ASSESSING INTER-JOINT COORDINATION DURING WALKING

by

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DISSERTATION ABSTRACT

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Title: Assessing Inter-joint Coordination during Walking

Coordination indicates the ability to assemble and maintain a series of proper relations between joints or segments during motions. In Dynamical Systems Theory (DST), movement patterns are results of a synergistic organization of the neuromuscular system based on the constraints of anatomical structures, environmental factors, and movement tasks. Human gait requires the high level of neuromuscular control to regulate the initiation, intensity and adaptability of movements. To better understand how the neuromuscular system organizes and coordinates movements during walking, examination of single joint kinematics and kinetics alone may not be sufficient. Studying inter-joint coordination will provide insights into the essential timing and sequencing of neuromuscular control over biomechanical degrees of freedom, and the variability of inter-joint coordination would reflect the adaptability of such control.

Previous studies assessing inter-joint coordination were mainly focused on neurological deficiencies, such as stroke or cerebral palsy. However, information on how inter-joint coordination is modulated with different constraints, such as walking speeds, aging, brain injury or joint dysfunctions, are limited. This knowledge could help us in identifying the potential risks during walking and improve the performance of individuals with movement impairments. The purpose of the present study was to investigate the

properties of inter-joint coordination pattern and variability during walking with different levels of neuromuscular system perturbations using a DST approach, including an overall neuromuscular systemic degeneration, a direct insult to the brain, and a joint disease.

We found that aging seemed to reduce the pattern adaptability of neuromuscular control. Isolated brain injury and joint disease altered the coordination pattern and exaggerated the variability, indicating a poor neuromuscular control. To improve gait performances for different populations, clinical rehabilitation should be carefully designed as different levels of neuromuscular system constraints would lead to different needs for facilitating appropriate coordinative movement.

This dissertation includes both previously published/unpublished and co-authored material.

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CHAPTER I

INTRODUCTION

Coordination and Locomotion

Coordination indicates the ability to assemble and maintain a series of proper relations between joints or segments during motions [1]**.** Motor coordination generally consists of several perspectives, including temporal and spatial patterns, multiple tasking and adaptation [2]. Temporal pattern is the latency or the relative timing between joints or segments positions. Spatial pattern is associated with the pattern selection or the relative position between joints or segments. Multiple tasking refers to the goal directed tasks that involve different limb movements. Adaptation is motion integration with sensory organization and refers to the variability of motor coordination, such as the ability to change or reorganize movements in response to external demands. Accordingly, coordination can be quantified by the spatial and temporal domains independently, and it has task-dependent as well as adaptation properties.

Reviewed from previous literatures, a common issue associated with movement coordination is how our central nervous system (CNS) controls multiple biomechanical degrees of freedom of the human body during locomotion. Human movements require coordinating approximately 10^2 joints, 10^3 muscles, and 10^4 cells for a goal directed task. Effectively organizing these multiple degrees of freedom and turning the joints, muscles, and motor units into a coordinative controllable system are necessary for a functional movement [3]. Movement coordination during skill learning and acquisition can be explained from this perspective with three stages. First is to freeze some degrees of freedom to simplify the control. This is followed by gradually release and integrate degrees of freedom into a target movement. Finally, the movement is achieved with further exploring forces and refinements.

Human gait requires multi-joint coordination to complete a precise end point motor control [4]. It involves higher levels of neuromuscular controls to regulate the initiation, intensity and adaptability of locomotion [5]. Previous studies have shown how locomotor coordination master or degenerate the degrees of freedom into a controllable system by constraints, such as movement tasks, environment, and visual stimulation. Lacquaniti et al. (1997) indicated that inter-joint coordination was essential to the end-point control of limbs and the maintenance of dynamic equilibrium during walking [6]. During walking, the degrees of freedom of lower limbs motions were reduced by planar constraints, specifically in the sagittal plane, and the planar co-variances of lower limb joint angles determined the spatio-temporal trajectories of center of mass [7,8]. Therefore, the control of lower limb kinematics would affect the conservation of total mechanical energy and the systemic balance control during walking [8,9]. Assaiante C. (1998) found that the degrees of freedom were gradually under control by the movement improvements with ages, and that the balance control during locomotion depended on the environmental requirements for the children [10]. Montagne et al. (2002), with the use of different virtual environments on treadmill walking to induce stepping reactions, demonstrated that the regulation of locomotion (control mechanism) could be influenced by task constraints [11].

To better understand how neuromuscular system organizes to coordinate movements during walking, examinations on a single joint kinematics, kinetics, and electromyography may not be sufficient. Barela et al. (2000) indicated that the relative phase between joints or segments might be a better variable to capture the organization of neuromuscular system during walking than single joint biomechanical measures (i.e., knee joint kinematics), and that the differences or changes in relative phase could provide insights into the neuromuscular deficits in stroke patients [12]. Similarly, several studies have argued that complex movements of human body cannot be merely constructed from the sum of single joint motions [13,14]. For example, in reaching and grasping tasks, cerebellar lesion patients could perform well controlled on single joint motion one at a time, however, the multiple joint movements were found not coordinated [15]. Cordo et al. (2004), with the use of sit-up and leg lifting exercises, demonstrated that there were control differences between single and multiple joint movements [13].

The notion of coordination among joints or segments is to functionally control over biomechanical degrees of freedom. Inability or inconsistency in such neuromuscular coordination may result in pathological movement patterns [4]. During walking, movement mobility and single joint property may not be directly associated with inter-joint or inter-segment coordination ability. For example, a stoke patient who walks fast does not necessary imply to have a more coordinative walking pattern than those who walks slow, as he/she may use some compensation strategies. Changes in a single joint motion during activities could still maintain similar relative motions with other joints. Therefore, investigations on the inter-joint or inter-segment coordination may provide insights into the essential timing and sequencing of neuromuscular system's control over biomechanical degrees of freedom, and the variability of coordination could reflect the adaptability of this control [17-19].

Dynamical Systems Theory

Dynamical Systems Theory (DST) states that brain and behaviors (movements) are self-organized coordinative structures. Movement patterns form from the synergetic organization of the neuromuscular system by the constraints of anatomical structures, environmental factors, and movement tasks. This would reduce the numbers of degrees of freedom and simplify the neuromuscular control. Furthermore, movements of body segments are linked functionally by coordinative structures, such as a group of neurons, muscles, and joints, which act together as co-variances as a result of shared afferent or efferent signals. Coordinative structures are like the emergent dynamic systems resulting from the convergence of constraints as pursuing tasks or functions [20,21]. Therefore, coordination of a movement presumably contains detailed information about the neural circuits in central nervous system [17].

An important feature of DST is that it proposed the nonlinear oscillation and limit cycle characteristics of behaviors (or movements) [20]. The nonlinear oscillation is that in an oscillating system (such as a physiological system), a perturbation to the system would result in a brief alternation on the oscillation but then followed by a quick return to its original oscillatory behavior, namely an attractor (a preferable movement pattern). A nonlinear oscillating system which exhibits a stable periodic behavior despite of small perturbations is called a limit cycle. The limit cycle oscillation is assumed to be autonomous and rhythmic [20,21]. As it had been indicated that the behavior of a system could be obtained from a variable and its first derivative over time [22], graphical examination of a movement can then be observed by constructing a phase portrait using its positions and velocities over time. The limit cycle can be described as a closed orbit to which all nearby trajectories are attracted on a phase portrait (Figure 1.1). The stability of the limit cycle is based on a balance between the excitation and inhibition from the nervous system as well as the dissipation. Excitation predominates the small variables inside the circle and thus causes an increase in amplitude to move toward the attractors, and vice versa [20,21].

The relationships between two limit cycle attractors can be presented by coupling based on the phasing relations [20,21]. These coupling relationships could be used to consider the intra-limb coordination or inter-limb coordination. In order to obtain the phasing relationships between two joints or segments during walking, the Cartesian coordinates of each data point on the phase portraits will have to be converted to equivalent polar coordinates which are also known as phase angles (φ) (Figure 1.1). The subtraction of the phase angles of one joint or segment from that of another is defined as the relative phase or coupling between two joints or segments [20,21,23,24]. Continuous relative phase (CRP), the difference in four quadrants arctangent phase angles between two joints or segments as shown in Figure 1.2, has been used to investigate the inter-joint or inter-segmental coordination pattern and variability in various activities [18,19,25-27].

In DST, there are two major stable coordinative patterns observed and defined systemically [20]. When CRP is close to 0° or +/- 360°, there is a synchronized motion of the two adjacent joints, namely in phase. Similarly, when CRP is close to 180°, the two adjacent joints are moving in the opposite direction (out of phase), implying that the two joints are moving in syncopation. The slope of CRP defines the leading and lagging relationships between two joints [28,29]. For example, CRP in figure 1.2 is determined by subtracting the phase angles of distal joint from that of proximal joint, so a positive slope represents the proximal joint is leading the distal joint, and vice versa. The self-organization of brain and behaviors (movements) refers to the switches between these two coordinative patterns (also known as phase transitions). Kelso et al. has demonstrated that human movements have the tendency to spontaneously transit from out of phase to in phase coordination at a critical movement rate, especially when the movement frequency is high. However, the in phase coordination will not tend to switch in out of phase coordination unless under intentions [20,30,31].

Variability of relative phase is associated with the variations of the trajectories on the phase portraits, which provides information regarding to the variability of the selected movement patterns [29]. Generally, an increase in relative phase variability is associated with a more unstable movement pattern, indicating a generation of new movement or switch to different movement pattern. A reduced variability is considered to be associated with pathological condition. However, the boundaries for higher or lower variability are still undetermined. Studies have shown that a certain amount of variability is required for the transition between movement conditions or gradual adaptation to a new movement pattern [17,32]. In general, a reduced variability could be recognized as lack of flexibility [18,32], and an excessive variability may reflect the presence of injuries, risks of injuries or diseases altering motor control patterns [25,26,28,33,34].

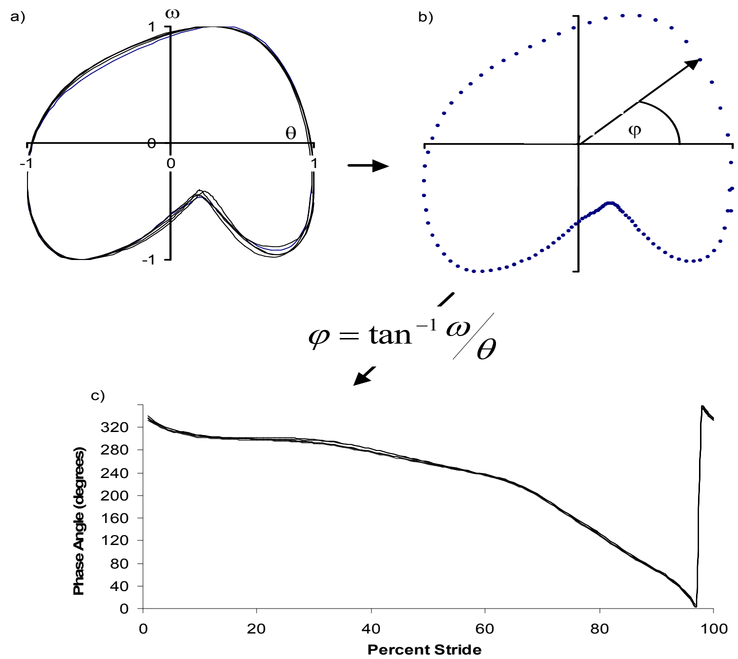


Fig. 1.1. Defining a phase angle (φ) on a phase portrait. (adapted from Haddad et al., 2006 [16])

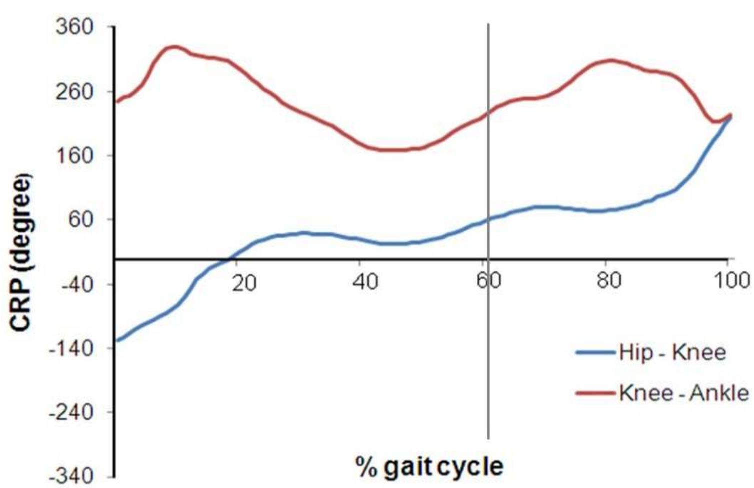


Fig. 1.2. Continuous relative phase (CRP) of Hip-knee and knee ankle inter-joint coordination. (Data constructed from a healthy young adult during walking)

Neurological Evidence for Brain Activation and Coordination

Neurophysiological studies that used full head superconducting quantum interference devise (SQUID), electroencephalographic (EEG), or functional MRI to monitor brain activities have shown that there is a direct and robust relationship between the relative timing of cortical activities and the movement and velocity (relative phase) [30,35,36]. These suggested that movement and velocity are represented in the discharge rate of motor cortical cells. The generation of a neural activity pattern also mirrors the movement velocity over time [31]. Fuch et al. (2000) demonstrated that movement velocity curves could be used to reconstruct brain activities curves in magnetoencephalography (MEG). Reorganization of brain activities during phase transitions can also be reconstructed from the movement velocity curves [37]. Further, the coordinative patterns and phase transitions (fluctuations and timing for switching coordination patterns) of a movement were found to synchronize with that of brain activities and EMG muscle activities [30,31,35,36], indicating that there are self-organization mechanisms underlying the coordinative actions of both brain and movements by switching one coordinative pattern to another [20].

