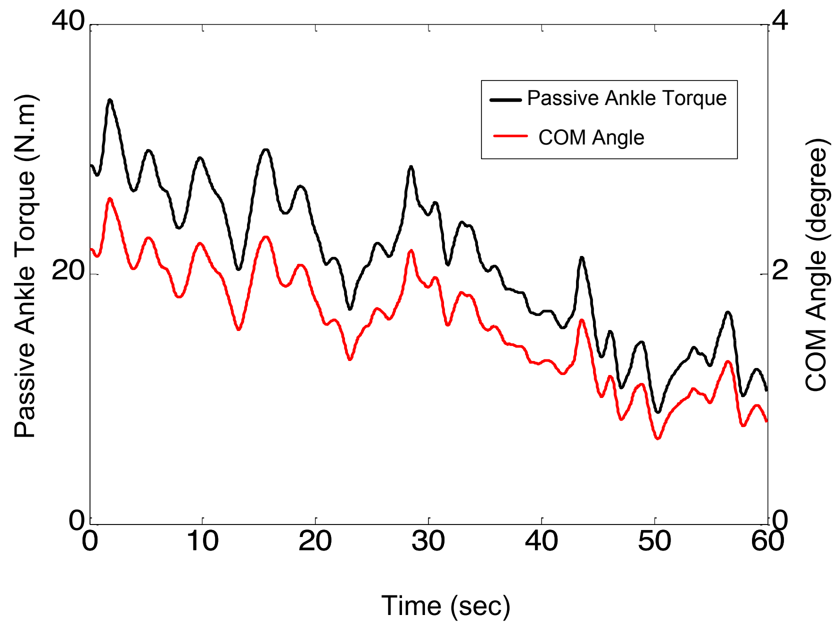

Figure 4.9: Passive ankle torque and COM angle for a representative trial.

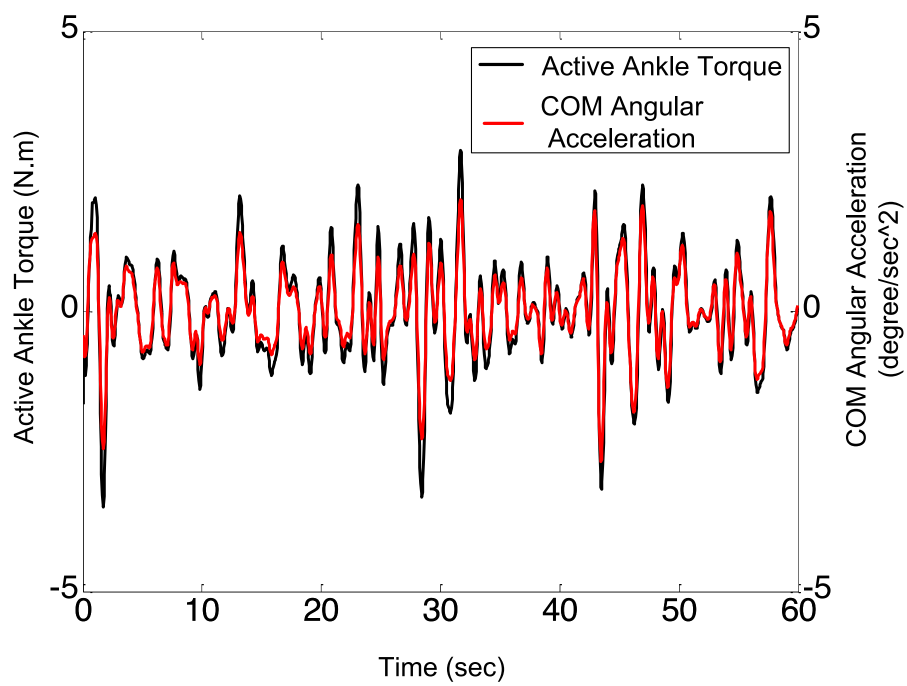

Figure 4.10: Active ankle torque and COM angular acceleration for a representative trial.

4.3.2 Statistical Results

A summary of statistical results for both pre-fatigue and fatigue-induced outcomes on both controller tracking performance and passive/active ankle torques is provided in Table 4.1, which also includes summary statistics for pre- and post-fatigue measures. Additional details are presented below, separately for pre-fatigue and fatigue-induced effects.

Table 4.1: Summary of statistical results (p values) for the effects of age (A), gender (G), and ankle LMF. M/E = Modeled/Experimental. Significant effects ($p < 0.05$) are bolded.

		Pre Fatigue				Fatigue			
		Mean(std)	A	G	AXG	A	G	AXG	Mean(std)
Controller Tracking Performance	COM Angle Tracking Error (degree)	0.51(0.24)	0.57	0.24	0.17	0.21	0.54	0.63	0.59(0.29)
	COM Angular Velocity Tracking Error (degree/sec)	0.41(0.19)	0.12	0.40	**0.020**	0.19	0.37	0.23	0.42(0.17)
	COM Angular Acceleration Tracking Error (degree/sec^2)	0.94(0.28)	**0.0032**	0.72	0.071	**0.020**	**0.0062**	0.41	1.01(0.28)
	Ankle Torque Tracking Error (N.m)	2.71(2.13)	**0.018**	**0.0064**	0.36	0.46	0.18	0.59	3.44(2.38)
	M/E Ankle Torque Correlation	0.95(0.04)	0.12	0.89	0.75	0.85	**0.026**	0.44	0.93(0.04)
Passive & Active	Passive Ankle Torque (N.m)	34.17(13.06)	0.86	0.52	**0.0093**	0.47	0.36	0.48	31.93(13.31)
	Active Ankle Torque (N.m)	0.91(0.48)	**0.0006**	**0.0004**	**0.022**	**0.0270**	**0.0083**	0.19	0.97(0.5)
	Passive/Active Torque Ratio	44.45(23.56)	**0.0069**	**<0.0001**	0.12	0.31	0.85	0.43	40.68(23.54)

4.3.3 Pre Fatigue

While COM angle ankle tracking errors and modeled vs. experimental ankle torque correlations were consistent across the groups, other measures of controller tracking performance differed. Specifically, errors in COM angular velocity tracking were larger for older males, followed by older females and younger females, and smallest among younger males (Figure 4.11-1). Similarly, COM angular acceleration tracking errors were larger in the older group (Figure 4.11-2), while ankle torque tracking errors were greater in the older group and among males (Figure 4.12).

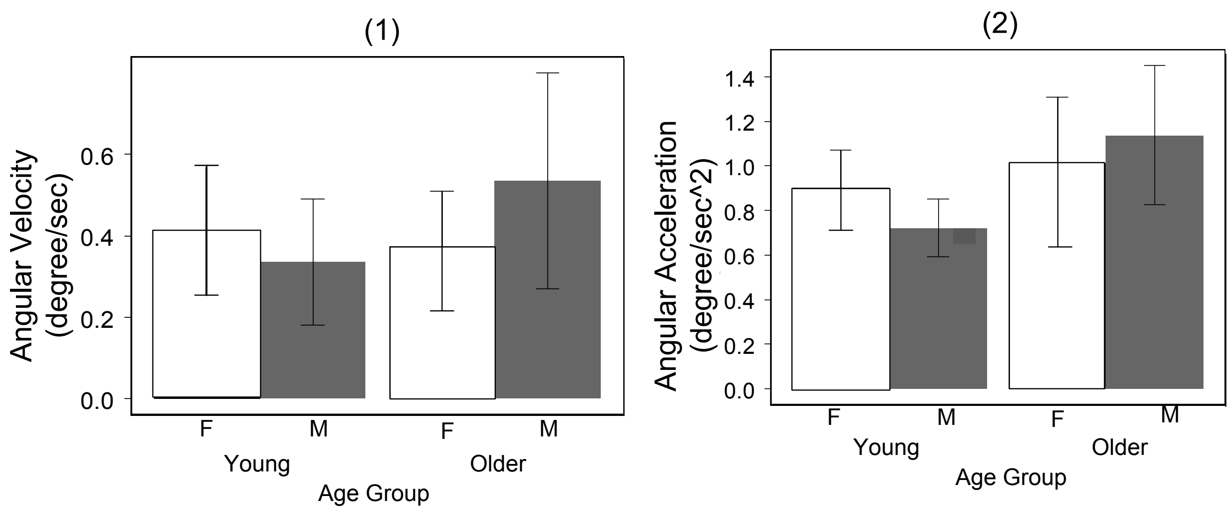

Figure 4.11: Pre-fatigue results for tracking errors in (1) angular velocity and (2) angular acceleration. Among them, (2) is significantly different between young and older adults.

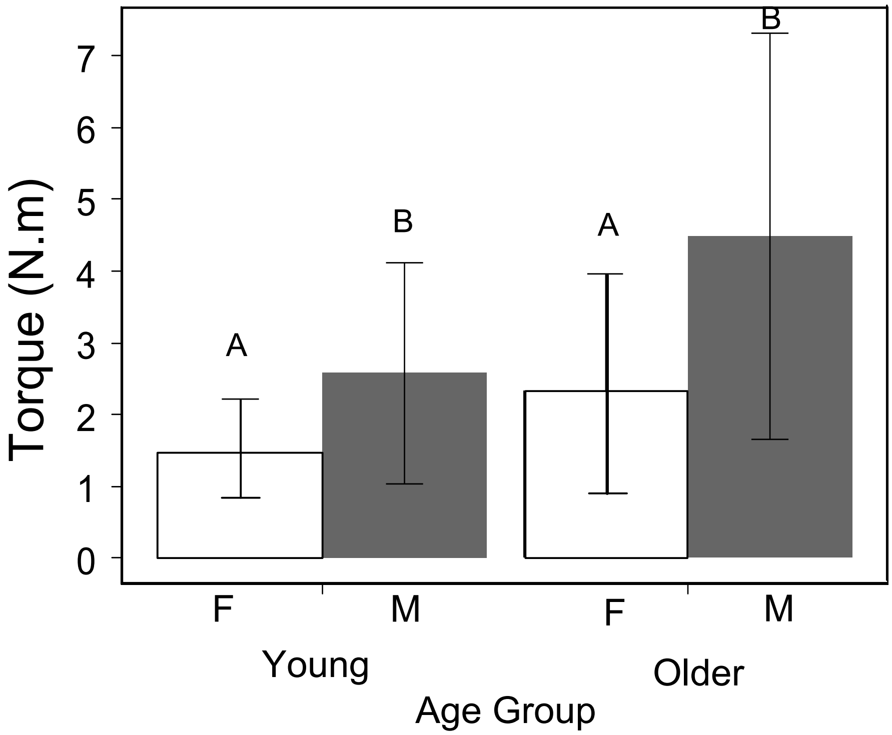

Figure 4.12: Pre-fatigue tracking errors for ankle torque. Young and older adult active ankle torque are significantly different. A and B denote active ankle torque differences between genders.

Passive ankle torque was larger among young females, followed by (in order) older females, older males, and younger males (Figure 4.13-1). Active ankle torque was greatest in the older group as well as among males (Figure 4.13-2); in contrast, the passive versus active ankle torque ratio was larger in the younger group and among females (Figure 4.13-3).

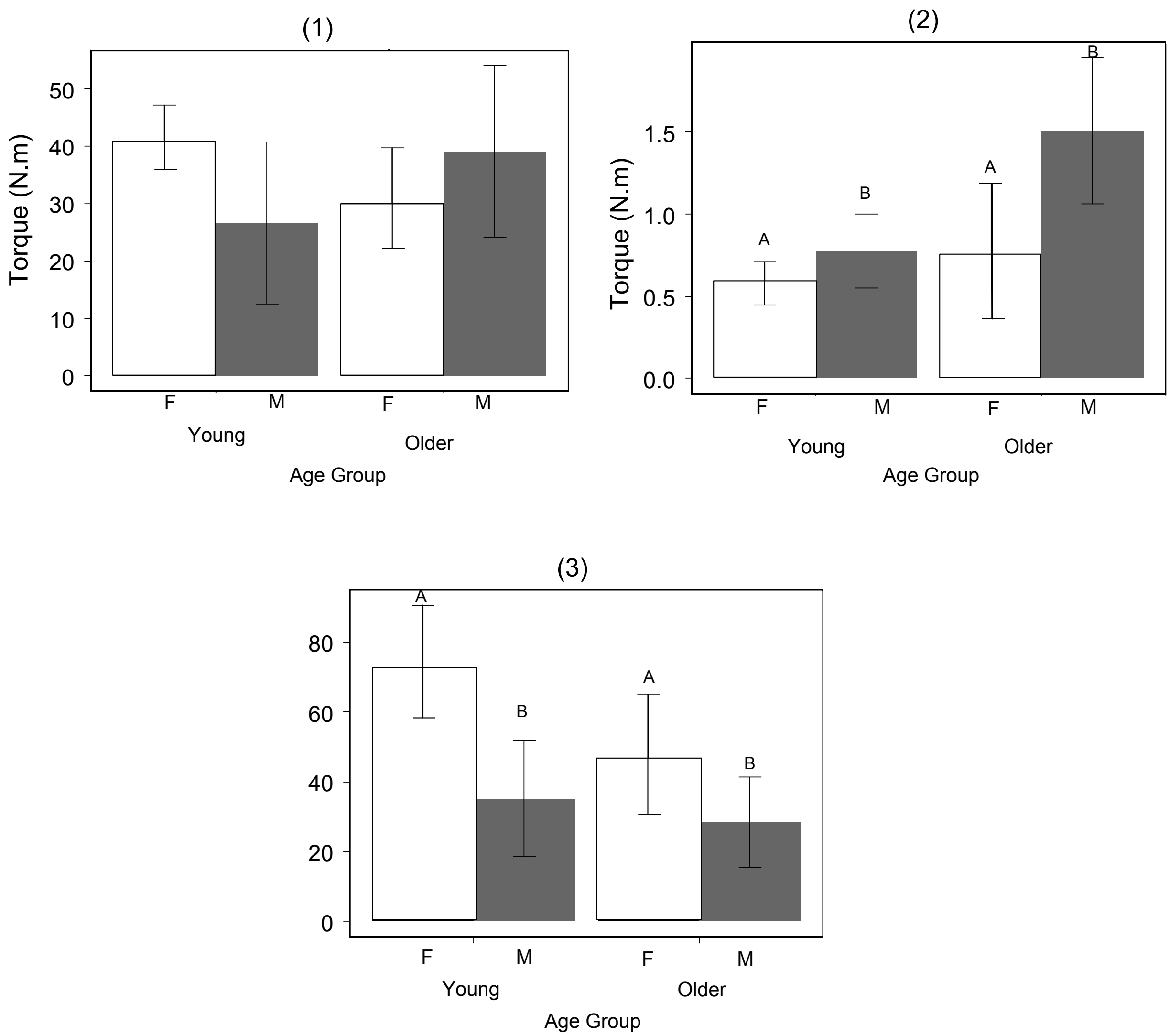

Figure 4.13: Pre-fatigue (1) passive ankle torque, (2) active ankle torque, and (3) passive/active ankle torque ratio. Among them, (2) and (3) are significantly different between young and older adults. A and B denote that (2) and (3) are different between genders.

4.3.4 Post Fatigue

All measures of controller tracking performance changed post-fatigue (Table 4.1), suggested a slight degradation in model tracking performance. Specifically, kinematic and torque tracking errors increased, and there was a decrease in the correlation between measured and experimental ankle torques. These changes were generally consistent across age and gender groups, with a few exceptions. COM angular acceleration tracking errors were larger in the younger group and among males (Figure 4.14-1), and females had larger decreases in post-fatigue correlations between modeled vs. experimental ankle torque (Figure 4.14-2). The passive ankle torque contribution decreased post-fatigue, consistently across age and gender groups. In contrast, the active contribution increased, and a larger increase occurred among the young group and among males (Figure 4.14-3).

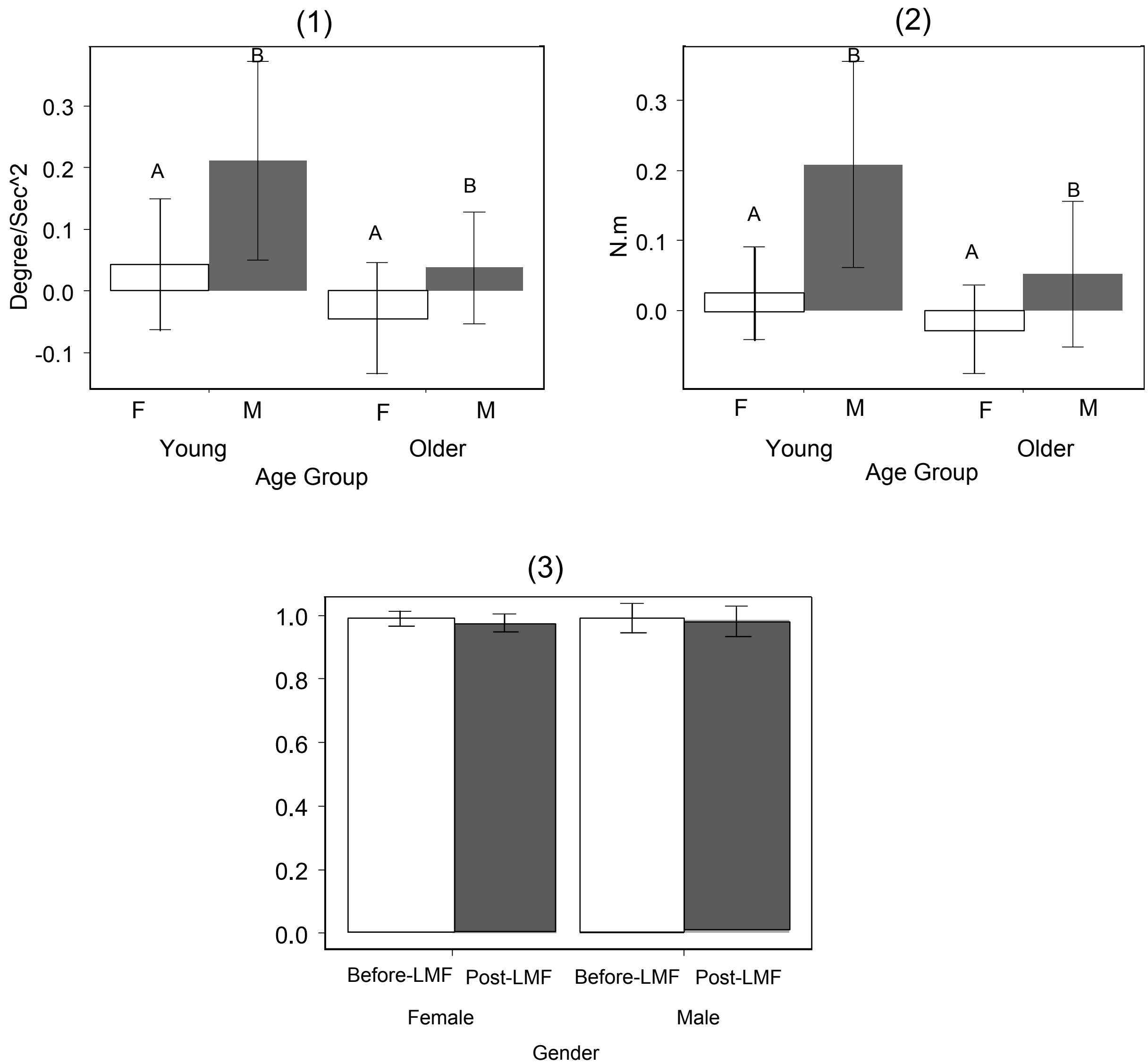

Figure 4.14: Post-fatigue changes in (1) COM angular acceleration tracking errors, (2) Active ankle torque, and (3) the correlation between modeled and experimental ankle torque. Among them, (1) and (2) are significantly different between young and older adults. A and B denote that (1) and (2) are differed between genders.

4.4 Discussion

The purpose of this study was to apply sliding mode control theory to model human upright stance. As noted above, this work was motivated by recent reports suggesting that upright stance is intermittently (vs. continuously) controlled (Loram et al., 2009, Gawthrop et al., 2011, Loram et al., 2011, Loram et al., 2012, Vieira et al., 2012). Because sliding mode is essentially an intermittent control method, it was considered appropriate for modeling upright stance. It should be noted that the current model is based on the assumption that the human balance controller controls the body based on a sliding surface condition. Thus, the control objective was to force the sliding surface to remain as close to zero as possible, thereby retaining control stability. Any perturbations that force the sliding surface away from zero would be pushed back to zero by the control input. Essentially, the sliding surface always oscillated at a near-zero value—although not equal to zero. Therefore, the control input also oscillated at approximately zero, thus yielding an intermittent force or moment—namely, an intermittent controller.

In this study, tracking errors were quantified (as root mean square errors), with results showing tracking errors between (1) experimental and estimated COM angle and (2) angular velocity that may be considered acceptable. Specifically, the mean tracking errors for these two kinematics measures were in the following respective ranges: [0.45, 0.7] degree and [0.33, 0.53] degree/s. Considering that the mean COM angle was 2 - 3 degrees, these tracking errors were comparatively small. Since angular velocity tracking errors account for 5 – 10% of experimental angular velocity, this means that the sliding mode control was able to reasonably track the desired COM angle and angular velocity.

The sliding mode controller tracking performance was comparable—in fact, slightly smaller—in comparison to the bang-bang intermittent controller developed by Bottaro et al. (2008). In terms of differences related to age and gender, COM angular velocity tracking errors were largest among older males and smallest for younger males. We attribute the reduced tracking performance among older males to age-related differences, such as the greater waist-to-hip ratio and larger intra-abdominal adiposity among older adults (Borkan et al., 1983, Schwartz et al., 1990, Hughes et al., 2004), and which resulted in a loss of accuracy when estimating COM angle. Mean COM angular acceleration errors were 0.72 degree/s^2 for older males, and 1.17 degree/s^2 for younger males, which accounted for 20 – 40 % of the experimental angular acceleration. The angular acceleration tracking performance was lower than that of either angle or angular velocity. This difference is likely tied to the fact that in the sliding mode controller the angle and angular velocity-based sliding surface condition was built into the controller. This design decision was made based on available literature suggesting that angle and angular velocity are important reference signals for a postural controller (Peterka, 2002, Masani et al., 2003, Qu et al., 2009). Consistent with angular velocity tracking results, and likely for the same reason, COM angular acceleration tracking errors were also larger for older adults.

Tracking errors between experimental and simulated ankle torque were also relatively small. Specifically, the mean ankle torque tracking error was in the range of 1.45 - 4.48 Nm. In comparison to overall ankle torque, which was in the range of 26 - 42 Nm, this error only accounted for 5 – 6% (and a maximum of 10%) of the entire ankle torque. A high correlation between experimental ankle torque and modeled ankle torque was also found. As a whole, these results support that the sliding mode approach can track experimental ankle torque while yielding relatively small tracking errors. Similar to kinematics tracking performance, ankle torque tracking performance was comparable to that of the bang-bang controller (Bottaro et al., 2008). Ankle torque tracking errors were also larger for males in both age groups. This difference may be linked to the different COM angle calculation errors for the two age and gender groups. Additionally, males typically have greater skeletal muscle mass, especially in the upper body (Janssen et al., 2000). Thus, the different COM angle estimates we obtained in this study are probably related to differences in mass distributions between genders, and which affected the center of mass position.

With respect to passive and active ankle torque, from the sliding mode torque equation (Eq 9),

$T = I\ddot{q}_d - \lambda I\dot{\tilde{q}} - k\eta sgn(s) - Mg\hbar sin(q)$

where $k = |\lambda I\dot{\tilde{q}}| + \eta, \eta > 0$

it is apparent that the fourth term, $Mg\hbar sin(q)$, which corresponds to gravitational torque, is linearly proportional to the sway angle (q). Such a linear relationship can either be due to continuous feedback from ankle angle or to passive ankle elasticity, which acts like a spring. With respect to upright stance, it has been suggested that there is no continuous feedback used to control sway stability (Loram et al., 2005a); as such, gravity torque should not be attributed to feedback control mechanisms. As indicated by Bottaro et al., (2008), the only possible physical source for gravity torque is related to the ankle passive stiffness structure, the series elastic elements (SEE), which is a primary source of ankle passive stiffness and damping. The other three terms in the torque equation above are related to desired angular acceleration $I\ddot{q}_d$ (1st term), the difference between the desired angle and the present state of angle $\lambda I\dot{\tilde{q}}$ (2nd term, angle feedback), and the sliding surface control term $k\eta sgn(s)$ (3rd term). These three variables are related to desired ankle angular acceleration, the derivative of desired and experimental ankle angular velocities, and active control gain, which are all associated with overall active ankle torque. We showed earlier that passive ankle torque is well correlated to the COM angle (Figure 4.9). In contrast, active torque is closely correlated to the COM angular acceleration (Figure 4.10). These findings indicate that active ankle torque is used to correct sway acceleration direction, by actively increasing or decreasing the ankle angular acceleration as indicated earlier (Zatsiorsky and King, 1997, Loram and Lakie, 2002b). In contrast, passive ankle torque acts more like a linear elastic spring.

The mean passive (intrinsic) ankle torque generated by the sliding mode was roughly 97% of the total torque for young adults; for older adults, the corresponding proportion was somewhat smaller (93%). This suggests a predominant contribution of passive torque during upright stance. Passive ankle torque is provided by a number of series elastic elements (SEE), including ankle muscles, the aponeurosis, Achilles tendons, and foot structures. Previous research has shown that SEE elements at the ankle are still very close to—or even overcome—the critical ankle torque needed to maintain upright stance stability (Winter et al., 2001, Loram et al., 2005a). However, our results suggest that active ankle torque still plays a role (even if a small one) during upright stance. Similar proportions of active vs. passive contributions were reported earlier (Loram and Lakie, 2002a), specifically that passive torque accounts for 91 ± 23% of critical toppling torque. However, our estimate of passive ankle torque was significantly larger than a more recent estimate of 64 ± 7.8 % for critical toppling torque (Casadio et al., 2005).

The differences between the current results and prior studies, in terms of active and passive ankle stiffness, can be attributed to the use of imposed external perturbations. Casadio et al., (2005) have argued that larger external perturbations induce smaller passive stiffness. Moreover, both Loram and Lakie (2002a) and Casadio et al. applied two specific perturbation magnitudes (0.055° and 1°, respectively) to upright stance in their studies. They found that passive stiffness tended to decrease due to increases in perturbation magnitude. It is likely, therefore, that even small perturbations (e.g., as small as 0.055°) can alter the actual underlying passive and active ankle stiffness. Due to these perturbations, the response mechanisms associated with upright stance could change—perhaps with increased muscle activity—thereby further increasing active ankle torque. Other perturbation-associated changes such as joint coordination have also been reported to alter the COM angle compared to a non-perturbation scenario (Patel et al., 2010). Since the COM angle is linearly proportional to passive ankle torque, any external perturbations could change the ratio of passive versus active ankle torque.

Here, older adults displayed greater active ankle torque compared to younger adults. We attribute this finding to the fact that increased active ankle torque is needed to maintain upright stance stability, which is compromised due to aging. And indeed, numerous studies have suggested that among older adults postural control is compromised by a variety of factors: diminished postural sensory sensitivity and latency (Shumway-Cook and Woollacott, 2000, Speers et al., 2002, Qu et al., 2009), increased muscle response latency (Woollacott et al., 1986, L Sturnieks et al., 2008), decreased lower extremity strength (Wolfson et al., 1996, Brown et al., 2000), and stronger demands for cognitive attention to maintain postural balance (Brown et al., 1999, Redfern et al., 2001, Huxhold et al., 2006). As such, the increased active ankle torque observed in the older group could reflect compensation for a loss of peripheral sensory sensitivity and decreased cognitive capacity. In contrast, the increased active ankle torque found here does not appear to be associated with the magnitude of passive ankle torque among older adults, which did not differ from the younger group.

With respect to gender differences, males had larger active ankle torques than females, and this gender difference was more pronounced in the older group. It is unclear as to why this was found and whether such a difference is consistent with postural control performance. Regarding the latter, some studies have shown that older females are at greater risk for (Wolfson et al., 1994), while other studies have suggested the opposite (Masui et al., 2005). Moreover, Bryant et al. (2005) was unable to document that either gender was at greater risk for falls. Our results, however, seem to support that males utilize more control resources—possibly an indicator of the increased risk of falls. Even though we could not account for all possible contributing factors, the greater active ankle torque observed among older males could be associated with gender-based differences in the neuromuscular system that play a role in physiological responses (Hunter, 2009).

Following ankle LMF, active ankle torque increased more among young adults. This finding means that ankle LMF resulted in more adverse effects in this group, eliciting a larger active ankle torque response. Studies have suggested that older adults have increased endurance (or resistance to muscle fatigue) because they rely less on anaerobic metabolic pathways for (Kent-Braun et al., 2002). Additionally, several researchers have described the higher percentage of type I (slower and fatigue-resistant) muscle fibers in older adults (Chan et al., 2000, Ditor and Hicks, 2000, Baudry et al., 2007). Other factors such as joint coordination could also play a role in explaining the higher active ankle torque among young adults. As recently reported by Paillard (2012), older adults might rely on a "hip-centered" strategy to counteract the adverse effects of ankle muscle fatigue. The age-related differences in post-fatigue active torque changes might not be attributed to the ankle fatigue protocol, since consistent levels of absolute sub-maximal voluntary contractions done by different age groups was found not lead to age-related differences in fatigue (Hunter et al., 2005).