The dynamic features of brain activation during inter- and/or intra-limb coordination are mediated by pattern and variability, rate/frequency of movement, and movement complexity. In addition, hemisphere specialization and the direction and amplitude of movement coordination may also have influence on the pattern of brain activation. For pattern and variability, syncopation pattern (out of phase) has greater variance (variability) and is associated with greater brain activities or higher blood oxygen level dependent activities compared to synchronization pattern (in phase), indicating that syncopation pattern may require higher levels of cognitive process. Compared to synchronization, syncopation was found to be associated with increased brain activities in bilateral dorsal premotor cortex, supplementary motor area (SMA), insula, bilateral cerebellum, bilateral basal ganglia and thalamus [38-42].

When movement frequency increases, switching pattern from syncopation to synchronization is dominated. This characteristic of movement was also associated with the brain activations. Studies have demonstrated that there were increased brain activities before and during pattern switching. Additionally, a more complex of movement was related to greater brain activities. [20,30,31,43,44].

Purpose of the Study

The majority of studies identified the biomechanical properties of lower extremities motions during walking with investigations on the isolated individual joint rather than assessing the interactions between joints. Previous studies addressing inter-joint coordination during walking were mainly focused on the neurological deficiencies, gait development in infants, obstacle crossing performances, and sports related injuries, such as stroke, cerebral palsy, and runners. However, information on how inter-joint coordination is modulated with different constraints during walking, such as walking speeds, aging, brain injuries, or isolated joint dysfunctions, are limited. Recently, Haddad el al. (2006), with the use of an asymmetrical leg loading, were able to demonstrate modifications in the inter- and intra-limb coordination and suggested the need of examining inter-joint or inter-segment coordination in gait analysis [16]. Since inconsistency in inter-joint coordination ability may induce gait deviations and result in tripping or slipping [4], investigating the modulation of inter-joint coordination due to different constraints could help us to identify the potential risks during walking and improve the gait performance of individuals.

Neurological and musculoskeletal constraints can change movement coordination [12,18]. However, the influences of these constraints on the neuromuscular controls are not well addressed. Since movement coordination is majorly accomplished by both CNS system and musculoskeletal system, any perturbation to either system could cause disturbances on the movement coordination. To reveal the underlying neuromuscular control mechanisms of movement coordination, we developed a series of studies investigating the properties of inter-joint coordination pattern and variability during walking with different levels of system perturbations. For the first and second studies, we examined the effect of an overall degeneration of neuromuscular system (aging) on lower limb inter-joint coordination and its adaptation abilities**.** The third study examined the effect of an insult to the central nervous system (concussion) on lower limb inter-joint coordination. This study also compared the influences of a cognitive perturbation versus a motor perturbation using a concurrent cognitive task and obstacle crossing, respectively. The fourth study examined the effect of a localized joint deficit (hip osteoarthritis followed by total hip arthroplasty) on lower limb inter-joint coordination. While the aging studies could help us understand the effects of overall systemic declinations on the neuromuscular control, the concussion study and the localized joint deficit study could help us identify the influences of neurological and musculoskeletal constraints on neuromuscular controls, respectively.

This dissertation includes previously published and co-authored material. The work in Chapter III was published in volume 45 of the *Journal of Biomechanics* in 2012. The coauthor, Dr. Li-Shan Chou, was the principle investigator. The work in Chapter VI was published in volume 32 of the *Gait and Posture* in 2010. The coauthor, Dr. Li-Shan Chou, was also the principle investigator. Another coauthor, Dr. Tung-Wu Lu, was consulted for the programming during data analysis.

Bridge

Before we conducted the first study in chapter III, we wanted to identify reliable evaluation tools that are available to quantify inter-joint coordination. The goal of the next chapter (Chapter II) was to provide knowledge about inter-joint coordination assessments and select our method for assessing inter-joint coordination during walking. Such study was accomplished by reviewing literatures and conducting a small pilot sub-study comparing the most two commonly used assessments. By reviewing and examining the methods for assessing inter-joint coordination, a thorough understanding and interpreting of inter-joint coordination during walking could be determined for further investigations.

CHAPTER II

MEASUREMENTS USED TO EVALUATE INTER-JOINTS OR INTER-SEGMENTAL COORDINATION DURING GAIT

Summaries of Measurements

There are many different methods being used to evaluate the inter-joint or inter-segment coordination during gait. These methods can be categorized by the kinematic variables they used to define and utilize to measure the coordination. A brief summary of the advantages, limitations, and clinical applications for each category is listed in Table 2.1.

The first category of measurements uses angle - angle diagram (cyclogram), such as vector coding (VC), normalized root mean square (NoRMS), and average coefficient of correspondence (ACC). This method utilizes the angle - angle diagram constructed from two adjacent joints or segments to describe the motor or locomotor coordination, such as an ankle joint angle - knee joint angle diagram. This diagram has been suggested to be a successful way to identify the differences between normal versus pathological gait [45]. The general advantages of using an angle - angle diagram are that it maintains the original spatial information (mostly without normalization), and that it is easy for the clinical interpretation and usage. However, comparing the angle - angle diagram between each individual may not be appropriate since the ranges of joint motions differ among individual. Moreover, this diagram provides only the spatial relationship between two joints, and the parameters used to evaluate this diagram are mostly qualitative (only looking at the variability, except vector coding). Missing the temporal evolution of inter-joint or inter-segment coordination may potentially reduce its sensitivity to the variability of coordination [46].

Clinically, angle - angle diagram has successful detected the inter-joint or inter-segment coordination differences between uninjured and injured runners, gait patterns between healthy individuals and patients with peripheral arterial disease, stroke, or spinal cord injury, etc. [33,47-49]. The rehabilitation training effects were also detected using this method.

Various parameters have been used to qualitatively quantify the angle - angle diagram, including vector coding, normalized root mean square (NoRMS), and average coefficient of correspondence (ACC), etc. Vector coding (VC) is to continuously calculate the direction and magnitude of the vectors connecting two consecutive data points on the angle - angle diagram throughout a gait cycle. The variability of these vectors is also identified to represent the variability of inter-joint or inter-segment coordination [49]. NoRMS technique is to calculate the normalized root mean square of a series of angle - angle cycles on the diagrams to obtain the variability of joint coupling coordination, but the pattern information is not provided [47]. ACC stands for the average consistency or coefficient of correspondence across all frames and all steps in multiple gait cycles. The mathematical calculation for obtaining this parameter is similar to VC, but it measures only the variability of inter-joint or inter-segment coupling motions [48].

Table 2.1. Summary of advantages, limitations, and clinical applications of the measurements used to evaluate inter-joint or inter-segment coordination during walking.

Measurements	Advantages	Limitations	Clinical
Angle-angle diagram (VC, NoRMS, ACC)	- original spatial information - easy for clinical interpretation and use - VC provides a continuous measure	- individual differences not accommodated - only the spatial relationship - mostly qualitative (variability)	- injured runners - peripheral arterial disease - stroke - spinal cord injury - post-rehabilitation
Elevation angles (planar covariation)	- explains the constraints for CNS controls during walking - relate to energy consumptions	- interpreting data is subjective, difficult, and less clinical meanings - only sagittal plane motions of lower extremities - lack of variability information	- limited to healthy subject performances
Angle-velocity diagram (CRP)	- theory based (DST) - provides a continuous measure - provides spatial, temporal, and variability - individual differences accommodated	- sinusoid assumption - affected by speed and normalization	- Parkinson disease - cerebral palsy - stroke - illiotibial band syndrome - running injuries - anterior cruciate ligament reconstruction - knee osteoarthritis - post-rehabilitation

The second category of measurements uses elevation angles or planar co-variation law. Planar co-variation law measures the co-variations of sagittal plane segmental elevation angles of lower extremities during walking [50]. Elevation angle of a segment is defined by the angle formed by the segment and the vertical line. The pattern of inter-segment coordination in sagittal plane of a limb is described by plotting the corresponding elevation angles of the thigh, shank, and foot segments. A statistical analysis, principal component analysis (PCA), is then performed to analyze the weighted sum of each segment, using a linear combination of principle component curves.

Planar co-variation law has been used to explain the planar constraints for CNS controls during walking, with strong couplings observed between segments [6-8,50]. It also has been used to relate the limb controls to energy consumptions during walking [9,51]. However, interpreting PCA data could be subjective and difficult. Further, planner co-variation law can only be applied to the sagittal plane motions of lower extremities, and the variability of inter-segment coordination is not provided. Clinical applications of this method are also limited. Most of the studies used this approach to examine the inter-segment coordination are focused on looking at the normal gait pattern and the obstructed gait pattern in healthy subjects.

The third category of measurement uses angle - velocity diagram, such as continuous relative phase. Continuous relative phase (CRP) is derived from an angle - velocity diagram or position - velocity diagram (also called phase portrait). Phase portrait provides the velocity as a function of position for each joint or segment and defines its position and direction of motions across multiple points of a gait cycle. The differences in four quadrants arctangent phase angles between two joints or segments, has been used to investigate the inter-joint or inter-segment coordination pattern and variability [28].

CRP is a theoretical based (DST) measurement, and it provides a continuous measurement of the interactions between segments or joints throughout an entire gait cycle to assess the overall profile of coordination [18]. The spatial and temporal pattern and the variability information of inter-joint or inter-segment coordination are provided using CRP. Normalization process is required in this method to facilitate the comparison between individuals. However, phase portrait is constructed under the assumption that the time series of joint or segment angles are like sinusoidal signals. CRP might be affected by the speeds of motions and the normalization technique. The use of normalized position and velocity may prohibit CRP to truly represent the real-time relationship between two joints or segments [52]. Further, interpreting CRP is not intuitive as well since the position and velocity information are involved.

Clinically, this method has been used a lot and is able to distinguish the inter-joint or inter-segment coordination differences between healthy and pathological movements as well as the rehabilitation training effects, such as Parkinson disease, cerebral palsy, stroke, illiotibial band syndrome, running injuries, anterior cruciate ligament reconstruction, knee osteoarthritis, etc. [12,18,53,25,26,32,54-57].

Comparisons of Continuous Relative Phase (CRP) and Vector Coding (VC) – A Preliminary Study

Introduction

Among the methodologies we reviewed, continuous relative phase (CRP) and vector coding (VC) are the most commonly used techniques to investigate the movement coordination in various activities as. Since walking involves continuously coordinating multiple joints, we selected CRP and VC as both techniques provide continuous measurements to evaluate the pattern and variability of inter-segment or inter-joint coordination over an entire gait or a movement cycle. The differences between CRP and VC are the kinematic variables used to construct the phase portraits and the quantification methods [5]. Whereas CRP is derived from an angle-velocity phase portrait that contains both spatial and temporal information of two joints, VC is derived from an angle-angle portrait that contains only the spatial information of two joint positions. Due to these differences, studies comparing the similarity or dissimilarity of coordination information provided by CRP and VC are rarely examined. To determine our method of assessment, facilitate our knowledge on how to appropriately interpret, and compare the coordination information using CRP and VC among studies, we investigated the pattern and variability of inter-joint coordination quantified by CRP and VC.