Compared with females, males displayed larger post-fatigue active ankle torque changes, which is consistent with a previous report (Kent-Braun et al., 2002). Multiple factors could account for this finding: lower muscle contraction forces, greater oxidative metabolism capacity, less likelihood to experience central fatigue (Paillard, 2012), reduced intramuscular pressures, and less occlusion of muscle blood flow (De Ruiter et al., 2007). Additionally, as reported by Hunter et al. (2004), females were able to perform longer duration sub-maximal elbow contractions, even when accounting for muscle contraction strength difference between genders. This finding indicates that metabolic pathways are more likely to account for gender-based differences in fatigue endurance (Enoka and Duchateau, 2008). In contrast to Hunter's (2009) report that older females have reduced fatigue endurance compared with older males, the older females in our study displayed smaller changes in active ankle torque than older males following ankle LMF.

There are several limitations associated with this study that must be acknowledged. We identified the hip joint center and L5/S1 locations, which provided essential data for this work, using ASIS surface markers. We acknowledge that even though care was used to ensure precision in locating these positions, some computational errors could be embedded in this process. As indicated in Appendix A, a constrained optimization process was utilized for three reasons. First, the hip joint center and L5/S1 positions differ between individuals (De Leva, 1996). Therefore, it was necessary to generalize these values across all participants, which presents some inherent drawbacks. Second, a small anatomical adjustment to the ASIS marker was performed. However, the magnitude and orientation of this adjustment correspond to previously described anatomical constraints used by other researchers (see Appendix A). Third, we argue that the adjustment used in this study still ensures a strong correlation between modeled ankle torque and experimental ankle torque both before and after the adjustment. In fact, the resulting COM angle following the adjustment process was consistent with prior reports. For example, our results showed that the COM angle was larger among females, which was also reported by (Panzer et al., 1995).

Another limitation in this study is that we simplified the model by using a single-segment inverted pendulum to represent human upright stance. Even though such a simplification has been criticized in the literature (Hsu et al., 2007, Kiemel et al., 2008), other studies have used similar strategies (Loram and Lakie, 2002b, Peterka and Loughlin, 2004, Qu et al., 2009). Additionally, there are good reasons to support this decision, including relatively small hip, trunk, shoulder, and neck motions during quiet upright stance. Nonetheless, the use of a single-segment inverted pendulum model likely resulted in some errors, and did not allow for inclusion of potentially important trunk and shoulder contributions to the intermittent postural controller. Thus, subsequent studies would benefit by employing a two-segment (or more) inverted pendulum model in sliding mode control.

4.5 Conclusions

For this study, a sliding mode intermittent controller was used to model upright stance. This work was motivated by experimental evidence suggesting that upright stance is under intermittent control. The modeling results demonstrated that the sliding mode was able to track experimental COM angle, COM angular velocity, and COM angular acceleration reasonably well. Moreover, the sliding mode controller was able to track experimental ankle torque with reasonable accuracy. The sliding mode controller was also used to estimate ankle torque, from which passive ankle torque and active ankle torque can be directly derived in analytical form. Our results for passive and active ankle torque were consistent with previous research findings. Use of a sliding mode intermittent controller is considered to a fruitful approach for understanding human postural control

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Appendix A

Determination of Hip Joint Center and L5/S1 Positions

Accurately locating joint centers through surface markers is critical for ensuring the precision of whole body COM location estimates. In the experiment from which the current data were obtained, reflective markers were attached to the following bony landmark positions: ankle malleoli, knee lateral femoral epicondyles, shoulder acromion, and head temple. The kinematics results showed that these markers are located superficially near the ankle, knee, shoulder joint centers, and head COM, respectively, in both the antero-posterio (AP) and vertical directions, and which did not need position adjustments (Majeske and Buchanan, 1984, Holden and Stanhope, 1998, Bruening et al., 2008). Specifically, for the head marker, preliminary analysis showed that the head reflective marker should be located about 12cm from the top of head and 9 cm from the back of head, supporting a prior report showing that the head CMS is located 10 -12cm from the top of head and 7 – 9 cm from the back of head (Yoganandan et al., 2009). As such, the head reflective marker position was also not adjusted.

For each participant, the hip joint center and L5/S1 positions were obtained by adjusting the locations of two markers attached to the anterior superior iliac spine (ASIS) via an optimization process. The marker adjustment process assumed negligible joint center migration during quiet upright stance. Therefore, any pelvis rotation was not accounted for during this adjustment process and only horizontal and vertical translations in the AP plane were conducted. These joint centers differ on an individual basis (De Leva, 1996), suggesting that a universal adjustment of the marker positions across participants that was utilized could have resulted in a faulty estimation of the joint centers. Accordingly, and to locate the hip joint center and L5/S1 based on the ASIS position, a constrained optimization process was undertaken (Figure 4.15).

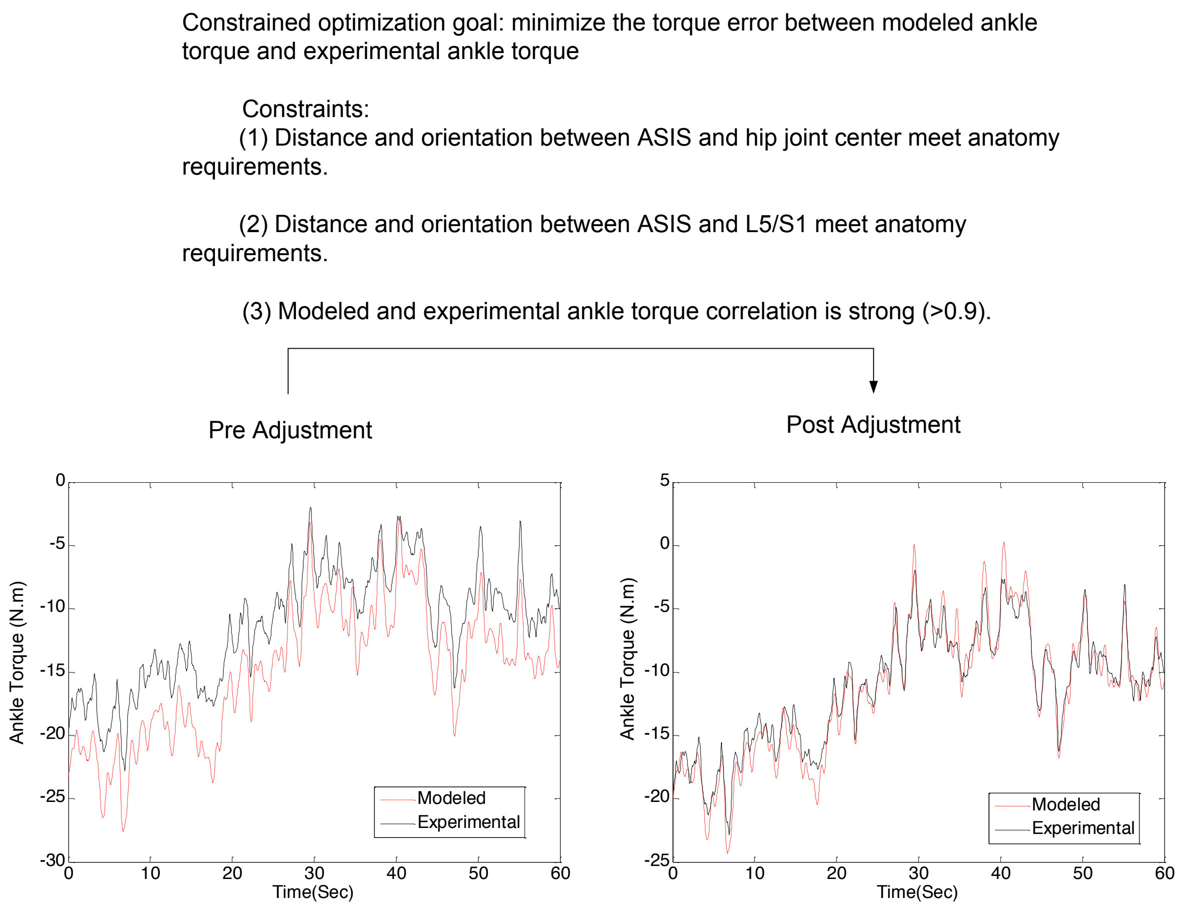

Figure 4.15: The constrained optimization process used to determine the hip joint center and L5/S1 positions based on ASIS markers.

The optimization process started with an initial rough estimate of the hip joint center and L5/S1 positions relative to ASIS. Based on this rough estimate, we then obtained an error associated with modeled ankle torque and experimental ankle torque. Subsequently, we attempted to minimize the error between experimental ankle torque and modeled ankle torque. Noted constraints were that the hip joint center and L5/S1 distances and orientations to ASIS had to be within accepted boundaries as described in the literature (Majeske and Buchanan, 1984, Seidel et al., 1995).

To satisfy anatomical constraints, starting from the ASIS, the hip joint center was located about 34% (about 3 – 9cm) from the pelvic depth posterior and 79% (about 5 – 8cm) from the pelvic height inferior (Seidel et al., 1995). The ASIS marker position was also used to project to the location of L5/S1, based on reports that L5/S1 should be located about 40% to 50% of the distance of the ASIS (beginning) to the posterior-superior iliac spine (PSIS) (end), and about 1.0cm - 1.5cm superior starting from the ASIS (Mackinnon and Winter, 1993, Reed et al., 1999, Larivière et al., 2001). Another important constraint was that a strong correlation (usually >0.9) between modeled and experimental ankle torque had to be ensured. The hip joint center and L5/S1 adjustment magnitudes and directions relative to the ASIS marker in the AP -plane are shown in Figure 4.16.

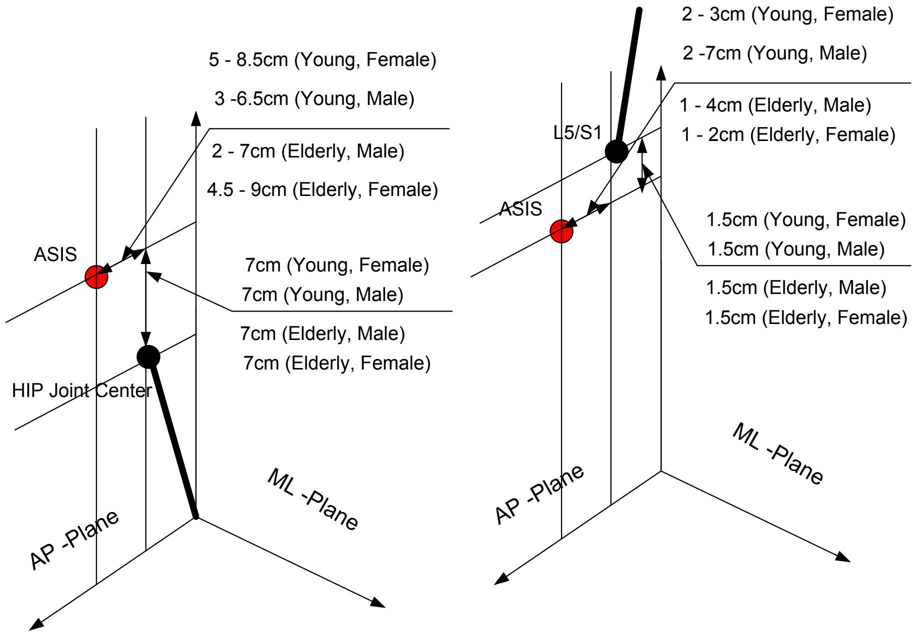

Figure 4.16: Adjustments of ASIS marker to hip joint center and L5/S1 in the AP plane.

Within this optimization process, a constant magnitude of adjustment was performed along the vertical direction, since it was found that the small amount of vertical adjustment (within 7cm) in the AP direction did not result in any significant variability in the joint angle. However, the joint angle in the AP plane was extremely sensitive to the joint center positions in the AP direction, indicating that their accuracy would greatly affect joint angle. As such, the adjustment in the AP direction was undertaken with a small step length (0.1cm) through an iterative process. In each step of the iteration, the root mean square error (RMSE) of the modeled ankle torque and experimental ankle torque was calculated. The final hip and L5/S1 positions were those that yielded the minimum of the RMSE.

Determination of COM Position

Following the surface marker position adjustment, the whole body center of mass (COM) angle along in the sagittal plane was calculated according to the following equation: $\sum_{i=1}^{7} \frac{m_i}{M}\ com_i$

where M is the whole body mass, and m_i and com_i are the mass and center of mass location of each body segment.

$$COM_{A-P}(\theta_{foot-AP}, \theta_{shank-AP}, \ldots, \theta_{lower\ arm-AP}) = x_{MT5} + M_{foot}\ C_{foot}\ L_{foot} \cos(\theta_{foot-AP}) + M_{shank}\ (L_{foot} \cos(\theta_{foot-AP}) + C_{shank} L_{shank} \cos(\theta_{shank-AP})) + M_{thigh}\ (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + C_{thigh} L_{thigh} \cos(\theta_{thigh-AP})) + M_{trunk}\ (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + L_{thigh} \cos(\theta_{thigh-AP}) + C_{trunk} L_{trunk} \cos(\theta_{trunk-AP})) + M_{head}\ (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + L_{thigh} \cos(\theta_{thigh-AP}) + L_{trunk} \cos(\theta_{trunk-AP}) + C_{head} L_{head} \cos(\theta_{head-AP})) + M_{upper\ arm}\ (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + L_{thigh} \cos(\theta_{thigh-AP}) + L_{trunk} \cos(\theta_{trunk-AP}) + C_{upper\ arm} L_{upper\ arm} \cos(\theta_{upper\ arm-AP})) + M_{lower\ arm}\ (L_{foot} \cos(\theta_{foot-AP}) + L_{shank} \cos(\theta_{shank-AP}) + L_{thigh} \cos(\theta_{thigh-AP}) + L_{trunk} \cos(\theta_{trunk-AP}) + L_{upper\ arm} \cos(\theta_{upper\ arm-AP}) + C_{lower\ arm} L_{lower\ arm} \cos(\theta_{lower\ arm-AP}))$$

Chapter 5 A New Method to Assess Passive and Active Ankle Stiffness during Quiet Upright Stance

Abstract

Both passive and active ankle torque contribute to postural stability during quiet upright stance, yet directly measuring their relative contributions is difficult. In this research, a new method was developed to estimate passive and active ankle stiffness (ST) and damping (DA). In contrast to earlier approaches, the proposed method does not require external mechanical or sensory perturbations. Instead, the method is based on the assumption that upright stance is intermittently controlled, and that active ankle torque is in-phase coherent with ankle angular acceleration. Thus, identifying the local maxima of ankle angular accelerations facilitates the identification of time windows that include substantial active ankle torque. After identifying these local maxima and associated windows, estimates of passive and active ankle ST and DA were obtained using linear regression analyses. Influences of age and gender on estimated passive and active ankle ST and DA were assessed, as well as the effects of localized muscle fatigue at the ankle. Consistent with earlier work, passive ankle torque was estimated to account for 94 - 97% of the total ankle torque, and to have linear relationships with the ankle angle and angular velocity. Older adults and males both had larger active and passive ankle ST and DA. Fatigue influenced active ankle torque, particularly among young adults and males. In conclusion, a new approach was developed to assess passive and active torque contributions, and which appears to be a promising tool to investigate individual differences and task-related effects such as fatigue.