Methods

Three healthy young adults were recruited (1 man, 2 women, age = 26.0 ± 6.5 yrs, BMI = 24.0 ± 1.2 kg/m^2). An 10-camera motion analysis system was used to collect whole body motion during level walking with preferred speed. A total of 29 reflective markers were placed on bony landmarks. Joint kinematics of the bilateral lower extremities was calculated by using OrthoTrak kinematic analysis software. CRP and VC were used to assess the inter-joint coordination pattern and variability throughout a gait cycle. For CRP calculation, the phase portrait for each joint or segments was first generated by plotting the normalized angular positions (θ) along the x-axis and normalized angular velocities (ω) along the y-axis (Fig. 2.1). Angular position and velocity data were normalized to a range of 1 and -1 along both dimensions of the phase plane, as suggested previously to minimize differences in amplitude and frequency [19,52,58]. Phase angles (φ) were calculated as $\varphi = \tan^{-1}(\omega/\theta)$ for each data point. To account for the phase discontinuities during arctangent angle computations, phase angles were unwrapped by adding multiples of 2π, and then the subtraction of the phase angles of one joint or segment from that of another was defined as the relative phase or coupling between two joints or segments [18,20]. VC was derived from constructing an angle - angle diagram with the distal joint or segment as vertical axis and the proximal joint or segment as horizontal axis to evaluate the coupling motions (Fig. 2.2). Coordination was defined as the coupling angle (γ) subtended from a vector connecting two consecutive data points related to the right horizontal, using $\gamma = \tan^{-1}(\frac{y_{i+1} - y_i}{x_{i+1} - x_i})$, $i = 1,2,3...,n-1$. These coupling angles represent instantaneous spatial relationship of two joints or segments at every instance [49]. The pattern of VC was unwrapped to avoid discontinuities. The pattern of inter-joint coordination of CRP and VC was compared by descriptive descriptions with in phase (CRP = 0° or ± 360°; VC = 45° or 225°) and out of phase (CRP = ± 180°; VC = 135° or 315°). The variability of inter-joint coordination of CRP and VC was calculated as the average standard deviation of all points on the ensemble CRP and VC curves over a gait cycle for each subject, namely the deviation phase (DP). DP values represent the cycle to cycle variability and a lower DP value indicates a more repeatable coordination between two joints [18].

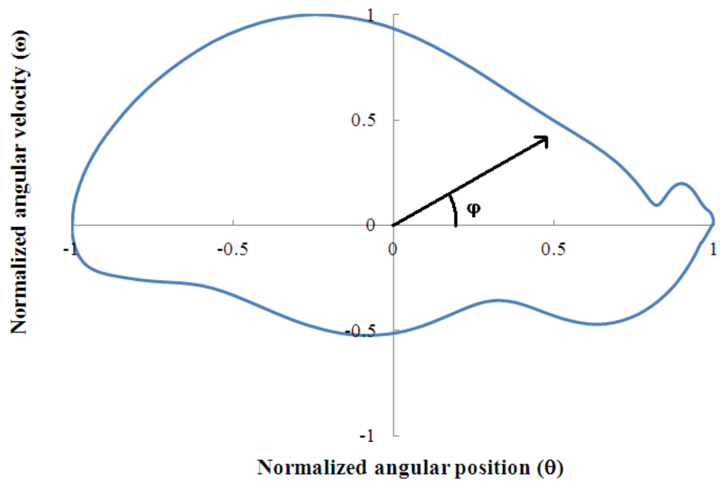


Fig. 2.1. Define phase angle (φ) on a phase portrait (with data from the hip joint motion of a representative subject).

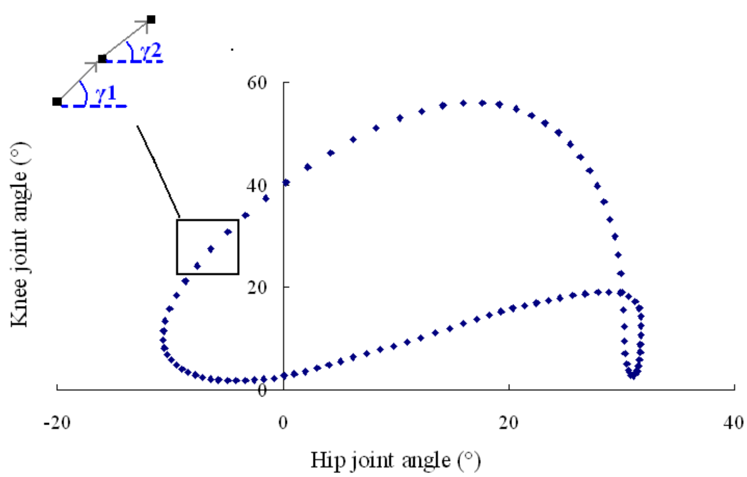


Fig. 2.2. Define the coupling angles (γ) on an angle - angle diagram.

Results

The alternations of coordination patterns between in phase and out of phase were generally in similar fashion in both techniques, except the initial contact of hip-knee coordination (Fig. 2.3). The relative motions between the hip and knee joints were in phase from the initial contact to around 60% of the gait cycle, and followed by out of phase gradually toward the end of the gait cycle. For knee-ankle inter-joint coordination, it started with out of phase relative motions, moved in phase to 20% of the gait cycle, switched to out of phase gradually from 20% of the gait cycle, sustained out of phase across 40% - 60% of the gait cycle, and then moved toward out of phase again to the end of the gait cycle. While VC seemed to have greater ranges of fluctuations on the patterns than CRP, CRP seemed to have sharper inflexion points on knee-ankle inter-joint coordination than VC. The DP values for both hip-knee and knee-ankle inter-joint coordination were similar using both techniques, respectively (Table 2.2).

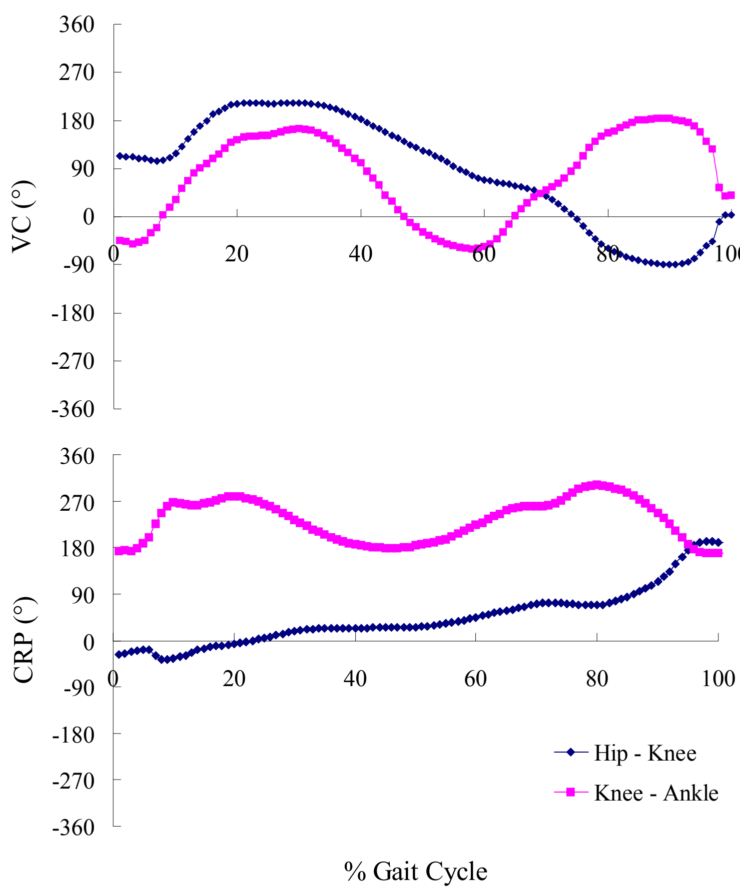


Fig. 2.3. Inter-joint coordination pattern across a gait cycle assessed by CRP and VC.

Table 2.2. DP values of inter-joint coordination (degrees)

Inter-joints	CRP	VC
Hip-knee	8.69 ± 3.27	7.91 ± 2.95
Knee-ankle	11.19 ± 2.23	11.78 ± 2.16

Discussion

Our findings suggested that there was a slight difference in the pattern and variability of inter-joint coordination presented by CRP and VC. This difference may be caused by the velocity (temporal evolution) and the normalization procedure involved in calculating CRP. The coordination information obtained from CRP and VC might be comparable with cautions. However, movement velocities were found to play an important role in finding the relationships between electromyography (EMG) and the joint kinematics properties on a phase portrait [59]. Previous studies had successfully demonstrated that the control of human movement can be validated by using phase portraits of the motions of joints or segments [20-24]. Hurmuzlu et al. (1994) has suggested that observing joint positions alone may be enough to identify the movement equilibrium during walking, however, phase portraits can be considered as useful tools to monitor the properties and changes of joints over time as they directly correlated the joint angles with respect to joint velocities [60]. Since it has been indicated that the afferent fibers in muscle receptors work most efficiently by sensing joint position and velocity and a parameter missing the temporal evolution may potentially reduce its sensitivity to the variability, CRP may provide a higher level assessment of neuromuscular control as it can define joint position and direction of motions across multiple points of a gait cycle when compared to VC [18,28]. Therefore, in the dissertation, we used CRP to investigate the inter-joint coordination.

Bridge

Chapter II identified the method we would use to assess inter-joint coordination during walking in the dissertation. The goal of the first study (Chapter III) was to understand the age-related differences in the inter-joint coordination during walking. This was accomplished by investigating the interaction of walking speed and aging. A thorough understanding and interpreting of inter-joint coordination during walking with different speeds could also be determined.

CHAPTER III

INVESTIGATION OF THE EFFECTS OF WALKING SPEED AND AGING ON INTER-JOINT COORDINATION - A PROBE INTO THE CHARACTERISTICS AND INTERPRETATION OF CRP

Introduction

Movement coordination represents the ability to assemble and maintain a series of proper relations between joints or segments during motions. According to the Dynamical Systems Theory (DST), movement patterns are synergetic organizations of the neuromuscular system based on anatomical constraints, environmental factors, and movement tasks. Such organization reduces the number of degrees of freedom and complexity of neuromuscular control [20]. A better understanding of how movement is coordinated between different joints or segments could provide insightful information about neural circuits employed by the central nervous system (CNS) [17,20].

Human gait requires a multi-joint coordination to control a precise foot trajectory [4]. Higher levels of CNS control are essential to regulate the initiation, termination, intensity and adaptability of locomotion [5]. However, examination individual joint kinematics and kinetics may not be sufficient to reveal how the neuromuscular system is organized to coordinate movement during walking [13,14]. Barela et al. (2000) demonstrated that the relation between joints or segments would serve better to capture such organization [12]. Investigations on inter-joint coordination could provide insights into the essential timing and sequencing of neuromuscular control over biomechanical degrees of freedom and variability of coordination that reflects adaptability of such control [17-19].

Walking speed is one of the factors significantly influencing joint motion [61]. Age-related differences in gait are frequently reported [62-64]. Elderly adults tend to walk slowly with signs of declined balance control [62]. Chung et al. (2010) reported that elderly adults demonstrated greater rectus femoris activities than young adults when walking at different speeds, especially when the speed exceeds 120% of the preferred walking speed [63]. Furthermore, elderly adults are frequently reported to have proprioceptive deficits in sensing joint movements [64], which are critical in planning and updating the inter-joint coordination when performing functional activities [65].

Many methods have been used to evaluate the inter-joint or inter-segment coordination during walking. Continuous relative phase (CRP) and vector coding (VC) are two of the commonly used techniques to investigate movement coordination in various activities [18,19,25-27,53]. Both techniques provide continuous measures to evaluate the coordination pattern and variability over a gait cycle. Whereas CRP is derived from the angle-velocity phase portrait that contains both spatial and temporal information of two joints, VC is derived from the angle-angle portrait which contains the spatial information of two joint positions. It has been suggested that observing joint positions alone might not be sufficient to identify the dynamic control of joint action during walking [60]. Since the movement between two joint positions could be executed with different velocities, phase portraits that correlate joint angles with velocities might better reveal such dynamic control. CRP could assess a higher order of neuromuscular control as the velocity is considered. Moreover, as afferent fibers in muscle receptors work most efficiently by sensing joint position and velocity [66], CRP could be a useful tool to quantify the inter-joint coordination [28].

Previous studies investigated relationships between gait speed and inter-joint or inter-segmental coordination were mainly focused on gait transitions between walking and running [67-70]. Although coordination between trunk and pelvis has been shown to be walking speed dependent [71,72], it is not clear whether joints of the lower extremity are coordinated similarly when walking with different speeds. Information is also limited on how the lower limb inter-joint coordination is modulated with aging. Age-related differences in inter-joint coordination during obstacle crossing [73], segmental coordination during turning [74], and lower trunk coordination during walking [75] have been investigated. Only one study reported that young and elderly adults accommodated lower limb inter-segmental coordination differently when walking on the ground with and without an asymmetrical leg loading [54]. Examining adjustments in inter-joint coordination at different walking speeds would allow us to identify modifications in neuromuscular control and to better reveal underlying mechanisms of age-related mobility impairments. Therefore, the purpose of this study was to investigate the effects of walking speed and age on the pattern and variability of inter-joint coordination in healthy adults. We hypothesized that (1) compared to the preferred walking speed, walking with a slower speed would induce more changes in inter-joint coordination than walking with a faster speed; (2) such speed effects on inter-joint coordination would be greater in young adults than in elderly adults.

Methods

Twenty adults were recruited and divided into two groups. Ten healthy young adults (5 men, 5 women, age = 24.7 ± 4.1 yrs, BMI = 22.1 ± 1.5 kg/m^2) and 10 healthy elderly adults (5 men, 5 women, age = 71.6 ± 5.2 yrs, BMI = 26.0 ± 2.9 kg/m^2). Using data reported by Bryne et al. (2002) [54], this sample size was determined to yield 80% power to detect group differences at the 0.05 level. Three dimensional gait analyses were performed. Prior to participation, each study participant provided signed consent to the experimental procedure approved by the Institutional Review Board. All participants had no histories of lower limb joint surgery, neurologic or musculoskeletal impairments, and stated no incidence of vertigo or arthritis.

Participants were asked to walk with barefoot along a 10-m walkway with three different self-selected paces: a preferred, faster, or slower speed. After each participant became familiar with the laboratory setting with several practice trials, data collection started with the preferred speed walking trials. The participant was then verbally instructed to walk as fast or slow as possible. Data collected from five trials of each walking speed were recorded for analysis. An eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) was used to collect whole body motion. A total of 29 reflective markers were placed on bony landmarks [76]. Markers were placed bilaterally on the dorsum of the foot (between 2nd and 3rd metatarsal heads), posterior aspect of the heel leveled with the foot markers, lateral malleoli, distal-medial aspect of the shank, lateral femoral epicondyles, distal-lateral aspect of the thigh, anterior superior iliac spine, and one sacrum marker (midway between posterior superior iliac spine). For the trunk and upper extremities, markers were placed bilaterally on the acromion process, lateral humeral epicondyles, dorsum of wrist line between radial and ulnar styloids, and the dorsum of 3rd metacarpal head. Markers were also placed on head at the superior apex, anterior midpoint, posterior midpoint, and bilateral temporal midpoints. Motion data were sampled at 60 Hz and smoothed using a low-pass, fourth order Butterworth filter with a cutoff frequency of 8 Hz. Gait velocity and joint angles of the lower extremities were calculated using the OrthoTrak (Motion Analysis Corp., Santa Rosa, CA).

Since walking speed and age-related changes in lower extremity joint kinematics are most dominant in the sagittal plane [62], we only examined the sagittal plane inter-joint coordination between hip and knee and between knee and ankle. A laboratory written Matlab program (The Mathworks Inc., Natick, MA) was used to calculate the CRP. Sagittal plane joint angles were interpolated to 100% of a gait cycle. To conform the applications of phase portraits and DST assumptions, which require the time series of joint angles to be like sinusoid signals [20], empirical mode decomposition (EMD) was applied to eliminate the riding waves and uneven amplitudes of joint angles in a gait cycle [77]. Joint angles were first decomposed into sinusoid waves with different frequencies. The lowest frequency component that did not satisfy with the sinusoid assumption was removed, and the remaining sinusoid waves were then reconstructed. Sagittal plane joint angular velocities were obtained by utilizing the generalized cross-validation spline smoothing and differentiation method [78]. Phase portrait for each joint throughout a gait cycle was generated by plotting the normalized angular positions (θ) (x-axis) and normalized angular velocities (ω) (y-axis). Angular positions and velocities were normalized using the following equations:

$$\theta_i = \frac{2\times[\theta_i - \min(\theta_i)]}{\max(\theta_i) - \min(\theta_i)} - 1$$

$$\omega_i = \frac{\omega_i}{\max\{|\omega_i|\}}$$

where θ_i and ω_i are the angular position and velocity for each data point during a gait cycle. Such normalization defined the angular positions and velocities between 1 and -1 along both dimensions of the phase plane and would minimize individual differences in amplitude and frequency [52,58]. Phase angles (φ) were calculated as $\varphi = \tan^{-1}(\omega/\theta)$ along each data point and unwrapped to correct discontinuities occurred during angle computation [16]. The CRP, which identifies the coordination between two adjacent joints, was then obtained by subtracting the phase angles of distal joint from that of proximal joint ($\varphi_{hip\text{-}knee}$, $\varphi_{knee\text{-}ankle}$) [28].

Walking speed effects on the similarity of CRP patterns across the entire gait cycle for each participant were examined with cross-correlation coefficients and root-mean-square (RMS) differences when comparing ensemble average CRP curves of the faster or slower speed to preferred speed walking. The cross-correlation coefficient and RMS differences quantified the temporal and magnitude differences between CRP curves, respectively. A high cross-correlation coefficient and a low RMS difference would indicate a good similarity between two curves [16]. The variability of inter-joint coordination for each participant was assessed with the average value of all standard deviations calculated for each data point over a gait cycle from all CRP curves, namely deviation phase (DP), which represents the cycle to cycle variability and compares the systemic inter-joint characteristics within a gait cycle. Therefore, a high DP indicates a more changeable coordination between two joints [19].

For all variables examined, no significant differences between limbs were detected using paired *t*-test for all participants ($p \geq 0.1$). Therefore, average values from both limbs of each participant were used for statistical analysis. Distribution of the cross-correlation coefficients was examined for its normality. If a violation was detected, the Fisher Z-transformation scores were used in further analysis [79]. A mixed-model analysis of variance with repeated measures was used to detect the effects of walking speed and age on cross-correlation coefficients, RMS differences, and DP values. Follow-up pair-wise comparison was conducted with Bonferroni adjustment to identify specific speed effects on DP values. All statistical analyses were performed with the SPSS (SPSS Inc., Chicago, IL).

Results

Average preferred walking speeds for young and elderly adults were 1.42 m/s and 1.30 m/s, respectively. When asked to walk faster, the walking speeds were 1.80 m/s for young adults and 1.63 m/s for elderly adults; while walking slower, the walking speeds were 1.01 m/s for young adults and 1.08 m/s for elderly adults. All participants were able to walk with significantly different speeds as instructed ($p = 0.001$; Fig. 3.1).

However, no significant group differences were detected at each corresponding walking condition.

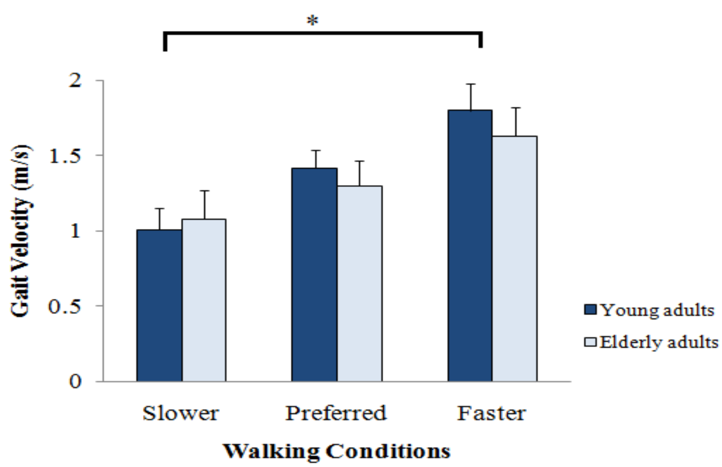


Fig. 3.1. Gait velocities of young and elderly groups at preferred, slower, and faster walking speeds (* indicates significant walking speed effects between slower and faster speeds).

For the hip-knee CRP pattern (Fig. 3.2), RMS differences between preferred and slower speeds was significantly greater than that between preferred and faster speeds in young adults (p = 0.003; Table 3.1), but were relatively invariant in elderly adults. Young adults also exhibited a significant greater RMS difference in the hip-knee CRP patterns between the preferred and slower walking than elderly adults (p = 0.004). For the knee-ankle CRP pattern, neither significant group nor speed effects were detected.

Cross-correlation coefficients between the slower or faster and preferred walking speeds in both groups were all greater than 0.96 for the hip-knee and greater than 0.77 for the knee-ankle CRP patterns (Table 3.1). When compared to young adults, elderly adults showed a significantly greater cross-correlation coefficient in the hip-knee CRP patterns (p = 0.023). Cross-correlation coefficients between the preferred and faster walking speeds were significantly greater than that between the preferred and slower walking speeds for hip-knee CRP patterns ($p = 0.001$).

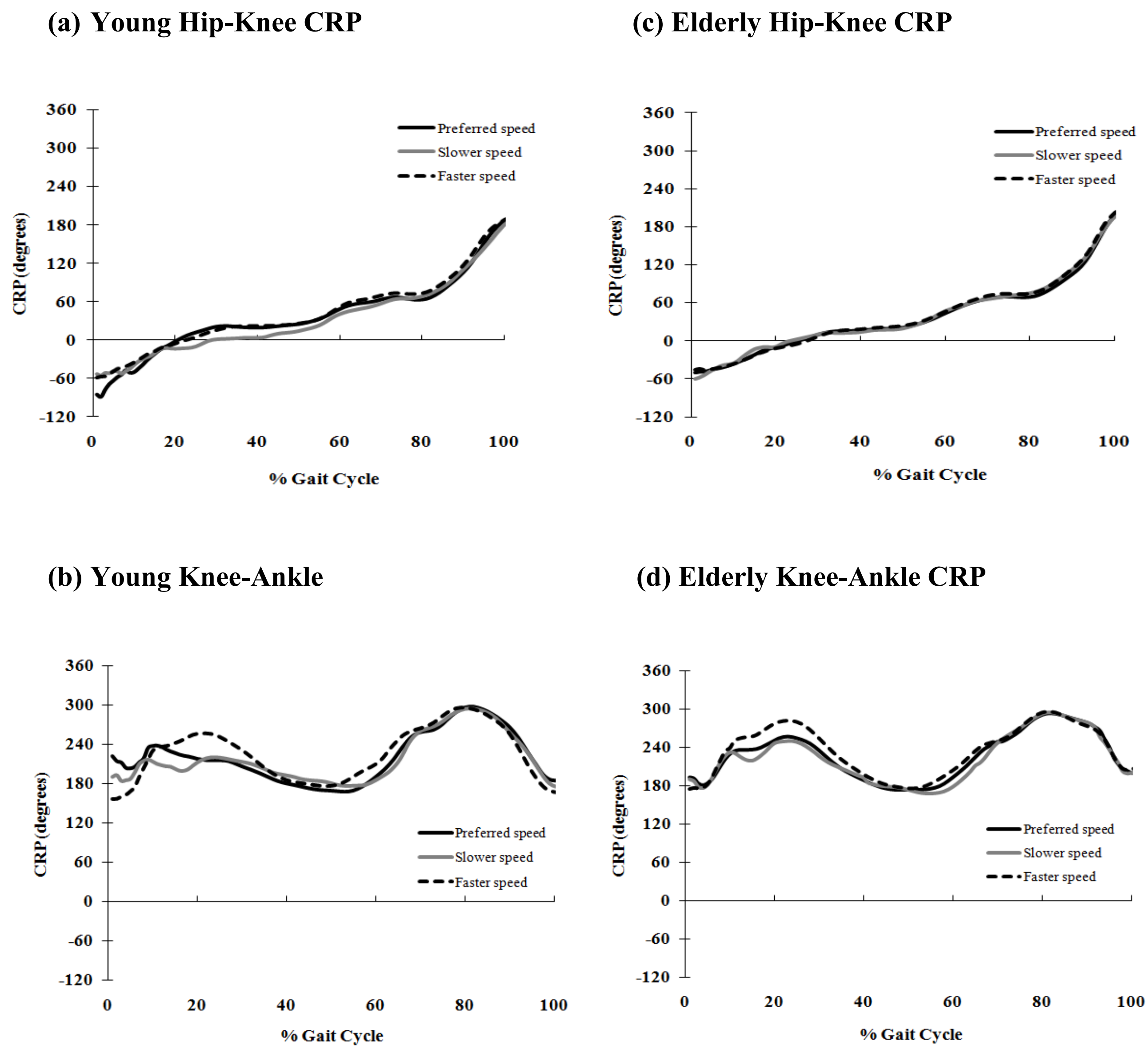


Fig. 3.2. Ensemble mean hip-knee and knee ankle CRP curves of young and elderly groups through a gait cycle at preferred, slower, and faster walking speeds. The solid black line represents the preferred speed (toe-off: Young 60%; Elderly 62%), the gray line represents the slower speed (toe-off: Young 62%; Elderly 62%), and the dash line represents the faster speed (toe-off: Young 59%; Elderly 61%), respectively. The gait cycle is defined from the heel strike to the next heel strike of the same foot.