5.1 Introduction

Control of balance during upright stance is achieved through corrective torques at multiple joints in the body (Cholewicki et al., 1997, Vuillerme et al., 2002, Schieppati et al., 2003, Gribble and Hertel, 2004, Kiemel et al., 2008), and which are generated through intrinsic passive, reflexive, and active voluntary muscular contractions. Among these, passive ankle torque is associated with viscoelastic characteristics of muscle, tendon, aponeurosis, and joint connective tissues, when these are stretched in the absence of reflex or voluntary recruitment of muscle units (Loram and Lakie, 2002a). Active ankle torque is generated through two mechanisms, active intrinsic stiffness and reflex stiffness (Sinkjær and Magnussen, 1994, Mirbagheri et al., 2001, Galiana et al., 2005). Specifically, active intrinsic stiffness is developed through the recruitment of muscle fibers through feedback and/or feed forward pathways (Fitzpatrick et al., 1994). Reflex stiffness is presumably activated through spinal or central reflex pathways that innervate muscle fibers during upright stance (Moorhouse and Granata, 2007).

Both passive and active components of ankle torque are responsible for the control of upright stance and postural stability. Previous muscle modeling utilized to simulate quiet upright stance suggested that passive ankle torque alone might be insufficient to maintain upright stance (Morasso and Sanguineti, 2002). Because the passive ankle torque is smaller than the gravitational torque, the use of only passive ankle torque can lead to a significant and impractical time delay in position control and thus increase instability, rendering the need for active control (Lakie et al., 2003). A number of studies have also confirmed the existence of active control torque during upright stance (Loram and Lakie, 2002a, Loram et al., 2005b). In terms of reflex activity, it occurs in response to tonic stretch due to the lengthening of a muscle during sway (Mirbagheri et al., 2000). However, this reflex activity remains quite difficult to measure, due to the minimal muscular activity associated with quiet upright stance (Fitzpatrick, 2003). In fact, ankle reflex activities have not been detected for small perturbations during quiet upright stance (Loram and Lakie, 2002a).

Research efforts have been made to quantify ankle stiffness and damping. Ankle stiffness has been measured in different postures, mostly involving the use of external mechanical or sensory perturbations. For instance, studies have employed supine postures—with plantar-flexion and dorsiflexion ankle motion—subjected to sequential and random perturbations, wherein passive, active, and reflexive ankle stiffness was measured (Kearney et al., 1997, Galiana et al., 2005). Previous work also measured passive ankle stiffness in a seated posture, using an ankle robot for generating plantar-flexion/dorsiflexion and inversion/eversion ankle motions (Roy et al., 2011). Passive and active ankle stiffness has also been measured in upright stance postures: 1) standing on a rotating force plate with very small bi-phasic random rotational perturbations (Loram and Lakie, 2002a); 2) standing on a rotating platform, with a relatively larger random rotational perturbations (Casadio et al., 2005); 3) standing on the ground, with sum-of-sines functional visual scene perturbations applied (Kiemel et al., 2008).

It is evident that most of these studies have involved the use of sequential or random perturbations. The use of such perturbations to estimate passive, active, and reflex muscular stiffness can ensure separation of the passive and active components from reflexive aspects of corrective ankle torque. However, ankle stiffness is dependent on its operating states and the magnitude of the external mechanical perturbations (Weiss et al., 1988, Latash and Zatsiorsky, 1993, Kearney et al., 1997, Loram and Lakie, 2002a). Active ankle stiffness measured with perturbations is larger than what has been obtained without using perturbations (i.e., during quiet upright stance scenarios). This difference is attributed to the fact that during perturbations the active intrinsic and reflex pathways constitute a greater percentage of the total joint stiffness (Mirbagheri et al., 2000), and much larger muscular responses are elicited than in quiet upright stance conditions (Loram et al., 2004, Casadio et al., 2005).

As noted, several existing studies have applied somewhat substantial perturbations to upright stance, which induced large active intrinsic corrective torques and likely led to unreliable estimates of joint stiffness during quiet upright stance. To resolve this, other researchers have applied very small (i.e., 0.05 degree, 1 degree/s) impulse perturbations to upright stance (Loram and Lakie, 2002a). These magnitudes were considered to be close to the magnitude of noise naturally present during quiet upright stance. Regardless of the magnitude, the differences between small perturbations versus non-perturbations on joint corrective torques are not well understood. The former are likely to induce elevated voluntary muscle activities, which hence could lead to imprecise estimates of actual joint stiffness and damping during quiet upright stance. As such, an identification method that does not use perturbation is considered most appropriate for estimating passive and active ankle stiffness during quiet upright stance.

Localized muscle fatigue (LMF) affects multiple aspects of the neuromuscular system, including: muscle relaxation rate, motor neuron discharge rate, excitation-contraction coupling, metabolic pathways, and force capacity (Enoka and Stuart, 1992). Most of these physiological changes are related to muscular mechanical performance, stiffness, and damping (Enoka and Stuart, 1992). For example, muscle reflex, contraction, and co-contraction level influence trunk stiffness (Granata and Marras, 1995, Granata and Marras, 2000, Moorhouse and Granata, 2007). Following repeated stretch-shortening cycle exercises, the peak eccentric stiffness of the soleus muscle declined (Avela and Komi, 1998). Similarly, ankle and knee joint stiffness also decreased following exhausting stretch-shortening cycle exercises (Kuitunen et al., 2002). In contrast, a prior study found that joint stiffness, damping, and inertia remain unchanged following LMF (Weiss et al., 1988). Nevertheless, to our knowledge, no prior studies have been specifically dedicated to addressing whether joint stiffness and damping change due to LMF during quiet upright stance, which is one goal of our research.

Aging has adverse to detrimental effects on muscular mechanical properties, inducing muscle atrophy, reduction of muscle fiber size and myosin function, reduced muscle mass/volume and cross sectional area, and decreased contractile capacity (Grimby and Saltin, 1983, Bruce et al., 1989, Akagi et al., 2009, Canepari et al., 2010). Aging also causes motor neuron changes leading to decreased motor unit discharge rates and increased motor unit discharge variability, decreased lumbosacral spinal cord motor neurons and numbers of motor neurons (Roos et al., 1997, Doherty, 2003). Muscle mechanical performance and motor neurons are associated with passive and active joint stiffness. As such, joint stiffness could change with age. Indeed, it was demonstrated that older adults have larger passive joint stiffness during walking (Silder et al., 2008). Due to loss of general muscle strength during locomotion, the active control of plantar flexors was compromised, inducing an increase of passive hip stiffness among older adults (Goldberg and Neptune, 2007). Despite these studies, age-related changes in passive and active ankle stiffness and damping were not well studied and still need to be investigated, specifically for the quiet upright stance condition and considering the effects of ankle LMF.

The primary purpose of this work was to create a method to estimate passive and active ankle stiffness and damping during quiet upright stance without the use of external perturbations. Secondary to this main goal, this new method was used to assess the effects of aging, gender, and ankle LMF on joint stiffness, damping, and inertia. The method was motivated by some existing studies suggesting that quiet upright stance is under intermittent control (Loram and Lakie, 2002b, Loram et al., 2005b, Loram et al., 2011). Based on the theory of intermittent control of upright stance, and our earlier work (Chapter 4), we developed a new algorithm to separate the passive and active components of active ankle torque.

5.2 Methods

5.2.1 Experimental and Modeled Ankle Torque

The experimental data used in this study were obtained from a previous study (Lin et al., 2009), in which trials of quiet upright stance were completed; relevant procedures were provided in Chapter 4. These data included whole-body kinematics and ground reactions forces. Using the model described below, an estimate of ankle torque, in plantar/dorsiflexion, was calculated with a three-segment model (leg, torso, and head). The modeled ankle torque was calculated using a top-down inverse dynamics approach, where the ankle, trunk, and head angles were estimated based on reflective markers in the existing dataset (Figure 5.1 – A; details in Appendix A). Actual (experimental) ankle torque was used to validate the modeled ankle torque, and was determined using a 2D model (Figure 5.1 - B). In this latter model, foot mass and center of mass location were obtained from existing anthropometric data (Enoka, 2008). Center of pressure location was determined from ground reaction forces obtained from a force platform. The experimental ankle torque was solely used for validating modeled ankle torque, and the modeled ankle torque was involved in the calculation of passive and active ankle stiffness. As such, estimates of passive and active ankle stiffness were based on a three segment model considered able to reasonably represent upright stance (Hsu et al., 2007, Kiemel et al., 2008).

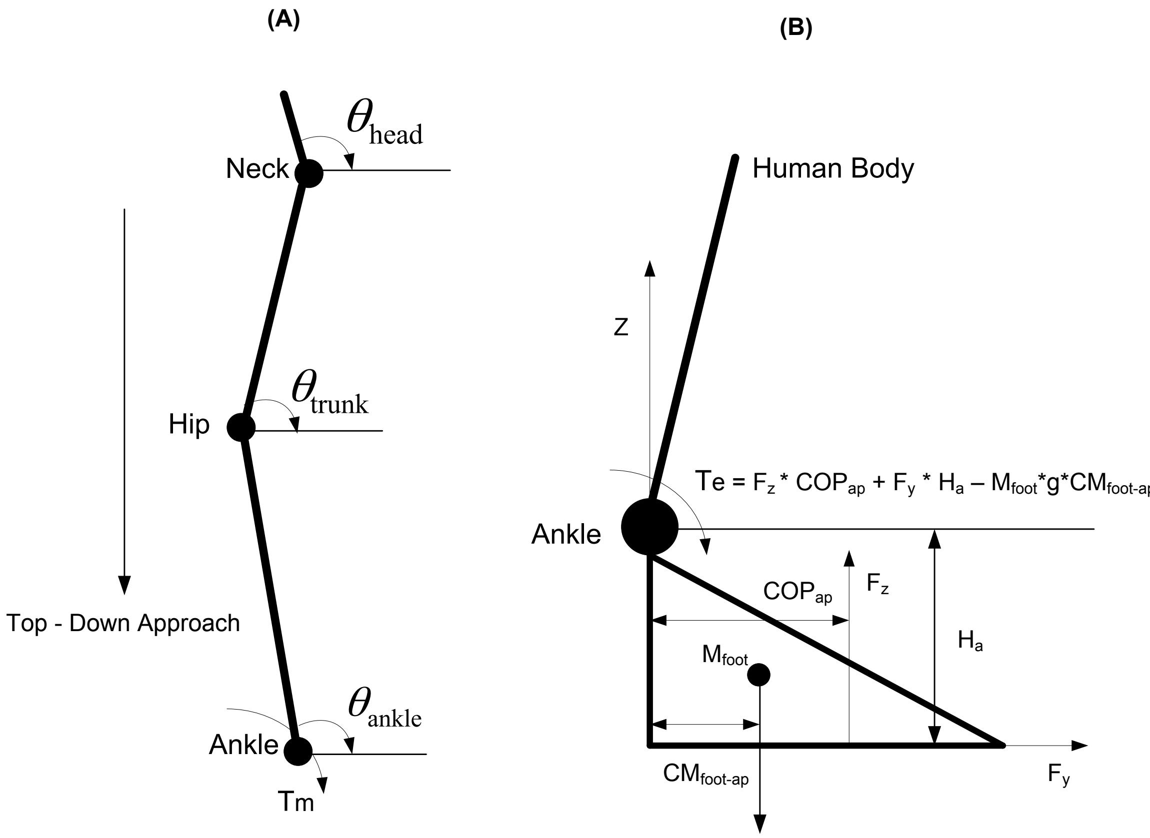

Figure 5.1: Approach for determining modeled and experimental ankle torques, (A) T_m: modeled ankle torque, $\boldsymbol{\theta_{ankle}}$: ankle angle, $\boldsymbol{\theta_{trunk}}$: trunk angle, $\boldsymbol{\theta_{head}}$: head angle; (B) T_e: experimental ankle torque, F_y: horizontal ground reaction force, F_z: vertical ground reaction force, COP_{ap}: COP location in the AP direction, H_a: ankle height, M_{foot}: mass of foot, CM_{foot}: center of mass of foot.

5.2.2 Passive and Active Zones and Ankle Torque

Previous studies indicate that quiet upright stance is controlled intermittently (Lakie et al., 2003, Loram et al., 2005b), and which involves only intermittent "bursts" of active (muscular) control of upright stance. In the absence of active control, there are passive contributions, due to tissue stretch and/or deformation. In turn, this intermittency in active control implies that distinct "zones" or time windows should exist, which involve either passive (P) or passive + active (P+A) control. While they may co-exist, the respective lengths of the passive and active zones are different. Active zones only exist when intermittent muscle bursts are present. In contrast, passive zones exist throughout a period of postural sway, excepting those situations near a neutral joint configuration. In P zones, while some active ankle torque might be present, the magnitude is likely small relative to in the P+A zone. This latter assumption is supported by results from Loram and Lakie (2002b), indicating that the soleus muscle contracts intermittently and no that major muscle activities could be detected between these intermittent episodes.

Because active zones involve bursts of muscle activity, one approach is to measure such activity directly. In the current method, an alternative method was used to identify the P+A zones. Based on our earlier results using a sliding mode control model (Chapter 4), active ankle torque (T_a) is strongly in-phase coherent with ankle angular acceleration ($A_{a\text{-}ac}$), as demonstrated in Figure 5.2. Thus, local maxima of active ankle torque — associated with muscle activity — are coincident with local maxima of $A_{a\text{-}ac}$. By identifying local maxima in the absolute values of $A_{a\text{-}ac}$, then the temporal centers of P+A zones can be identified, and these zones should range (temporally) on both sides of the local maxima.