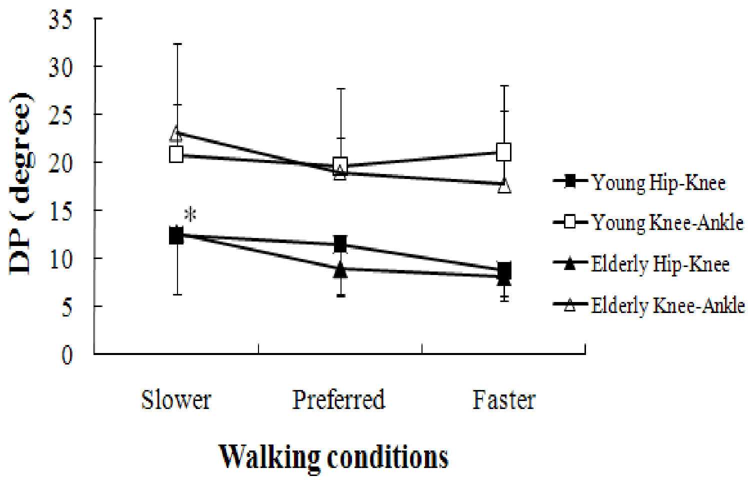


Fig. 3.3. Deviation phase (DP) values for CRP curves of young and elderly groups at different walking speeds. (* indicates significant walking speed effects on hip-knee DP values between slower and faster speeds).

Table 3.1. Root mean square (RMS) differences (°) and cross-correlation coefficients (r) in hip-knee and knee-ankle CRP patterns over a gait cycle.

	Young		Elderly		*p* values
Mean (SD)	Preferred-Slower	Preferred-Faster	Preferred-Slower	Preferred-Faster	
RMS differences					
Hip-Knee	23.3*† (9.0)	16.0 (6.4)	12.9 (4.2)	12.1 (4.1)	$P_{g\times s}$=0.046
Knee-Ankle	29.0 (8.7)	24.3 (9.4)	27.4 (7.8)	29.0 (11.4)	
Cross-correlation					
Hip-Knee	0.96 (0.02)	0.97 (0.03)	0.98 (0.01)	0.99 (0.01)	P_g=0.023 P_s=0.001
Knee-Ankle	0.77 (0.13)	0.78 (0.15)	0.78 (0.12)	0.79 (0.15)	

$P_{g\times s}$: group×speed interaction**;** P_g : group main effect; P_s : speed main effect.
* significantly different than Preferred-Faster speed; † significantly different than elderly group.

Slower and faster speed walking had significantly different effects on the hip-knee DP values ($p = 0.04$). Pair-wise comparison showed that the significant speed effects were detected in hip-knee DP values between slower and faster speed walking ($p = 0.03$). No significant group differences in DP values were detected for all three speeds (Fig. 3.3).

Discussion

In this study the pattern and variability of inter-joint coordination of young and elderly adults at three different walking speeds were examined. We hypothesized that the pattern and variability of inter-joint coordination would be significantly affected by the walking speed and age. Our results suggested that gait velocity had significant effects on the pattern of the hip-knee inter-joint coordination in young adult and on the variability of the hip-knee inter-joint coordination. Compared to young adults, elderly adults seemed to maintain similar patterns in hip-knee coordination when walking with different speeds. Since inter-joint coordination ensures the maintenance of dynamic equilibrium during locomotion [6] and a lack of ability in modulating multi-joint coordination may induce gait deviations [62], it is possible that such lacks of adopting different coordination patterns in elderly adults are associated with their declines in gait functions.

With the use of planar covariant of joint angles, Hicheur et al. (2006) demonstrated that changes in the pattern of inter-segmental coordination were associated with the changes of walking speeds [80]. Similar results were found in young adults of this study using CRP measurements. In young adults, the adjustment for walking with different speeds was mainly accomplished by changing the CRP angles between the hip and knee joints (RMS differences). Cross-correlation coefficients also showed that there were changes in temporal evolution of hip-knee CRP over a gait cycle. This indicates that changing walking speed is capable of eliciting different motor control patterns in young adults. However, elderly adults were found to maintain a relatively invariant RMS differences when walking with different speeds. This may imply that elderly adults have difficulties in adopting different coordination patterns in response to changes in motor tasks, which could be similar to age-related loss of complexity [81]. Age-related declines in proprioception and central nervous processing abilities could affect the ability of continuously planning, processing and updating their inter-joint coordination with various activities [64,65]. This result supports our hypothesis that the pattern of inter-joint coordination could be affected by the walking speed and age.

Although studies showed that a low variability was indicative of injuries related to repetitive stresses on anatomical structures [18], a excessive variability could also reflect the presence of injuries, risks of injuries or diseases altering motor control patterns [25,26]. Although elderly adults were found to demonstrate a greater variability in individual joint motion than young adults [82], such age effects on inter-joint coordination were not predominant in this study. However, young adults seem to be capable of modifying the coordination pattern accordingly to sustain their adaptabilities when asked to walk faster or slower. This may suggest that clinical efforts can utilize various walking speeds to facilitate adaptive and reliable inter-joint coordination to improve gait performances of elderly adults, and CRP can be used as an objective measure to evaluate such training effects [17].

In this study, the variability of hip-knee inter-joint coordination was significantly affected by the walking speed in both young and elderly groups (Fig. 3.3). Variability in coordination could reflect the adaptability of movement controls [17,32]. A greater variability observed during slower walking may indicate that the CNS is modulating the movement pattern to accommodate changes in walking speeds [18,32]. This suggests that the slower speed walking is more challenging to the neuromuscular control. A greater effort is required to maintain the dynamic balance and support the whole body during slower walking as the single leg support time increases.

We found that the modulation of inter-joint coordination when walking with different speeds was mainly accomplished by changing the coordination between proximal joints (hip and knee), which corresponds to previous findings reporting that hip joint motions were critical for human locomotion in timing control [83]. Proximal joints have been shown to play a greater role in balance control during walking when compared to distal joints [84]. It was also reported that a redistribution of joint moments and powers towards the hip joints during walking in elderly adults [85]. Our CNS seems to select a control strategy that modulates the proximal joint motion to effectively change the walking speed. In addition, we observed graphical differences between slower or faster and preferred speeds were mostly noticeable in stance phase of hip-knee and knee-ankle CRP patterns (Fig. 3.2). This may indicate that, when walking speed changes, modifications of inter-joint coordination patterns in stance phase are more crucial than that in swing phase.

There are limitations in this study. The pattern and variability of inter-joint coordination were only examined in the sagittal plane motion. Given this is the primary plane of motion during gait, the inter-joint coordination are expected to be more robustly controlled and illustrated. Second, we only assessed three discrete gait velocities, and all speeds were self-selected. It was our intention to examine variations in the inter-joint coordination within a range of self-selected paces and to minimize interferences in the participant's performance.

In summary, our results suggested that gait velocity had significant effects on the pattern and variability of inter-joint coordination, especially when walking at slower speeds. Compared to young adults, elderly adults seemed to accommodate to walking speeds with different neuromuscular control strategies. Assessing the inter-joint coordination as well as facilitating the coordination at different walking speeds in clinical rehabilitations may improve the gait performances of elderly adults.

Bridge

Chapter III demonstrated a novel finding for age-related declines on inter-joint coordination ability during walking with different speeds. The study showed that walking speed was a significant factor influencing the pattern and variability of inter-joint coordination, suggesting us to consider walking speeds as a covariate or to control the walking speed when using CRP to assess inter-joint coordination.

Chapter IV had a further investigation on whether such age-related declines on inter-joint coordination ability during walking were related to outcomes from commonly used clinical balance assessments for the fall risk detection in the elderly adults. Also, the single joint variability was calculated and compared to further demonstrate the significance of investigating coordination. The coordination characteristics of faller and non-fallers were further identified.

CHAPTER IV

VARIABILITY OF INTER-JOINT COORDINATION DURING WALKING IN ELDERLY ADULTS AND ITS ASSOCIATION WITH CLINICAL BALANCE MEASURES

Introduction

Falling is a serious problem in elderly adults. It is estimated that one third of those who are older than 65 years would experience at least one fall each year, and falling is also one of the leading causes of death in the elderly population [86,87]. Studies have shown that two thirds of the deaths due to falls are potentially preventable, therefore, early identification of high fall risk individuals and the development and implement of prevention interventions are important strategies to reduce the incidence of falling [86-88].

Tripping over an object during walking is one of the major causes of falling in elderly adults. Twenty percent of falls reported by the elderly adults were a result of tripping or stumbling with an object in their walking path [87,89]. Previous studies have evaluated the variability of biomechanical measures, such as gait speed, stance time, swing time, stride width and length, minimum foot clearance during foot swing, and isolated joint kinematic and kinetic variability [90-97], to differentiate fall-prone and low risk elderly adults. An increased variability has been reported to be associated with a higher risk of falling. However, a reduced variability in the step width has also been linked to a higher fall risk [98]. These associations between gait variability measures and fall risks appear to be nonlinear [98,99], and the mechanisms of such associations remain to be investigated.

Walking is a dynamic task that involves high level of neuromuscular controls integrating with visual, vestibular and proprioceptive sensors to produce coordinated limb movement. Inconsistent or poor coordination among joints during gait could result in tripping or imbalance [4]. Examining the spatio-temporal gait parameters or individual joint kinematics in isolation may not be sufficient to identify the neuromuscular control or represent the coordination ability during walking [60]. Previous studies have successfully demonstrated that dynamic control of human movement can be assessed with phase portraits constructed by the motions of two joints or segments [20,21], and movement velocities were found to play an important role in relating muscle activations to joint kinematics on a phase portrait [59]. Similar nonlinear dynamic analyses have been suggested to be a more robust approach to discriminate the fall prone from low fall risk individuals [101-102]. Therefore, investigating inter-joint coordination using continuous relative phase (CRP), which incorporates the joint position and velocity of two adjacent joints on phase portraits might be helpful in identifying elderly adults who are at risk for falls.

Inter-joint coordination during walking may be linked to the balance control ability of an individual by affecting the center of mass (COM) motion [6-8]. However, the clinical relationship between inter-joint coordination and balance control has not been established. Assessing the inter-joint coordination that directly reflects the neuromuscular control and comparing its variability between healthy and fall-prone elderly adults may provide insights into the underlying mechanisms of poor balance control. The variability of inter-joint coordination provides information regarding to the variation of the selected movement patterns. Changes in the variability of inter-joint coordination could be considered as either an external perturbation or an inherent property of neuromuscular system control, indicating signs of injuries, risks of injuries, diseases [18,25,26]. This information may help differentiate elderly fallers from non-fallers as well as enhance the development of rehabilitation interventions for them through facilitating reliable inter-joint coordination.

The purpose of this study was to investigate the characteristics of inter-joint coordination variability in healthy (non-fallers) and fall-prone (fallers) elderly adults and to determine its correlations to commonly used clinical balance measures. We hypothesized that (1) the variability of inter-joint coordination of fallers would be significantly greater than that of non-fallers; (2) high variability of inter-joint coordination would be significantly correlated with poor balance control as indicated by commonly used clinical tests.

Methods

Thirty elderly adults were recruited and divided into two groups in this study. Fifteen healthy elderly subjects (8 men/7 women, age = 75.7 ± 4.7 yrs, BMI = 26.7 ± 4.1 kg/m^2) and 15 elderly subjects who had experienced at least two unintentional falls to the ground in the year prior to the testing date (3 men/12 women, age = 72.9 ± 4.1 yrs, BMI = 29.6 ± 8.4 kg/m^2, # of falls = 3.2 ± 1.2). All subjects had no current or histories of neurological or musculoskeletal deficits that might contribute to the inability to walk independently, and no uncorrectable visual impairment, vestibular dysfunction, dementia, or depression. Gait analysis was performed to assess gait performance and joint kinematics of the lower extremities, and three clinical balance tests were used to evaluate balance control. Prior to testing, each subject provided a signed consent to the experimental procedure approved by the Institutional Review Board.

Subjects were asked to walk with barefoot along a 10-m walkway with self-selected comfortable speeds. A 10-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) was used to collect the whole body motion data during level ground walking. A total of 29 reflective markers were placed on bony landmarks [103]. Motion data were sampled at 60 Hz and filtered using a low-pass, fourth order Butterworth filter with a cutoff frequency of 8 Hz. Five successful walking trials were collected from each subject for data analysis. Clinical balance tests, including Berg Balance Test (BBS), Dynamic Gait Index (DGI) and Time Up and GO (TUG), were performed for each subject after data collection. These tests are commonly used to evaluate balance control and fall risks in elderly adults. Fallers have been predicted by low scores on BBS and DGI, and well identified by TUG tests [104-107].

Gait temporal-distance parameters and sagittal plane joint kinematics of the lower extremities were calculated using OrthoTrak kinematic analysis software (Motion Analysis Corp., Santa Rosa, CA). A Matlab program (The Mathworks Inc., Natick, MA) was used to calculate the CRP. Sagittal plane joint angular positions were interpolated to 100% of a gait cycle, and were decomposed and reconstructed through empirical mode decomposition (EMD) to conform the assumptions of applying phase portraits

[20,77,103]. Sagittal plane joint angular velocities were obtained by utilizing the generalized cross-validation spline smoothing and differentiation method [78]. To minimize individual differences in amplitude and frequency, angular positions and velocities were normalized to values between 1 and -1 [58]. Phase portrait for each joint throughout a gait cycle was formed by plotting the normalized angular positions (θ) (x-axis) and normalized angular velocities (ω) (y-axis). Phase angles (φ) were calculated as $\varphi = \tan^{-1}(\omega/\theta)$ along each data point and unwrapped to correct for discontinuities occurred during arctangent angle computations by adding multiples of 2π [16]. The CRP was then obtained by subtracting the phase angles of distal joint from that of proximal joint, such as $\varphi_{\text{hip-knee}}$ and $\varphi_{\text{knee-ankle}}$, to identify the coordination between two adjacent joints (hip-knee or knee-ankle) [28].

The variability of inter-joint coordination over the stance and swing phases of a gait cycle, derived from CRP were assessed with the average value of all standard deviations calculated from all data points, the deviation phase (DP). The variability of single joint angular positions over the stance and swing phases of a gait cycle were examined with the same method: the mean standard deviation (mean SD). Such variability represents the cycle-to-cycle fluctuations and compares the systemic single joint and inter-joint characteristics within the stance or swing phase. A greater DP or mean SD value indicates a more variant performance among different gait cycles [18].

For all DP values of CRP curves and mean SD values of single joint angular positions examined, no significant differences between limbs were detected with the paired *t*-test ($p > 0.11$ and $p > 0.15$, respectively). Thus, average values from both limbs of each subject were used for statistical analyses. Independent *t*-test was used to detect group differences in walking speeds, measures from clinical balance tests, and mean SD values. Group differences in DP values were examined using an ANOVA with walking speeds as covariates [103]. Pearson correlation coefficient was used to determine the correlations between the DP values and clinical balance tests of all subjects. All statistical analyses were performed with PASW 18 (SPSS Inc., Chicago, IL).

Results

The non-fallers walked significantly faster than the fallers ($p = 0.003$; Table 4.1). The fallers demonstrated significantly poorer performances in all three clinical balance tests when compared to the non-fallers ($p < 0.003$; Table 4.1).

For the variability of inter-joint coordination, significant group differences were detected in knee-ankle DP values (Fig. 4.1b). The fallers had significantly greater knee-ankle DP values in the stance phase but smaller knee-ankle DP values in the swing phase, as compared to the non-fallers ($p = 0.03$ and $p = 0.04$, respectively). No significant group differences were detected in hip-knee DP values during either the stance or swing phase (Fig. 4.1a).

No significant group differences were detected in the variability of the hip or knee joint angular position during the entire gait cycle (Fig. 4.2a & 4.2b), nor in the swing phase of ankle joint motion (Fig. 4.2c). When compared to the non-fallers, the fallers demonstrated a significantly greater mean SD value in ankle joint angular position during the stance phase ($p = 0.03$; Fig. 4.2c).

Pearson correlation coefficients showed that the stance phase hip-knee DP values of CRP curves demonstrated a significant correlation with DGI scores ($p = 0.001$; Table 4.2). The stance phase knee-ankle DP values of the CRP curves also correlated significantly with DGI scores ($p = 0.01$) as well as the TUG performance time ($p = 0.04$). However, no significant correlations were detected between hip-knee or knee-ankle DP values and any clinical balance measure during the swing phase.

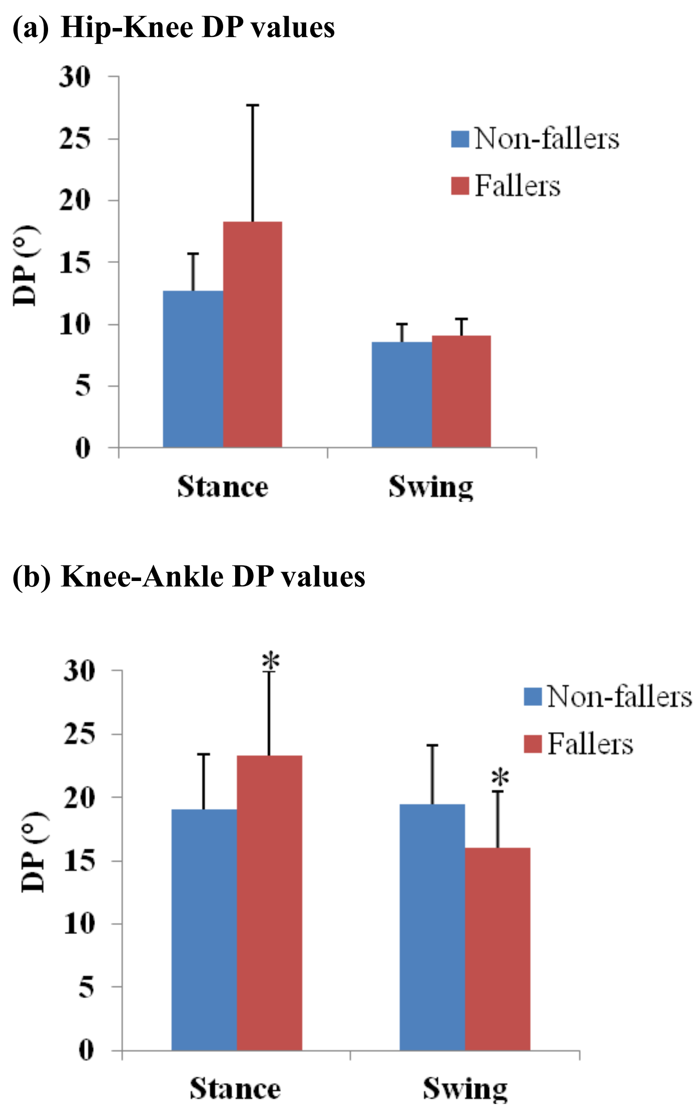


Fig. 4.1. Variability of (a) hip-knee and (b) knee-ankle inter-joint coordination in stance and swing phases

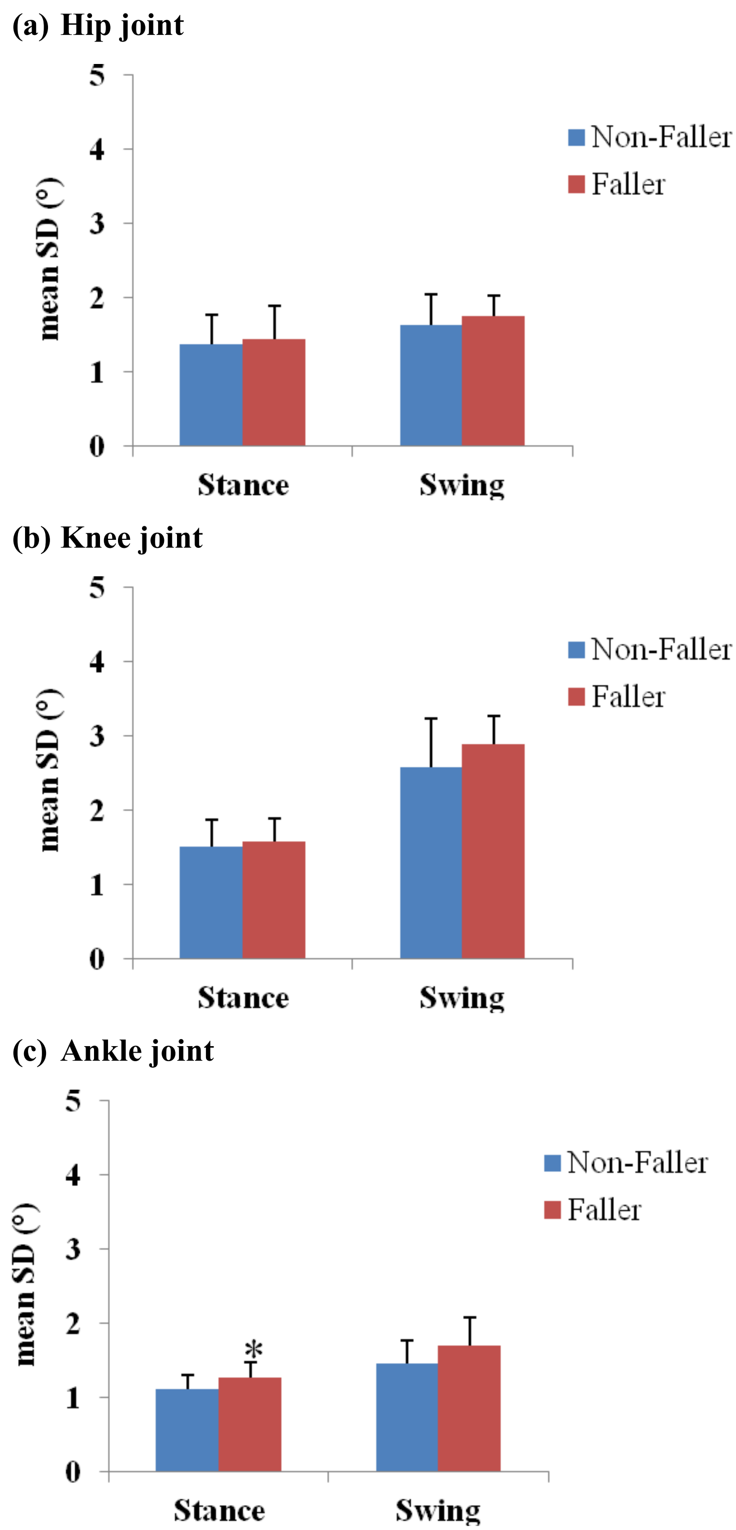


Fig. 4.2. Variability of (a) hip, (b) knee, and (c) ankle joint angles in stance and swing phases

Table 4.1. Mean (SD) of walking speeds and clinical balance tests for non-fallers and fallers.

	Non-faller	**Faller**
Gait velocity (m/s)	1.22 (0.14)	1.07 (0.12) *
BBS	55.47 (0.74)	53.44 (1.15) *
DGI	23.60 (0.74)	22.56 (1.03) *
TUG (s)	7.49 (1.20)	9.06 (1.26) *

*Significant group differences, $p < 0.05$.

Table 4.2. Pearson correlation coefficient (r) between overall DP values and clinical balance tests.

	Stance		**Swing**	
Clinical tests	Hip-Knee	Knee-Ankle	Hip-Knee	Knee-Ankle
BBS	-0.187	-0.284	-0.062	0.275
DGI	-0.559 *	-0.437 *	-0.100	-0.021
TUG (s)	0.236	0.362 *	0.313	-0.235

*Significant correlation, $p < 0.05$.

Discussion

In this study we investigated the variability of inter-joint coordination in healthy (non-fallers) and fall-prone (fallers) elderly adults and its correlations to commonly used clinical balance measures. Our results showed that, compared to healthy elderly adults, fallers demonstrated a significant high variability of knee-ankle inter-joint coordination in the stance phase and a lower variability of knee-ankle inter-joint coordination in the swing phase. In addition, the variability of ankle joint motion control in fallers was higher than non-fallers.