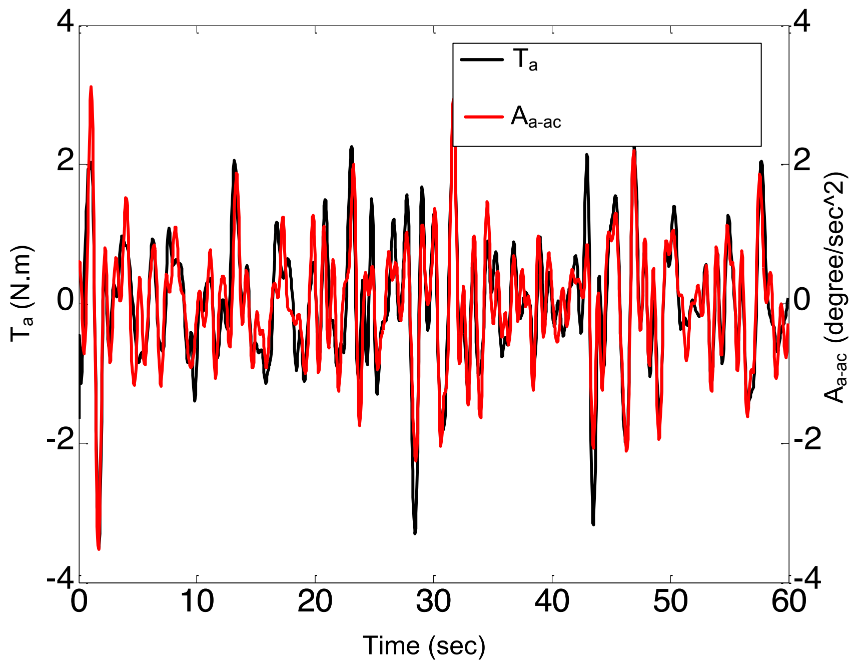

Figure 5.2: A sample of model-estimated active ankle torque (T_a) and ankle angular acceleration ($A_{a\text{-}ac}$).

Three steps were used to determine, in each trial, the temporal location and size of the P+A zones (Figure 5.3). First, as noted earlier, the center points of each P+A zone were identified using local maxima of absolute values of $A_{a\text{-}ac}$ time series. Second, the size (duration) of each P+A zone was determined. As the experimental data were sampled at 20Hz, the interval between data points is 50ms. As such, the size of a P+A window can only be in increments of 100 ms. From earlier work (Loram and Lakie, 2002b, Lakie et al., 2003, Loram et al., 2005b, Loram et al., 2011), the time interval between soleus muscle contractions during upright stance is 383 ± 55 ms. To yield comparable intervals here (i.e., 200 – 400 ms), the P+A window size was set to 200 ms (see below for a sensitivity analysis). Third, and given the locations/durations of the P+A zone, the P zones were simply the time windows adjacent to P+A zones.

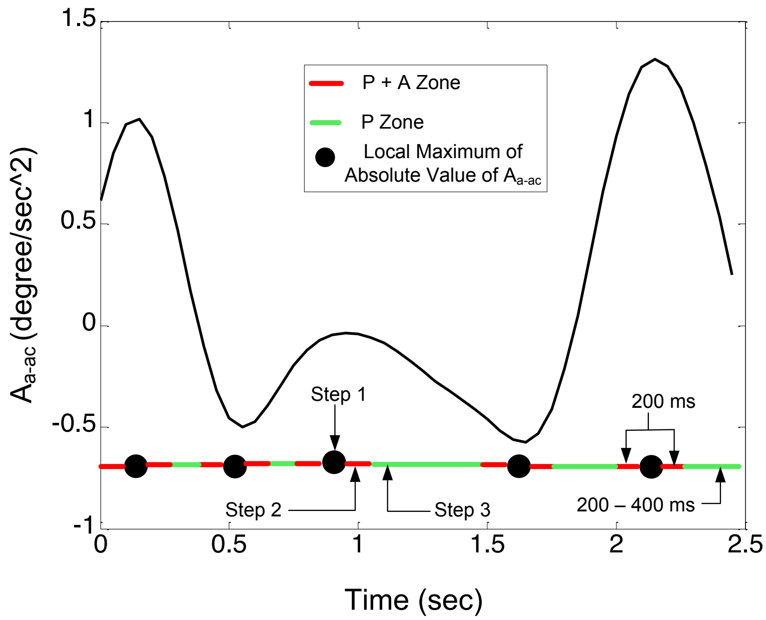

Figure 5.3: Steps used to determine the local maxima, passive (P) zones, and passive + active (P+A) zones. The P+A zone size is fixed in size (200 ms), yet the P zone sizes vary. Step 1: identify the local maxima of absolute values of ankle angular acceleration ($A_{a\text{-}ac}$); Step 2: Determine the locations of P+A zones; Step 3: determine the locations of P zones.

After identifying the locations of the P and P+A zones, these zones were mapped to ankle torque. Specifically, passive ankle torque was extracted from the P zones, and combined passive plus active ankle torque was extracted from the P+A zones.

5.2.3 Passive and Active Ankle Stiffness and Damping

Following the procedures demonstrated above, numerous P zones, P+A zones, and passive and active ankle torques are available. It is thus possible to undertake regression analyses between ankle kinematics and kinetics, and thereby estimate stiffness and damping contributions to ankle torque. Preliminary analysis showed that linear regression models were appropriate. In P zones, there is only passive ankle torque, and passive ankle torque (T_p) has the following relationship with ankle angle and angular velocity:

$$T_p = k_p \, \theta_{ankle} + m_P \dot{\theta}_{ankle} \quad (1)$$

In P+A zones, there are both passive (T_p) and active (T_a) ankle torque contributions, thus:

$T_{p+a} = T_p + T_a$

(2)

with T_a represented as:

$$T_a = \quad k_{a-ankle} \theta_{ankle} + +k_{a-trunk} \theta_{trunk} + m_a \dot{\theta}_{ankle} + I \ddot{\theta}_{ankle} \quad (3)$$

where

T_p: Passive ankle torque in P zones

T_a: Active ankle torque in P+A zones

T_{p+a}: Ankle torque in P+A zones (passive + active ankle torque)

θ_{ankle}: Ankle angle, $\dot{\theta}_{ankle}$: Ankle angular velocity,$\ddot{\theta}$: Ankle angular acceleration

θ_{trunk} : Trunk angle

k_p: Passive ankle stiffness

$k_{a-ankle}$: Active ankle stiffness related to ankle angle

$k_{a-trunk}$: Active ankle stiffness related to trunk angle

m_p: Passive ankle damping

m_a : Active ankle damping

I: leg moment of inertia

For linear regression analyses, the Matlab *regress* function was used and R^2 and p values were obtained for evaluation of model fits. The specific predictor variables in the equations above were determined using an iterative explorative method. For both passive and active ankle torque equations (equations 1 and 3), the first step included only ankle angle and ankle angular velocity. For passive ankle torque (equation 1), inclusion of only these two predictors was sufficient to obtain significance of the model ($p < 0.05$) and high R^2 values (>0.9). For active ankle torque (equation 3), using only the same two predictors was not effective ($R^2 < 0.01$, p>0.05). Adding trunk angle was more effective ($R^2 \sim 0.3$ - 0.6, p<0.05) as was adding ankle angular acceleration ($R^2 \sim 0.5 – 0.8$, p<0.05), and thus these two were included in the model. Addition of other predictors and interactive effects, such as $\theta_{ankle}\times\theta_{trunk}$, $\dot{\theta}_{ankle}\times\dot{\theta}_{trunk}$ θ_{head}, $\dot{\theta}_{head}$, $\dot{\theta}_{trunk}$, did not substantially improve the model predictions. In contrast, removal of any of the other predictors in the equations above substantially impaired model predictions.

To verify the P+A zone identification process described above, a sensitivity analysis was conducted to determine the effects of assumed P+A zone size on passive and active ankle stiffness and damping, and leg inertia. As noted earlier, the P+A zone sizes start from 100ms and can increase in 100 ms increments (given the data available). Here, there are only three possible choices of the time window (i.e. 100, 200, and 300ms) that still allow for reasonable time intervals between muscle bursts (P+A zones). None of the noted variables were highly sensitive to the specific time window size in this range (Figure 5.4). As an additional assessment, specifically of the intermittency of modeled postural control, the mean time interval between P+A zones was obtained in each trial (see Results).

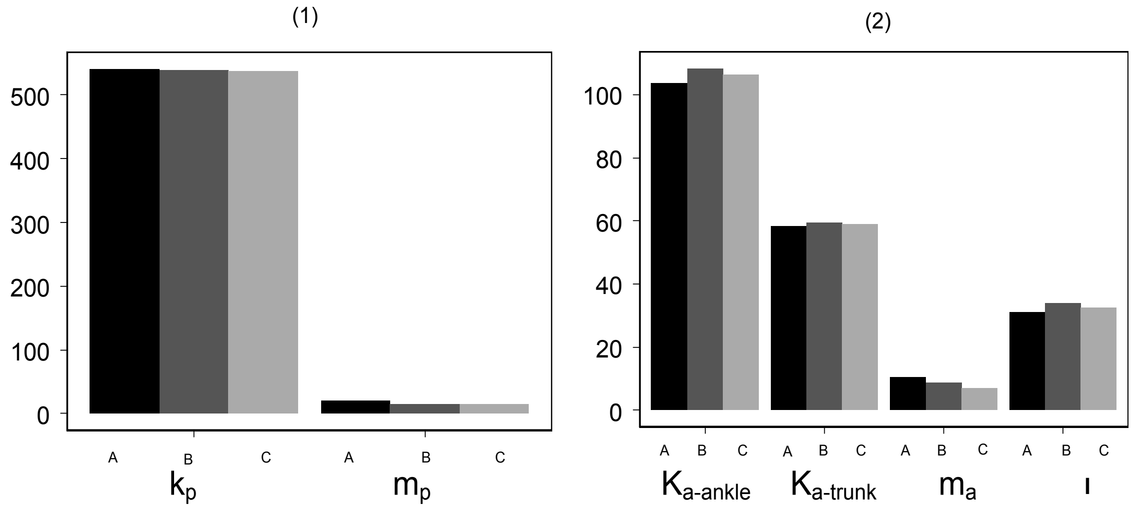

Figure 5.4: The influences of assumed P+A time window size (A:100ms, B:200ms, C:300ms) on estimated passive ankle stiffness－k_p (Nm/rad) and damping－m_p (Nms/rad) (1), and active ankle stiffness－$k_{a\text{-ankle}}$, $k_{a\text{-trunk}}$, ankle damping－m_a, and leg inertia－I (N.m.s^2/rad) for a representative trial.

5.2.4 Analysis

To evaluate the current approach, root mean square errors (RMSE) and coefficients of determination (R^2) were obtained between modeled and experimental ankle torques in each trial. Also for each trial, R^2 was obtained to assessing the quality of the fitted passive and active ankle torque regression equations (Eqs 1 and 3). Two groups of dependent variables were obtained: 1) passive ankle torque, stiffness, and damping of the ankle; and 2) active ankle torque, stiffness, damping, and moment inertia of the leg. Respective means of the dependent variables were obtained for the three pre-fatigue and three post-fatigue trials, and these means were used subsequently in statistical analyses. Effects of age and gender were evaluated in pre-fatigue trials using separate two-way ANOVAs. Differential effects of fatigue associated with age and gender were analyzed using separate two-way ANCOVAs on change scores (post-fatigue – pre-fatigue) for each dependent variable, with independent variables of age and gender, and the pre-fatigue measure as a covariate. The level of significance for all tests was set at $p < 0.05$, and all statistical analyses were performed using JMP 9.02 (SAS Inc., Cary, NC, USA).

5.3 Results

5.3.1 Representative Trial

Both modeled and experimental total ankle torque are shown in Figure 5.5 for a representative trial. In this, the torque values represent the summed contribution from both ankles. A small difference is evident between modeled and experimental ankle torques (RMSE = 1.1 Nm, $R^2 = 0.93$).

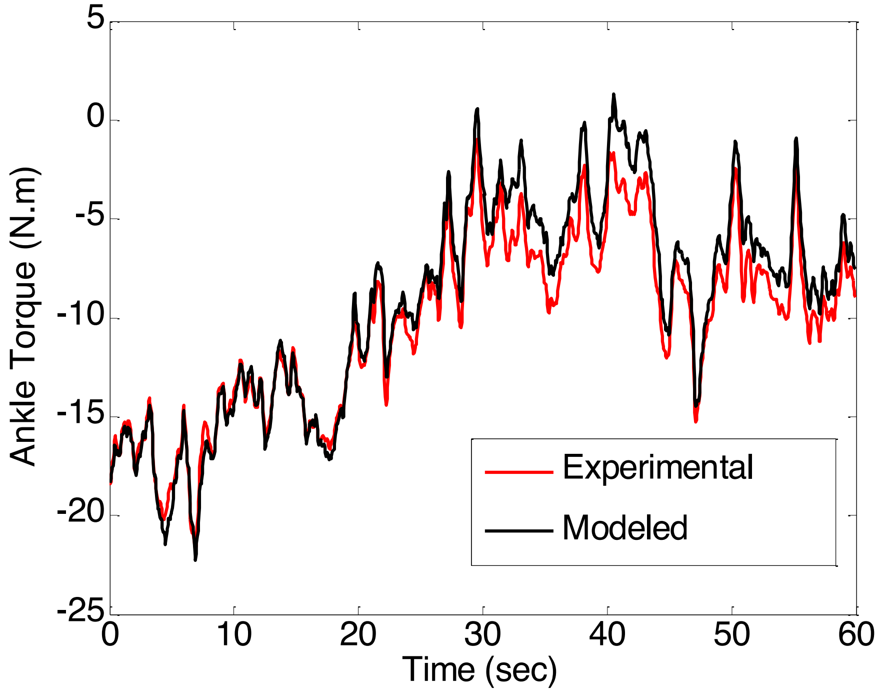

Figure 5.5: Modeled and experimental ankle torque for a representative trial.

Ankle A-AC, ankle torque, and the passive and P + A zones for the same representative trial are shown in Figure 5.6. It is evident that all of local maxima have been identified as indicated by the assistant curve (top Figure). The distribution of the local maxima is also fairly uniform through the duration shown. The mean time interval between the P+A zones (muscle bursts) was 545ms for this trial.