During walking, the lower extremities have different functions and purposes during the stance and swing phases, respectively [4]. Although most of fall were reported to be due to a tripping or stumbling of the swing foot, a robust control of the supporting limb is required for balance maintenance and safe ambulation. The timing and the amount of loadings of the supporting limb modulates the sensory transmission for force production, speed, step perturbations, and stability during walking. Additionally, this would further influence the extent and timing of the contralateral swing limb through afferent presynaptic inhibitions [108]. The greater variability of knee-ankle inter-joint coordination observed during the stance phase in fallers may indicate an inconsistency in neuromuscular control for weight-bearing during gait. Such elevated variability or unsteadiness could affect movement control of the swing limb through the presynaptic inhibition mechanism and perturb the swing foot trajectory.

In contrast, the variability of knee-ankle inter-joint coordination was found to be smaller in fallers during the swing phase. Studies have shown that a certain amount of variability is required for the transition between movement conditions or the adaptation to a new movement pattern [17]. However, a reduced variability could be recognized as lack of flexibility [18], and an excessive variability may also reflect the presence of injuries, risks of injuries or diseases altering motor control patterns [18,25,26]. Similar to the variability of gait parameters, there seems to be a certain range associating the variability and low risk of falls [98,99]. In addition, fallers tended to walk with reduced ankle ranges of motion and a delayed onset in the dorsiflexion of the ankle at the swing phase [109]. It is possible that the reduced variability of knee-ankle inter-joint coordination in fallers is the result of a poor adaptive neuromuscular control for swing foot trajectories, which prohibit the execution of flexible avoidance control strategies when negotiating with unexpected disturbances or obstacles while walking. Coupled together with the excessive variability of the supporting limb, changes in the inter-joint coordination could be one of the contributing factors to the elevated risk of tripping or stumbling during walking in the fall-prone elderly.

Fallers also demonstrated a greater variability of ankle joint motions in the stance phase when compared to non-fallers. This could be caused by the loss of muscle strength and joint flexibility during aging [34], and may potentially contribute to the elevated variability of knee-ankle inter-joint coordination observed in the stance phase. However, this was not observed in the swing phase. This suggests that examining single joint angular positions alone might not be sufficient to identify the swing foot control problems in fallers. Since walking is a dynamic condition with precise foot trajectory controls and dynamic equilibrium maintenance [6], investigating the variability of inter-joint coordination may provide insights on the neuromuscular changes contributing to fall risks in elderly adults.

Significant correlations between a higher variability of inter-joint coordination during gait and poor clinical balance assessment scores in elderly adults were only observed in the stance phase. This result is reasonable given that balance maintenance is the primary function of the supporting limb during gait. Such an association could indicate that increased variability of inter-joint coordination in the supporting limb could lead to a poor posture or balance control. Furthermore, the correlation between variability of inter-joint coordination and clinical balance tests was mainly observed with the DGI. While BBS evaluates the ability of balance control during static posture changes, DGI examines an individual's ability of balance control during walking with different perturbations, such as head turns and crossing objects [104,105]. Therefore, it is not surprising to see that DGI demonstrated a significant correlation with the variability of inter-joint coordination of the supporting limb. On the other hand, the TUG test, which examines sit-to-walk transition, turning and walking [106,107], was found to significantly correlate with the variability of knee-ankle inter-joint coordination. This might be explained by the reduced knee extension angles, knee extensor and ankle plantarflexor moments observed in elderly adults during sit-to-stand transition [110].

We also found that the differences observed in the variability of hip-knee inter-joint coordination between fallers and non-fallers were due to different walking speeds. This result corresponds to previous reports indicating that proximal joints play an important role in maintaining dynamic balance and modulating walking speeds during walking [4,103]. Furthermore, our findings from the variability of knee-ankle inter-joint coordination indicate that the controls of distal joints are critical for a safe locomotion in elderly adults.

This present study demonstrated the feasibility of utilizing inter-joint coordination to distinguish fallers from non-fallers. Clinically, the DGI could be used as an indicator to reflect an individual’s ability in controlling inter-joint coordination. Nevertheless, we only examined the inter-joint coordination in the sagittal plane. Future works that investigate the frontal plane inter-joint coordination properties and evaluate the training effects of facilitating reliable inter-joint coordination will enhance the development of fall prevention interventions.

In conclusion, our results suggest that the clinical implication for declines in the inter-joint coordination during walking could be indicated by a poor DGI. The excessive variability of the supporting limb and reduced variability of the swing limb in knee-ankle inter-joint coordination of fallers may contribute to their elevated risk of imbalance, tripping or stumbling during walking. Clinical efforts should be devoted to improve the neuromuscular control of knee-ankle inter-joint coordination during walking for fall prevention.

Bridge

Chapter IV demonstrated that the age-related fall risks were associated with the variability of inter-joint coordination. The clinical relationship constructed could be used to identify elderly adults as potential fallers and non-fallers as well as provide a reference for developing an intervention of fall prevention.

After investigating the general degeneration of neuromuscular system (aging) effects on lower limb inter-joint coordination, we specified two different levels of perturbations on the neuromuscular system in the following studies, including a neurological impairment (Chapter V) and a musculoskeletal impairment (Chapter VI). The study in Chapter V examined the effect of a mild central nervous system perturbation (concussion) on lower limb inter-joint coordination. Additionally, this study compared the influences of increased physical and mental demands (obstacle crossing and dual-task) on inter-joint coordination to help us further investigate the neuromuscular control properties during walking.

CHAPTER V

CONCUSSION INDUCES AN EXCESSIVE VARIABILITY IN INTER-JOINT COORDINATION DURING WALKING WITH MOTOR AND COGNITIVE DISTURBANCES

Introduction

Traumatic brain injury (TBI) is one of the major causes of disability and death in the United States with an estimation of 1.7 million incidences every year, and of these approximately 75% of TBIs are mild traumatic brain injury (concussion) [111]. It has been reported that those suffering a concussion have increased static and dynamic posture instability [112], increased errors and reaction time of responses during motor tasks [113,114], and inefficient cognitive process [115]. Neurophysiological studies have also reported transient or permanent structural and functional changes in the brain associated with motor control and coordination following concussion [116].

Tripping incidences are hazardous following a concussion [117]. When walking and crossing an obstacle, the central nervous system has to adjust the control of locomotion to the obstacle's location, height, width, and foot approximation via vision to complete a safe crossing [118]. Previous studies have examined the biomechanical gait characteristics in concussed individuals, including the single joint kinematics, kinetics, and center of mass (COM) and center of pressure (COP) control. During level walking, the concussed individuals tended to walk slower with shorter steps [119,120], and had a greater medio-lateral (ML) but reduced antero-posterior (AP) COM motion [119] than controls. During obstructed gait, concussed individuals similarly had slower walking speeds [120], greater ML COM motions, smaller AP COM motions [121], greater ML COM-COP separation in single limb stance [121], greater peak hip flexion during crossing [120], larger variability in obstacle clearances [117], and more obstacle contact incidences [117] than control subjects. While these studies have demonstrated that a mild traumatic brain injury would induce dynamic balance deficits, the single joint kinematics of concussed individuals during level walking were found unchanged [120].

Human locomotion is a complex task involving high level of cognitive and cortical controls that integrate with sensory inputs to produce coordinated limb movements for a smooth and secured transit. Discordant lower limb coordination may influence the trajectories of COM and result in gait deviations or tripping [4-7]. However, observing only single joint kinematics, kinetics, and COM-COP control may not provide sufficient information on walking lower limb coordination of concussed individuals [60]. Examinations of lower limb inter-joint coordination during walking and obstacle crossing have not yet been conducted in concussed subjects, but could provide insights about neuromuscular control strategies in that population.

Previous studies had successfully identified the pattern and variability of lower limb inter-joint or inter-segment coordination in various populations using continuous relative phase (CRP) [18,25,26]. CRP, the difference in four quadrants arctangent phase angels between two joints or segments, is a continuous measurement of the interaction between joints or segments throughout the gait cycle using the joint angle-velocity phase portraits. Phase portraits provide velocity as a function of position for each joint or segment and define its position and direction of motions across multiple points of a gait cycle. As it has been indicated that the afferent fibers in muscle receptors work most efficiently by sensing joint position and velocity, CRP is recommended as a useful tool to quantify the inter-joint coordination due to its usage of joint angle-velocity phase portraits [28].

Attention is an important mediator for motor coordination [122]. Bimanual coordination, which is a high attention demanding task, has been associated with high coordination variability [122], and induces a trade-off between coordination variability and reaction time [123]. Although posture control seemed to take priority over concurrent cognitive processing in a dual-task situation [124], the cognitive tests had significantly influences on the dynamic balance control following concussion. Such dual-task gait tests have been suggested to better distinguish concussed individuals from healthy individuals [114, 125]. However, to our knowledge, no study has yet reported on lower limb inter-joint coordination during dual-tasking following a concussion.

Individuals with a prior concussion were reported to have a higher risk for suffering repetitive concussions, which could lead to permanent brain damages [126]. Assessing lower limb inter-joint coordination that directly reflects the neuromuscular control during walking under different conditions and comparing its variability between healthy and concussed individuals may reveal the underlying mechanisms of such elevated risks. The information could enhance the development of rehabilitation interventions and improve the assessment of the motor recovery for concussed individuals. Therefore, the purpose of this study was to investigate the effect of concussion on inter-joint coordination of the lower limbs during single-task walking, obstacle crossing, and walking while simultaneously performing a concurrent cognitive task (dual-task walking). It was hypothesized that (1) compared to single-task walking, changes in the pattern of inter-joint coordination during obstacle crossing or dual-task walking of concussed individuals would be different from that of matched controls; and (2) the variability of inter-joint coordination in concussed individuals would be significantly greater than the matched controls in all three tasks.

Methods

In this study, forty-six young adults were recruited and divided into two groups. The participants included twenty-three healthy young adults (14 males, 9 females, age = 21.4 ± 3.2 yrs, BMI = 26.5 ± 5.0 kg/m^2) and 23 young adults who had suffered a grade 2 concussion (14 males, 9 females, age = 21.8 ± 3.6 yrs, BMI = 26.7 ± 5.3 kg/m^2). The concussed subjects (CONC) underwent laboratory testing within 48 hours after sustaining a concussion identified by certified athletic trainers or attending medical doctors. The severity of injury was categorized as Grade 2 according to the American Academy of Neurology [127], which defined Grade 2 as concussion-induced disorientation lasting more than 15 minutes without losing consciousness. The healthy subjects (NORM) were individually matched to a concussed subject by age, gender, sports, education level, height, and weight, and were excluded if they had any symptom or history of concussion within the past year. All subjects had no histories of lower limb injuries or disorders affecting gait. Prior to testing, each subject provided a signed consent to the experimental procedure approved by the Institutional Review Board.

A 10-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) was used to collect data on the subjects who were outfitted with a total of 29 reflective markers placed on bony landmarks [103]. All subjects were asked to walk along a 10-m walkway at self-selected comfortable speeds. Whole body motion data were collected during (1) single-task level ground walking (Level); (2) walking and crossing an obstacle corresponding to 10% body height (OB); and (3) walking while simultaneously answering continuously administered question and answer tasks (ATT). The question and answer tasks included spelling a common five-letter word in reverse, reciting the month of the year in reverse, or subtracting a certain numbers continuously [125]. Motion data were sampled at 60 Hz and filtered using a low-pass, fourth order Butterworth filter with a cutoff frequency of 8 Hz. Five successful walking trials were collected for each condition from each subject for data analysis. Gait temporal distance and sagittal plane joint kinematics of the bilateral lower extremities was calculated by using OrthoTrak kinematic analysis software (Motion Analysis Corp., Santa Rosa, CA). Horizontal toe-obstacle distance of the trailing limb, heel-obstacle distance of the leading limb, toe obstacle clearances for leading and trailing limb during OB (Fig. 5.1), and percentage of correct answers during ATT were examined.