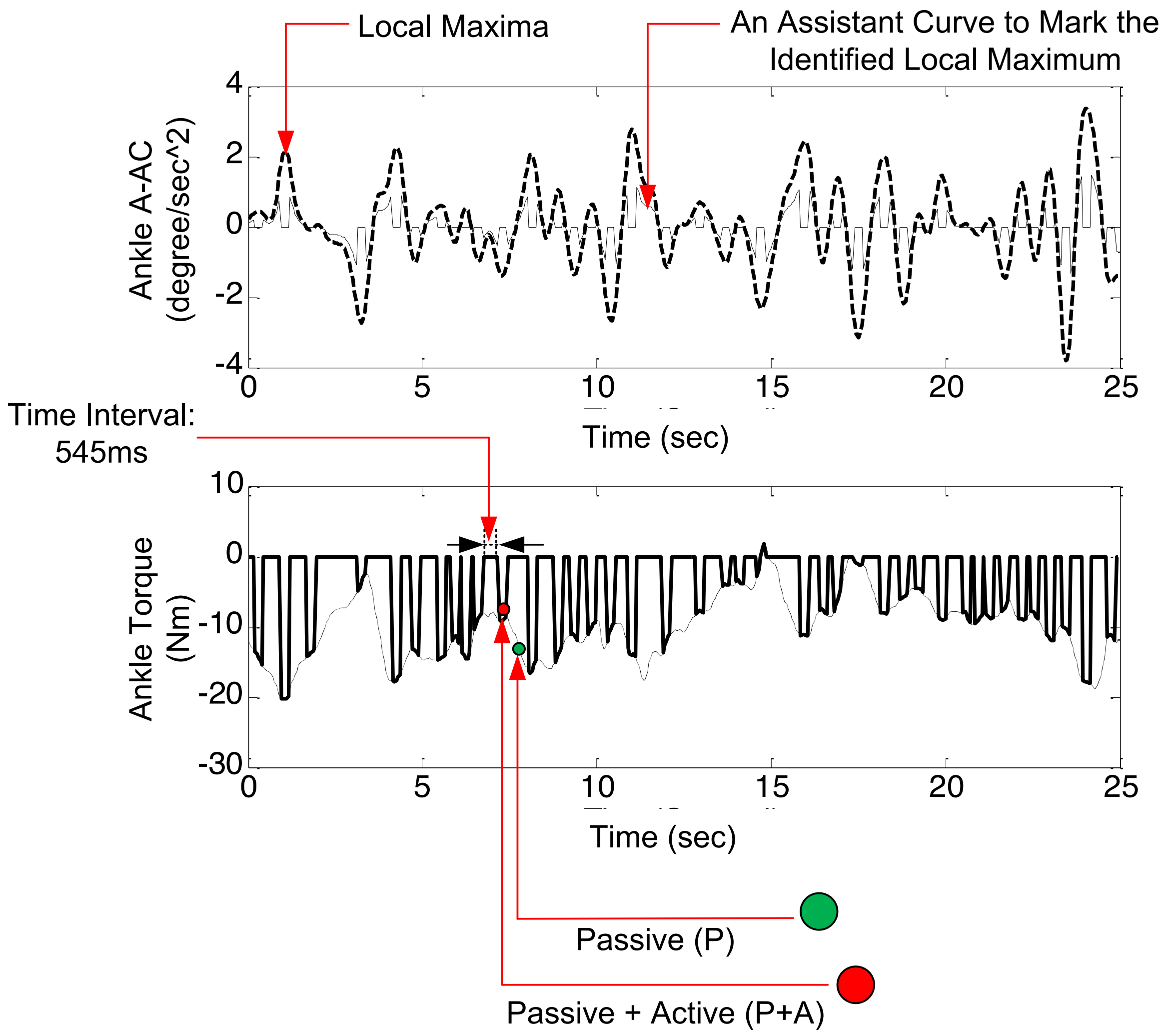

Figure 5.6: Passive and passive + active zones on ankle angular acceleration and ankle torque for a representative trial. (Time interval refers to the mean time interval for the trial).

The relationships between ankle angle and passive ankle torque, as well as between ankle angular acceleration and active ankle torque, are shown in Figure 5.7. A linear relationship between passive ankle torque and ankle angle is evident, with a strong correspondence between the two measures. Some linear correspondence also exists between ankle angular acceleration and active ankle torque. The low correspondence, though, indicates that other predictors are needed (e.g., ankle angle and angular velocity and trunk angle). Similar outcomes were found in the majority of trials.

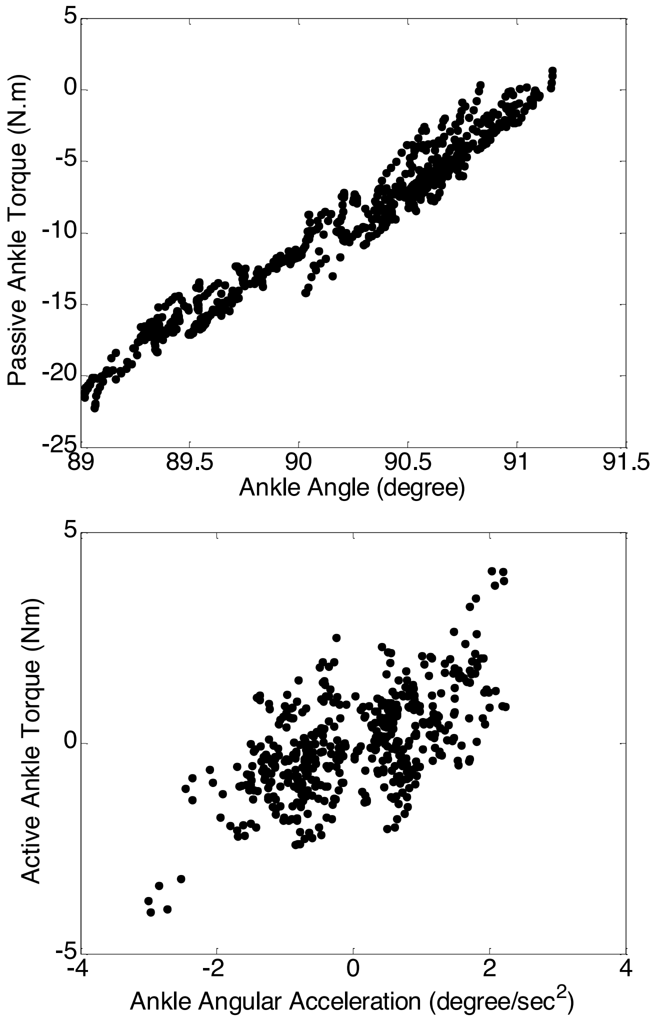

Figure 5.7: Relationships between ankle angle and passive ankle torque, and ankle angular acceleration and active ankle torque, for a representative trial.

For the same representative trial, both experimental and predicted passive and active ankle torques are presented in Figure 5.8, where the latter were obtained from the prediction equations (1 and 3). In this trial, a high correspondence was found between the two sets of passive torques, and a somewhat weaker correspondence for active torques.

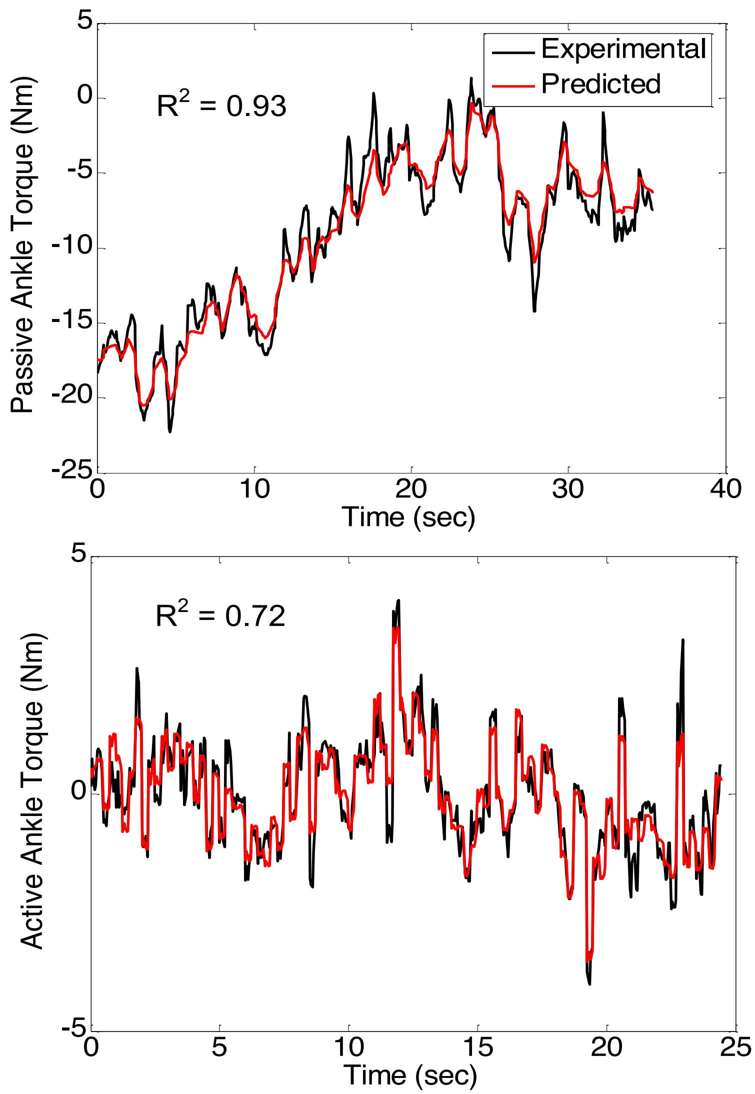

Figure 5.8: Experimental and predicted passive (top) and active (bottom) ankle torques for a representative trial.

5.3.2 Model Performance

Across all trials, the mean (SD) coefficient of determination R^2 between experimental and modeled ankle torques was 0.89 (0.25). Root mean square errors (RMSE) between modeled and experimental ankle torques were 2.75 (2.12) Nm (3 -10% of overall ankle torque). Coefficients of determination R^2 from passive and active ankle torque model fits were 0.91 (0.08) and 0.61 (0.15), respectively. The time intervals between identified P+A zones was 548(27) ms.

5.3.3 Statistical Results

A summary of statistical results, regarding the effects of age, gender, and fatigue, is presented in Table 5.1. These results are presented in more detail below, separately for pre-fatigue trials and regarding the effects of fatigue on passive stiffness, passive damping, passive torque, active stiffness (ankle contribution), active stiffness (trunk contribution), active damping, active torque, and leg inertia.

Table 5.1: Summary of statistical results. For each dependent measure, summary statistics are given for pre- and post-fatigue trials, along with the effects of age (A) and gender (G) on pre-fatigue measures and fatigue-induced changes. Significant ($p < 0.05$) effects are bolded, and those approaching significance ($p < 0.06$) are underlined. k_p: passive stiffness, k_a: active stiffness, T_p: passive torque, $k_{a\text{-ankle}}$: active stiffness (ankle contribution), $k_{a\text{-trunk}}$: active stiffness (trunk contribution), m_a: active damping, T_a: active torque, I: moment inertia of the leg.

		Pre-Fatigue				Fatigue			
		Mean(Std)	A	G	AXG	A	G	AXG	Mean(Std)
Passive	k_p (N.m/rad)	661(199)	**0.0087**	**<0.0001**	0.29	0.39	0.65	0.46	669(199)
	m_p (N.m.s/rad)	27.7(28.0)	**0.010**	**0.0073**	0.64	0.17	0.79	0.12	31.5(35.8)
	T_p (N.m)	33.8(13.4)	0.58	0.78	**0.016**	0.66	0.30	0.56	**31.3(13.4)**
Active	$k_{a\text{-ankle}}$ (N.m/rad)	102(44.8)	**0.0007**	**0.0005**	0.079	0.09	0.50	0.057	97.9(40.3)
	$k_{a\text{-trunk}}$ (N.m/rad)	104(83.8)	0.52	**<0.0001**	0.36	0.49	0.81	0.13	106(79.5)
	m_a (N.m.s/rad)	19.2(19.4)	0.088	**0.0027**	0.52	0.23	0.92	0.13	24.3(23.2)
	I (N.m.s^2/rad)	29.9(18.8)	0.051	**0.0017**	0.18	0.72	0.67	0.72	28.6(18.2)
	T_a (N.m)	1.07(0.64)	**0.003**	**0.0004**	**0.0206**	**0.033**	0.33	**0.013**	1.16(0.63)

5.3.4 Pre-Fatigue

Age and gender had significant main and/or interactive effects on all of the dependent measures, with specific quantitative differences summarized in Table 5.2. Passive ankle stiffness and damping were both larger among older adults and among males. Passive ankle torque was substantially higher among older vs. young males, with a smaller and opposite age effect evident when comparing between older and young females. Active ankle stiffness (ankle contribution) was larger among older adults. Males had higher values of active ankle stiffness (both ankle and trunk contributions), damping, and leg inertia. Active ankle torque was higher among males in both age groups, though this gender difference was more substantial in the older group. Further, both young and older females had comparable values, whereas older males had much larger values than young males. Though only approaching significance, leg inertia was higher among older vs. young adults (p = 0.051).

Table 5.2: Summary of age and gender differences in the dependent measures (see Table 5.1 legend). Only significant pre-fatigue differences (between age groups and genders) are presented.

	Passive		**Active**				
	k_p (N.m/rad)	m_p (N.m.s/rad)	$K_{a\text{-ankle}}$ (N.m/rad)	$K_{a\text{-trunk}}$ (N.m/rad)	m_a (N.m.s/rad)	I (N.m.s^2/rad)	T_a (N.m)
Young	589 (139)	20.8 (14.3)	97.4 (33.5)				0.81 (0.31)
Older	732 (224)	34.6 (35.9)	105 (53.4)				1.33 (0.78)
Male	775 (224)	35.5 (32.0)	129 (44.2)	137 (98.9)	26.1 (20.9)	36.3 (20.9)	1.34 (0.66)
Female	555 (224)	20.6 (21.8)	75.3 (25.8)	72.9 (50.9)	12.9 (15.5)	23.9 (14.4)	0.82 (0.53)

5.3.5 Effects of Fatigue

Only passive ankle torque was significantly affected by fatigue, and which decreased by ~7%. While not a significant fatigue effect overall, active ankle torque increased, and the magnitude of this increase was influenced by age and the age x gender interaction (Table 5.3). Fatigue led to an increased active ankle torque within the young group and a slightly lower active torque in the older group. Within age groups, the effects of fatigue were divergent between genders. While only approaching significance, the effects of fatigue on active ankle stiffness (ankle contribution) was more pronounced among males in comparison to females, and the divergence is more substantial in the older group ($p = 0.057$).

Table 5.3: Active ankle torque during pre- and post-fatigue trials.

	Active Ankle Torque (Nm)				
	Pre Fatigue	**Post Fatigue**	**Post - Pre**	**Post - Pre**	
Young	0.81 (0.31)	1.03 (0.56)	0.23 (0.36)	Male	0.39 (0.44)
				Female	0.053 (0.12)
Older	1.33 (0.78)	1.28 (0.67)	-0.052 (0.18)	Male	-0.17 (0.17)
				Female	0.068 (0.095)
Male	1.34 (0.66)	1.46 (0.61)	0.11 (0.44)		
Female	0.82 (0.53)	0.88 (0.49)	0.06 (0.11)		

5.4 Discussion

This study sought to develop a new method (model) for calculating passive and active ankle stiffness and damping during quiet upright stance. In contrast to existing methods, this new approach does not require the application of external mechanical or sensory perturbations. It was motivated primarily by experimental findings that upright stance is intermittently controlled by lower extremity muscle groups (Loram and Lakie, 2002b). Results using a sliding mode control model (Chapter 4) also contributed, and which indicated that ankle angular acceleration is in-phase coherent with active ankle torque. In this new method, several “zones” were identified, specifically involving passive or passive-plus-active contribution, and associated passive and active ankle torques. Subsequently, passive and active ankle stiffness and damping were calculated by fitting regression models between ankle kinematics and kinetics.

The proposed method requires identifying the passive and active zones to determine the passive and active ankle torques. As indicated from the results, the local maxima of the ankle angular acceleration were identified with acceptable accuracy, i.e. majority of local maximum values of the ankle angular acceleration have been identified (Figure 5.7, top). Across trials, the time interval between P+A zones, which corresponds to the intermittent voluntary muscle contraction interval during quiet upright stance, was ~ 540 - 550 ms. This value is larger than, but comparable to, previous estimates indicating that the soleus muscle contraction time interval is about 383 ± 55 ms during quiet upright stance (Loram and Lakie, 2002b, Lakie et al., 2003, Loram et al., 2005a, Loram et al., 2011). Considering that there is a time delay between muscle contraction and the resultant joint motion, 540 - 550 ms may represent an acceptable intermittent control time interval. The difference between sets of values indicates that the time delay between voluntary muscle contraction and the joint motion is ~ 150 ms.

Our results indicated that passive ankle torque accounted for ~ 94-97% of total ankle torque. This estimate is similar to (although larger than) prior results indicating that passive torque accounts for ~ 91 ± 23% of critical toppling torque (Loram and Lakie, 2002a). The current estimate also exceeds two other reported estimates of 64 ± 8% (Casadio et al., 2005) and 50 – 60% (Kiemel et al., 2008), and which are explained as follows. As described by Casadio et al. (2005), larger mechanical perturbations induced greater active muscular responses, thereby increasing muscular activity and associated active ankle torque. For example, 0.055° and 1° perturbations were used by Loram and Lakie (2002a), and Casadio et al. (2005), respectively. The 0.055° perturbation was considered to be very small, with a magnitude close to that of hemodynamic noise during quiet upright stance; as a result, the measured passive ankle stiffness was larger. Kiemel et al. (2008) applied visual scene perturbations to quiet upright standing, which induced relatively large leg and trunk muscle activity that was easily detectable through EMG. As such, it revealed an even greater active ankle torque and subsequently smaller passive ankle torque. In the current study, however, no perturbations were employed, which in turn likely resulted in a larger passive ankle torque than prior studies, but also likely represented passive ankle torque more validly.

Both passive and active ankle torque contributed toward maintaining upright stance. Here, passive ankle torque had peaks up to 17 Nm (single ankle contribution) and active ankle torque was ~ 1 - 1.5 Nm. The question emerges as to why such a small magnitude of active ankle torque is important for upright stance stability control. Due to the very small, slow, and bounded sway motion, active torque (even small) is likely to be responsible for influencing the sway motion acceleration direction, thereby altering the motion direction. This rationale has already been demonstrated (and supported) by the GLP algorithm (Zatsiorsky and King, 1997), which essentially viewed upright stance motion as a series of accelerating and braking segments. Furthermore, quiet upright stance has also been decomposed into a series of ballistic throw-and-catch motions, consisting of small accelerating and braking motion components (Loram and Lakie, 2002b). As such, during quiet upright stance passive joint torque is used primarily to counteract the body's gravity toppling torque, while active joint torque is principally associated with changing the sway direction and responding to hemodynamics noise (and other small perturbations).

Results from the current study also confirmed that passive ankle torque has a linear relationship with ankle angle, and indicating the primarily stiffness-based control of upright stance as indicated by Winter et al. (2001). In contrast to this simple linear relationship, though, the relationship between active ankle torque and ankle angle was rather complex. We found that active joint torque has a linear relationship with multiple variables—namely, joint angle, angular velocity, and angular acceleration. Interestingly, we found that not only was the ankle angle involved, but the trunk angle as well. Such multiple linear relationships between active ankle torque an ankle and trunk angles suggest that active ankle torque is dependent on both ankle and trunk motion indicating the possible existence of ankle-trunk interaction and coordination. This finding may also reinforce the argument that ankle-trunk interaction/coordination is essential for maintaining quiet upright stance, as indicated by earlier studies (Creath et al., 2005, Krishnamoorthy et al., 2005, Hsu et al., 2007). The leg moment of inertia also contributed to active ankle torque. This is likely because active ankle torque has a strong correlation with ankle angular acceleration, which in turn is related to leg moment of inertia (Figure 5.2). Furthermore, previous research showed that joint torque is often governed by a second-order function of joint motion, indicating the importance of leg or trunk inertia in the active control model (Hunter and Kearney, 1982, Moorhouse and Granata, 2007).

Results from this study also confirmed that passive and active ankle torque and stiffness varies by age and gender. For example, older adults had greater passive ankle stiffness and damping, and greater active ankle torque and stiffness. The increase in passive/active ankle torque, stiffness, and damping due to aging has been noted by other studies, which have indicated that older adults have larger active ankle plantar flexor muscle stiffness (Ochala et al., 2004), greater active lower extremity stiffness (Hortobágyi and DeVita, 1999), and greater passive ankle dorsiflexor stiffness (Vandervoort et al., 1992a, Gajdosik et al., 2004). In contrast, smaller passive plantar-flexion torque (Simoneau et al., 2005), ankle tendon stiffness(Narici and Maganaris, 2006), and passive ankle torque (Gajdosik et al., 1999) have also been reported among older adults. Several considerations are offered to address these contrasting results. First, it should be noted that the decrease in ankle tendon stiffness noted by Narici and Maganaris, 2006 is not equal to the decrease in muscle tendon related stiffness, the structures of which become stiffer in the older group (Vandervoort et al., 1992a, Kubo et al., 2001). Additionally, older adults generally experience smaller ranges of ankle motion, meaning that the aging ankles are stiffer and less compliant (Vandervoort et al., 1992b, Gajdosik et al., 2004) . Second, though smaller ankle torque was observed among older adults in Gajdosik et al., (1999), it does not necessarily lead to smaller stiffness and damping. This is because ankle stiffness depends on both ankle torque and ankle angle; a smaller ankle torque can be associated with even smaller ankle angle and therefore, elevated ankle stiffness.

Regarding gender-related differences, this study showed that males had greater passive ankle stiffness and damping, active ankle torque, active ankle stiffness (both ankle and trunk contributions), active ankle damping, and leg inertia. Such gender differences in stiffness have also been identified across several lower-extremity muscle groups by prior studies. For example, females typically have smaller active hamstring and quadriceps muscle stiffness and leg stiffness than males (Granata et al., 2002), while males have greater passive and active knee flexor muscular stiffness (Blackburn et al., 2004) and larger co-contraction-based active knee torque than females (Wojtys et al., 2002). Males also have larger passive ankle dorsiflexion, and full stretch ankle torque and stiffness (Vandervoort et al., 1992a, Gajdosik et al., 2006). These gender differences correspond to injury epidemiology, for example the evidence that female athletes have a 25% greater risk of ankle sprain injury and older females accounted for over 70% of non-fatal unintentional fall related injuries during emergency department visits in 2001 (Hosea et al., 2000, Stevens and Sogolow, 2005). Though other factors can also contribute to this gender difference, this difference in injury risk may be related to the relatively smaller passive and active ankle stiffness and damping among females.

In the presence of ankle LMF, passive ankle torque decreased, indicating compromised performance related to passive ankle viscoelastic tissues, and which is also in agreement with previous work (Kuitunen et al., 2002). Apart from the noted changes in terms of passive ankle torque, a larger increase in active ankle torque was evident among young adults. Additionally, the increase with fatigue was more pronounced in young vs. older males. Such differences may be attributed to the higher percentage of fatigue resistant (Type I) muscle fibers and altered anaerobic metabolic pathways present among older adults (Chan et al., 2000, Ditor and Hicks, 2000, Baudry et al., 2007, Kent-Braun, 2009). Ankle LMF also affected active ankle torque differently between genders, with a substantially greater impact on young males than young females. Multiple factors could be associated with this gender difference in the effects of fatigue, including underlying gender differences in contractile capacity, oxidative metabolic capacity, blood flow occlusion, and central fatigue (Barry and Enoka, 2007, Paillard, 2012). The gender and age differences resulting from the fatigue highlight the need to distinguish the intervention strategies for these populations regarding the occupational fall preventions (Courtney et al., 2001, Layne and Pollack, 2004, Mitchell et al., 2010, Schwatka et al., 2012).

Several limitations associated with this study should be discussed. First, we simplified the entire body to a three-segment linkage, and which viewed the knee joint as a rigid unit. This assumption could lead to imprecise estimates of ankle torque. However, since the relative motion between the knee and ankle or trunk joints during quiet upright stance is small, the loss of precision should thereby be small. Second, ankle LMF was conducted only unilaterally; however, both ankles were averaged into one ankle joint in the three-segment inverse dynamics model. This potential limitation might introduce some bias to the ankle torque calculation. During quiet upright stance, though, the magnitude of ankle motion is small so that such bias is likely minimal. Third, the hip joint center and L5/S1 locations were determined based on existing literature, with an optimization process described in Chapter 4. The potentially imprecise estimates of these locations could introduce some errors to both the modeled ankle torque and the ankle angle, which are essential for estimating passive and active ankle torque, stiffness, and damping. Methods were used to reduce such errors, as described earlier, but such errors still likely had adverse effects on model predictions. Finally, during the iterative process of determining passive plus active zone size, the iterative step size was limited to 50ms— the time interval between sample, and which was imposed by sampling frequency of 20Hz. A higher sampling frequency would allow for a small step size, and which could improve the resolution of the P+A zone identification process.

5.5 Conclusions

Thus study sought to create a method for calculating passive and active ankle torque, stiffness, and damping without using external mechanical and sensory perturbations. A new method motivated by intermittent control theory was proposed to accomplish this goal. With the new method, passive and active ankle torques were separately estimated, after which passive and active ankle stiffness were calculated based on multiple linear regression equations. The results showed that passive ankle torque accounted for a large percentage of total ankle torque. Interestingly, we found that active ankle torque may be not solely a function of ankle angle, but dependent on trunk angle as well. Results also showed that passive and active torque, stiffness, and damping were larger among males, with more substantial divergence between genders within the older group. Following ankle LMF, a larger increase of active ankle torque among young adults and males was evident indicating the need of age and gender specific interventions for occupational fall preventions.

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Appendix A

In this section, the inverse dynamics model employed in the study is briefly presented. Only ankle, trunk, and neck joints are considered in the model. During quiet upright stance, knee joint motion is relatively small relative to ankle motion, and as such the knee joint is treated as a rigid link.

The three segment inverse dynamics free body diagram is shown in Figure 5.9.

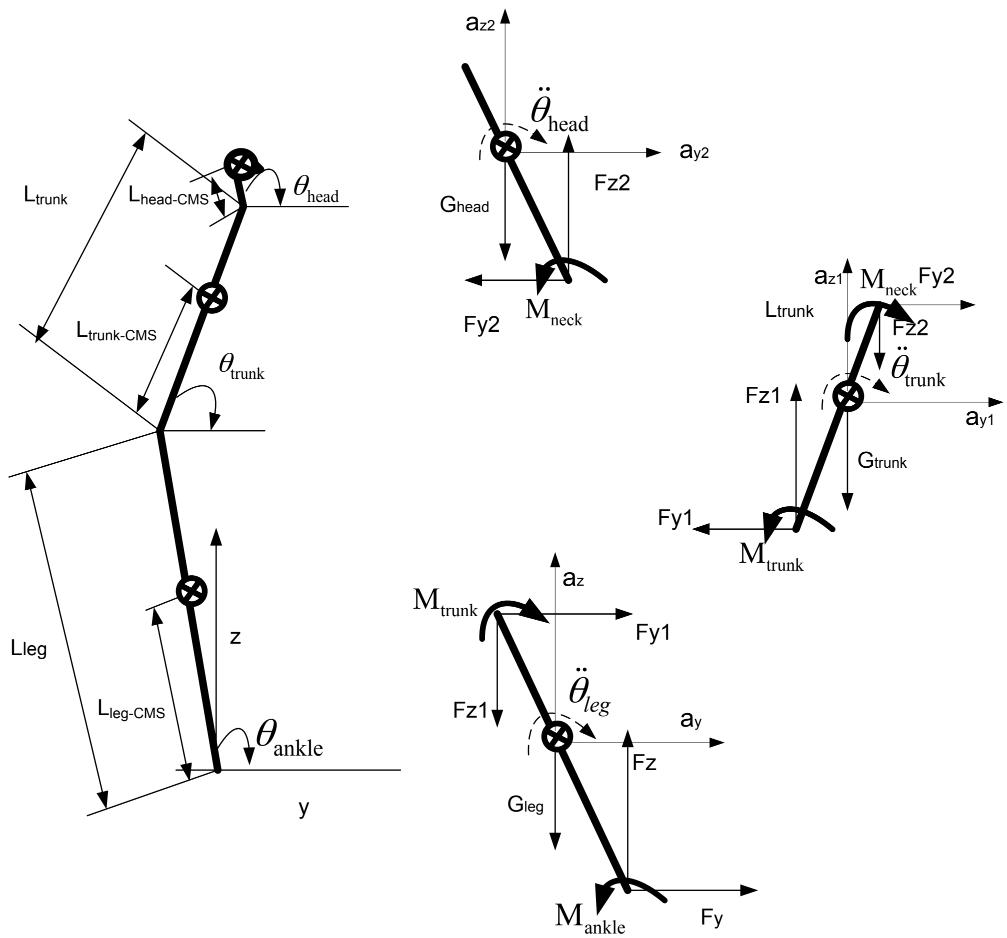

Figure 5.9 The forces and moments of a three segment inverse dynamics model.

Where the notations are listed as follows:

1.M_{ankle} is the ankle moment;M_{trunk} is the trunk moment;M_{head} is the neck moment;

2. G_{leg} is the leg weight; G_{trunk} is the leg weight; G_{head} is the neck weight;

3. F_y and F_z are ground reaction forces; F_{y1} and F_{z1} are trunk joint forces; F_{y2} and F_{z2} are neck joint forces;

4. θ_{ankle}, θ_{trunk} , θ_{neck} are the ankle angle, trunk, and neck angles;

5. $\ddot{\theta}_{leg}$, $\ddot{\theta}_{trunk}$, $\ddot{\theta}_{neck}$ are the angular acceleration of the leg, trunk, and neck;

6. a_y is the leg center of mass linear acceleration along the sagittal plane; a_z is the leg center of mass linear acceleration along the frontal plane; a_{y1} is the trunk center of mass linear acceleration along the saggital plane; a_{z1} is the trunk center of mass linear acceleration along the frontal plane; a_{y2} is the neck center of mass linear acceleration along the saggital plane; a_{z2} is the neck center of mass linear acceleration along the frontal plane;

7. L_{leg} is the leg length; L_{leg-cm} is the distance of the leg center of mass to ankle. L_{trunk} is the trunk length; $L_{trunk-cm}$ is the distance of trunk center of mass to the proximity end of the trunk.

For ankle, trunk, and neck, the following force and moment equilibrium equations are applicable.

$$\sum F_y = 0$$

$$F_{y2} + a_{y2} \times m_{head} = 0$$

$$\sum F_z = 0$$

$$F_{z2} + a_{z2} \times m_{head} + G_{head} = 0$$

$$\sum M_{sagittal} = 0$$

$$M_{neck} + I_{head} \times \ddot{\theta}_{head} + G_{head} \times L_{head-cm} = 0$$

$$\sum F_y = 0$$

$$F_{y1} + F_{y2} + a_{y1} \times m_{trunk} = 0$$

$$\sum F_z = 0$$

$$F_{z1} + F_{z2} + a_{z1} \times m_{trunk} + G_{trunk} = 0$$

$$\sum M_{saggital} = 0$$

$$M_{trunk} + I_{runk} \times \ddot{\theta}_{trunk} + G_{trunk} \times L_{trunk-cm} + L_{trunk} \times F_{y2} \times Cos(\theta_{trunk})$$

$$+ L_{trunk} \times F_{z2} \times sin(\theta_{trunk}) + M_{neck} = 0$$

$$\sum F_y = 0$$

$$F_y + F_{y1} + a_y \times m_{leg} = 0$$

$$\sum F_z = 0$$

$$F_z + F_{z1} + a_z \times m_{leg} + G_{leg} = 0$$

$$\sum M_{sagittal} = 0$$

$$M_{ankle} + M_{trunk} + I_{leg} \times \ddot{\theta}_{leg} + G_{leg} \times L_{leg-cm} + L_{leg} \times F_{y1} \times Cos(\theta_{ankle}) + L_{leg} \times F_{z1} \times sin(\theta_{ankle}) = 0$$

In these equations, the force plate forces F_y and F_z are ground reaction forces, which were obtained from the force platform reaction forces measurement. The joint angle, angular velocity, and angular acceleration as well as the segment linear acceleration are obtained from reflective marker data. The other parameters, such as the segment mass, center of mass position, and segment length, were obtained from the literature (Enoka, 2008).

A top (neck) to down (ankle) approach was employed when calculating joint forces and moments. It was assumed that by applying a top-down approach, errors in estimated neck torques could be reduced, which would otherwise be sensitive to inaccuracies in estimated trunk and leg masses.

Chapter 6 Conclusions, Limitations, and Future Research

6.1 Research Revisits

In this dissertation, three studies were conducted that investigated joint coordination, the central postural controller, and passive and active ankle stiffness components. Though separate analyses were conducted, modeling the control of upright stance was central to all three. In this work, majors goal were to help understand better the mechanisms of postural control, to identify the effects of localized muscle fatigue, gender, and aging on the postural control process, and to quantify the roles of ankle stiffness in the postural control pathway. It is expected that the results of these studies may facilitate the design of new theoretical postural control models and possibly offer some insights to fall prevention training programs. The major research conclusions, limitations, and suggested directions for future research are discussed below.

6.2 Research Conclusions

6.2.1 Two-Joint and Whole Body Coordination

Joint (or body segment) coordination is important for achieving effective postural control, or balance, in diverse daily and occupational activities. It has been suggested that the ankle, trunk, shoulder, and head are coordinated together in various postures such as upright stance and sit-to-stand movements (Scholz and Schöner, 1999, Black et al., 2007, Hsu et al., 2007). While a fundamental understanding has been gained regarding coordination patterns in time or frequency domains separately, little is known in the time-frequency domain. To address this, two-joint coordination was studied using a wavelet coherence method, with the goal to characterize the coordination patterns in both time and frequency domains.

Results from this method indicated that the wavelet coherence of ankle-knee, ankle-trunk, and ankle-head were significant over a wide frequency range (i.e., 1- 8 Hz). Furthermore, the wavelet coherence was not continuous, but rather was intermittent with a time interval of about 1 sec between instances of significant wavelet coherences. The phase relationship between pairs of joints also appeared intermittent, with both in-phase and anti-phase behaviors appearing at certain frequencies. Such an intermittent two-joint coordination pattern was identified for both adjacent joints (ankle-knee) and non-adjacent joints (ankle-head), suggesting that joint coordination is possibly governed by both mechanical constraints and certain central control mechanisms. With respect to the effects of aging, older adults displayed a larger ankle-knee time-interval in the sagittal plane, as well as a larger ankle-trunk time-interval in the frontal plane, suggesting an age-related reduction of joint coordination.

In addition to two-joint coordination analysis, whole body coordination was studied using the uncontrolled manifold (UCM) method. The shoulder, head, and whole-body center of mass (COM) were analyzed, to assess whether one or more of these might serve as the control goal(s) of whole body coordination. Both shoulder and head UCM values were greater than 1, whereas the COM UCM values were smaller than 1. Based on earlier literature, if a variable's UCM is smaller than 1 then it is unlikely that the variable is the control goal of the whole body coordination (Scholz and Schöner, 1999, Black et al., 2007). Hence, the current results suggest that the shoulder or head are more likely the control goals, as opposed to the whole-body COM. These findings differed from previous research results, which demonstrated that the whole-body COM is likely the coordination control goal (Black et al., 2007, Hsu et al., 2007). As such, further research is needed to investigate whether COM is indeed the whole body coordination control goal.

Following localized muscle fatigue (LMF) at the ankle, the intermittent time interval generally increased, indicating reduced two-joint coordination post-fatigue. This fatigue-induced decrease in coordination was especially notable in young adults. Such an age-related difference is in agreement with previous studies, which showed that exercise-induced muscle fatigue is more substantial among young vs. older adults (Chan et al., 2000, Ditor and Hicks, 2000, Kent-Braun et al., 2002). A similar outcome was also evident for the results of UCM analysis, as the values were significantly decreased following ankle LMF, and implying a post-fatigue decrease in whole body coordination.

6.2.2 Two-Joint and Whole Body Coordination: Implications

Reduced joint coordination induced by fatigue and aging may suggest a reduced ability for detecting and/or recovering from falls. In fact, the importance of joint coordination in fall prevention has also been noted by previous research. It is known, for example, that training programs that emphasize body flexibility and coordination, such as Tai Chi and dynamic walking, are useful for fall prevention (Li et al., 2005, Gregory and Watson, 2009). Hence, the current results provide some explanation and a potential basis regarding such fall prevention and post-fall rehabilitation programs among older adults.

Furthermore, since muscle fatigue also compromises whole body coordination which suggests a potential increase in postural instability following prolonged and/or physically-demanding work, occupational fall prevention programs might need to emphasize joint coordination to reduce the risks of occupational falls, especially among older workers.

6.2.3 An Intermittent Sliding Mode Postural Controller

Quiet upright stance has been modeled using continuous postural controllers (Kuo, 1995, Peterka and Loughlin, 2004, Kiemel et al., 2008). However, previous experimental have also indicated that quiet upright stance is likely maintained through intermittent rather than continuous control mechanisms (Loram and Lakie, 2002, Loram et al., 2004, 2005, Loram et al., 2011). Inspired by these experimental findings, an intermittent postural controller was constructed and applied to model human upright stance. Specifically, a single-segment inverted pendulum was used, and control of this was modeled with a sliding mode intermittent controller. The motion of the sliding mode controller was driven by an intermittent control signal, which was used to minimize errors between desired and estimated trajectories.

Based on the results obtained, the sliding mode controller can reasonably track upright stance (e.g. the COM angle, angular velocity, and angular acceleration time series). Furthermore, the model yielded estimates of required ankle torques that were consistent with the actual ankle torques derived from experimental data. The estimated ankle torques were also decomposed into passive and active components, based on whether the torque was associated with feedback control signals (passive: no feedback; active: with feedback). Overall, the estimated passive component of ankle torque accounted for 97% of the entire ankle torque, which was in a good agreement with estimates reported previously (Loram and Lakie, 2002, Casadio et al., 2005).

Following the onset of ankle LMF, an increase of the error in ankle angular acceleration tracking was evident. The reduced correlation between modeled and experimental ankle torques suggests that the control model implemented here may be insufficient to account for the effects of fatigue. Despite this, and based on the outputs from the postural control model, males and young adults may have larger increases in active ankle torque post fatigue. As suggested previously (Stevens and Sogolow, 2005, Stevens et al., 2009), such differences in active ankle torque with respect to age and gender can aid in the design targeted fall prevention training programs.

6.2.4 Intermittent Controller: Implications

This study demonstrated a relatively high tracking performance of the sliding mode controller. It is thus feasible that the sliding mode could, at least abstractly, represent the actual underlying postural controller for upright stance. Alternatively, the sliding mode controller could be comparable to the actual body controller in terms of its kinematic and kinetic performance. In addition, it appears that the passive component of ankle torque plays a critical role (97% of the entire ankle torque) in maintaining quiet upright stance. As such, fall prevention training programs may benefit from attention to improving passive musculoskeletal mechanics to enhance posture stability and reduce the risks of falls.

6.2.5 A New method for Calculating Passive and Active Ankle Torque, Stiffness, and Damping

Both passive and active ankle torques contribute to maintaining upright stance (Morasso and Sanguineti, 2002, Kiemel et al., 2008). In this study, a new method was developed to separate the passive and active ankle torque contributions. This method is novel in that it does not require complex experimental settings to deliver mechanical and (or) sensory perturbations, as was required in previous studies. Instead, it is based only on measurements of whole body kinematics and ground reaction forces during quiet upright stance.

This new method was based on the assumption of intermittent postural control, as discussed above. Active ankle torque was assumed to be developed intermittently and to exist only during contractions of skeletal muscles (Loram et al., 2004, 2005). Based on earlier modeling results using the sliding mode controller, it was also assumed that active ankle torque is coherent with ankle angular acceleration. Thus, intermittent active zones were identified from regions around local maxima of ankle angular acceleration, and estimates of passive and active ankle torque were extracted from the identified passive and active zones. Subsequently, linear regression models were used to estimate passive and active ankle stiffness from segmental kinematics and extracted torque data.

In additional to the current approach, passive and active ankle torques were also calculated using the sliding mode controller. Although quite divergent methods were used, the results were very similar, indeed almost identical between the two studies, supporting the accuracy of the new method. From applying this new approach, larger passive and active ankle stiffness and damping were evident among older adults and males. Following the onset of ankle muscle fatigue, active ankle torque increased, particularly among young adults and males. Furthermore, the estimated time intervals between active zones was slightly larger but still comparable to an earlier report of measured intervals between muscle bursts (Loram et al., 2005).

6.2.6 Passive and Active Ankle Torque, Stiffness, and Damping: Implications

A new method was developed to estimate passive and active ankle torques, stiffness, and damping. The method is relatively easy to implement and is computationally efficient. Importantly, this new method does not require perturbations and so is also considered experimentally efficient.

Larger passive and active stiffness and damping were evident among older adults. These results are seemingly in contradiction to the general lower muscle strength with age, and thus the potentially lower ankle torque among older adults (Jones et al., 1999, Newman et al., 2003). However, very low levels of muscle activity are needed to maintain quiet upright stance (Runge et al., 1999, Loram et al., 2004), and thus muscle strength does not appear to contribute greatly to active ankle stiffness during quiet upright stance. The greater ankle stiffness predicted among older adults may be due to the use of more rigid and less coordinated postures, which was also demonstrated through the earlier study of two-joint and whole body coordination.

Given the increased ankle stiffness among older adults, one relevant question is whether this difference is also associated with a higher risks of falls (L Sturnieks et al., 2008, Lacour et al., 2008). As reported earlier, a higher stiffness is associated with less flexibility of the underlying system (Latash and Zatsiorsky, 1993). Thus, higher ankle stiffness in the older group may indicate that the physical system of older adults is more rigid and thus less able to counteract any unexpected environmental perturbations such as slips and trips. However, larger ankle stiffness is unlikely a sole predictor of risks of falls, since the current results also indicated that males (especially older males) have larger passive and active ankle stiffness, yet males also have less risks of falls (Hannan et al., 1995, Kiely et al., 1998, Stevens and Sogolow, 2005). As such, ankle stiffness should be used in conjunction with other measures such cognitive resources and muscle strength for predicting risks of falls.

6.3 Research Limitations

6.3.1 Experiment Limitations

The results of this investigation are subjected to some limitations that warrant discussion, and in particular, two limitations that are consistent across the all studies are notable. Both were somewhat unavoidable in the current work, since an existing dataset was employed throughout. First, the ankle plantar flexors were fatigued unilaterally, yet the control of bilateral upright stance requires both legs. As such, any conclusions regarding the effects of fatigue from the current work may not generalize to bilateral fatigue conditions. Second, during kinematic/kinetic modeling the anterior superior iliac spine (ASIS) was used to quantify the pelvis kinematics. This made it difficult to estimate the true hip joint and L5/S1 location. While care was taken, through estimating these locations by referencing to anatomical literature and geometrical assumptions/relationships, it is still possible that some errors occurred during the estimation process.

6.3.2 Methodological Limitations

Regarding the methods used here, one limitation related to the use of only a single-segment inverted pendulum model to assess quiet upright stance. Though several other studies have used a similar approach (Kuo, 1995, Peterka and Loughlin, 2004, Maurer et al., 2006, Qu and Nussbaum, 2010), even for the small motions involved during quiet upright stance a multi-segment approach can provide higher accuracy for estimating joint kinematics and kinetics (Kiemel et al., 2008). As such, a multiple segment approach is expected to address the limitation, and thus should be one focus of future research. A second important limitation is related to the wavelet coherence analysis, during which coherence associated with only frequencies of 2.5 - 4 Hz was analyzed, despite evidence that other frequency bands may also be either interesting or important (Hsu et al., 2007). As such, future investigation of lower and higher frequency bands would be of benefit.

6.4 Future Work

All three of these studies built upon an assumptions that human upright stance involved intermittent instead of continuous postural control. In turn, the three studies provided support for this assumption. First, two-joint coordination appeared intermittent. Second, an intermittent sliding mode controller was able to track upright stance motions with good accuracy. Third, and inspired by the intermittent control theory, a new method was used to estimate passive and active ankle torque, stiffness, and damping, and the results were comparable to other studies using other methods.

Building on this, and on related studies by other researchers, several directions for future work are recommended. Moving beyond existing studies, which only measured a few of the ankle muscles using ultrasound techniques, activities of additional muscles should be measured, both within the ankle but also more generally in the lower extremity and trunk. Through measurements of these muscle groups, both ankle and trunk muscle activities can be obtained, which can be used to validate ankle-trunk coordination patterns, which have not otherwise been experimentally verified. In addition to this, it also can provide experimental evidence for future modeling of quiet upright stance using two-segment inverted pendulum models. The new method described here for calculating passive and active ankle stiffness can also be extended to the trunk joint. Overall, such work can help contribute to validating intermittent control theory and/or may inspire new and improved intermittent postural control theories and models.

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