A laboratory written Matlab program (The Mathworks Inc., Natick, MA) was used to calculate CRP. Sagittal plane joint angles were interpolated to 100% of a gait cycle. Empirical mode decomposition (EMD) was applied to eliminate the riding waves and uneven amplitudes of joint angles in a gait cycle to conform the applications of phase portraits [20,77,103]. Generalized cross-validation spline smoothing and differentiation method was used to acquire sagittal plane joint angular velocities [78]. To minimize individual differences in amplitude and frequency, normalization was performed to define the values of angular positions and velocities between 1 and -1 [58]. Phase portrait for each joint throughout a gait cycle was then constructed by plotting the normalized angular positions (θ) (x-axis) and normalized angular velocities (ω) (y-axis). Phase angles (φ) were calculated as $\varphi = \tan^{-1}(\omega/\theta)$ along each data point and unwrapped to correct discontinuities occurred during angle computation [16]. The subtraction of the phase angles of distal joint from that of proximal joint ($\varphi_{hip\text{-}knee}$, $\varphi_{knee\text{-}ankle}$), the CRP, was obtained to identity the coordination between two adjacent joints (hip-knee or knee-ankle) [28].

Differences in the inter-joint coordination CRP patterns across the entire gait cycle were examined between gait conditions with cross-correlation coefficients and root-mean-square (RMS) differences when comparing ensemble average CRP curves of OB and ATT to Level. The cross-correlation coefficient and RMS differences quantified the temporal and magnitude differences between CRP curves, respectively. A higher cross-correlation coefficient and a lower RMS difference indicated a good similarity between two curves [16]. The deviation phase (DP), which is the average standard deviation calculated from all points from all points of the ensemble CRP curve over the stance and swing phases of a gait cycle, was used to calculate the variability of inter-joint coordination for each gait condition. The DP represented the trial-to-trial variability and was used to compare the systemic inter-joint characteristics within the stance and swing phases of a gait cycle. A high DP indicated high coordination variability between two joints [18].

Preliminary analyses revealed no significant differences between limbs for Level or ATT condition using the paired *t*-test ($p > 0.30$). Therefore, average values from both limbs of each subject during Level and ATT gait conditions were used for statistical analysis. The trailing limb (TOB) and leading limb (LOB) during OB were examined separately. Independent *t*-test was used to detect group differences in toe obstacle distance of the trailing limb, heel obstacle distance of the leading limb, toe obstacle clearances for leading and trailing limb during OB, and percentage of correct answers during ATT. A mixed-model analysis of variance with repeated measures was used to detect the effects of group and task on walking speeds, cross-correlation coefficients, RMS differences, and DP values. The statistical analysis for cross-correlation coefficients, RMS differences, and DP values were performed with walking speeds as covariates [103]. Follow-up pair-wise comparison was conducted with Bonferroni adjustment to identify between task effects on cross-correlation coefficients, RMS differences, and DP values. All statistical analyses were performed with the PASW 18 (SPSS Inc., Chicago, IL).

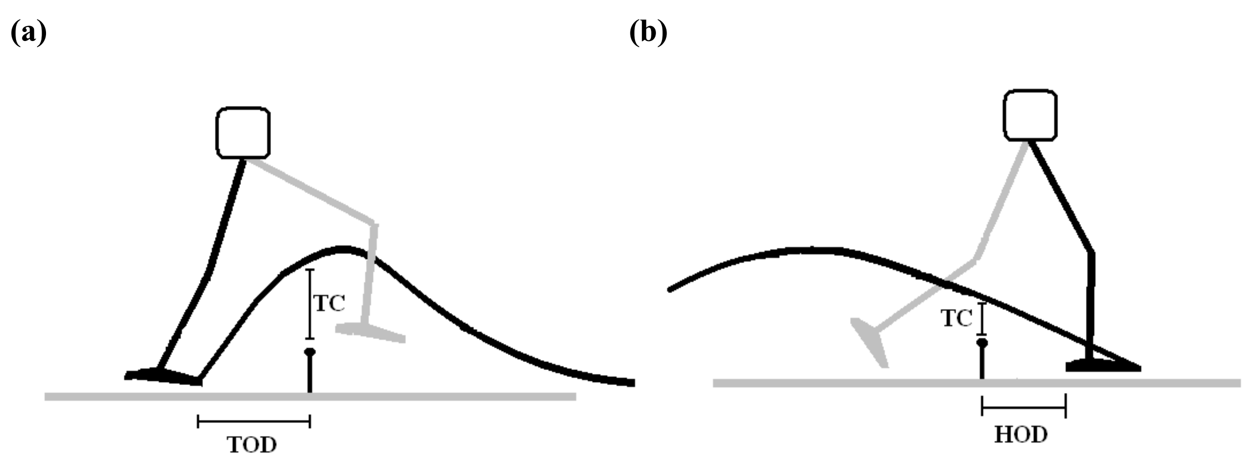


Fig. 5.1. An illustration of the spatial measures for the (a) trailing limb and (b) leading limb during obstacle crossing. (TC: toe obstacle clearance; TOD: toe-obstacle distance; HOD: heel-obstacle distance)

Results

The walking speeds for CONC and NORM during Level were 1.25 ± 0.15 m/s and 1.27 ± 0.14 m/s, respectively, and no significant group differences were detected (Fig. 5.2). During OB and ATT, CONC walked significantly slower than NORM ($p = 0.015$ and $p = 0.04$, respectively; Fig. 5.2). However, no significant group differences were detected in the horizontal toe-obstacle distance of the trailing limb, heel-obstacle distance of the leading limb, toe obstacle clearances for leading and trailing limb during OB, or percentage of correct answers during ATT (Table 5.1). When compared Level, both CONC and NORM had significantly slower walking speeds during OB and ATT ($p < 0.001$).

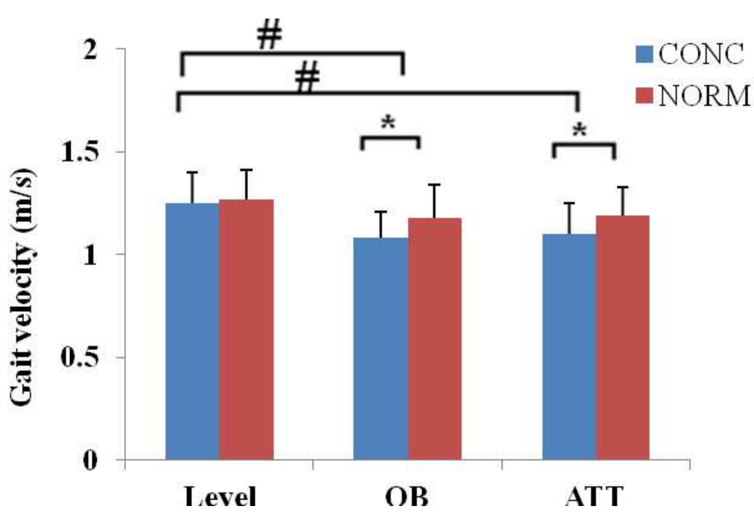


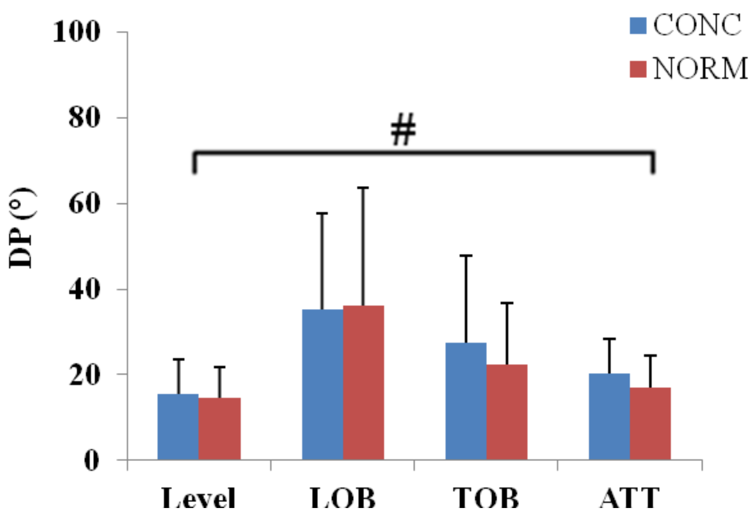
Fig. 5.2. Gait velocities of NORM and CONC groups during level ground walking (Level), walking and crossing an obstacle (OB), and walking with a concurrent attention test (ATT). (* indicates significant group difference; # indicates significant task effects)

When compared to the single-task level walking condition, RMS differences of the CRP curves were found to be the largest for the leading limb during obstacle crossing (LOB-L) of both groups, followed by RMS differences of the trailing limb during obstacle crossing (TOB-L) and then the dual-task gait condition (ATT-L) ($p < 0.001$; Table 5.2). Subjects with concussion demonstrated significantly larger RMS differences (LOB-L, TOB-L, and ATT-L) than healthy subjects for both the hip-knee and knee-ankle CRP patterns ($p = 0.003$ and $p = 0.012$, respectively; Table 5.2).

Significant task main effects were detected in cross-correlation measures of hip-knee and knee-ankle CRP patterns between obstacle crossing and level walking and between dual and single task walking for all pair-wise comparisons ($p < 0.001$; Table 5.2) For the cross-correlation measures of hip-knee CRP patterns, ATT-L had the greatest values, followed by LOB-L and TOB-L. For the cross-correlation measures of knee-ankle CRP patterns, ATT-L had the greatest values, followed by TOB-L and LOB-L (Table 5.2). No significant group differences in the cross-correlation measures were detected.

Significant task main effects for stance phase hip-knee DP values were detected between all pair-wise comparisons ($p < 0.05$; Fig 5-3a), but no significant group differences were detected. For knee-ankle inter-joint coordination variability in stance phase, there was a significant group by task interaction ($p = 0.01$). CONC showed significantly greater knee-ankle DP values than NORM during TOB and ATT ($p = 0.009$ and $p = 0.037$, respectively). Task had significant effects on the knee-ankle DP values in CONC for all pair-wise comparisons ($p < 0.04$). However, significant task effects on the knee-ankle DP values in NORM were only detected between Level and LOB, LOB and TOB, and LOB and ATT ($p < 0.001$; Fig 5-3b)

(a)



(b)

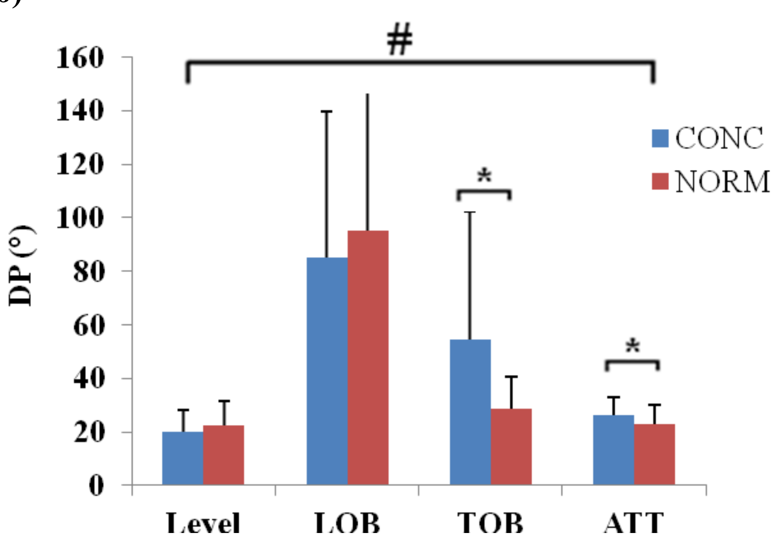
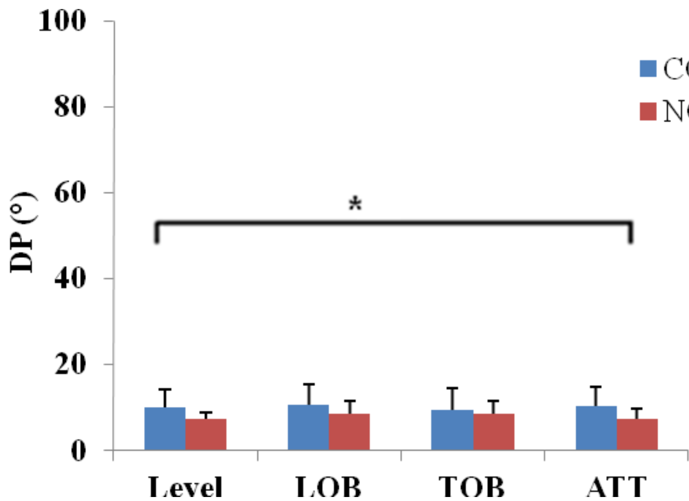


Fig. 5.3. Deviation phase (DP) values of (a) hip-knee and (b) knee-ankle inter-joint coordination for NORM and CONC groups in stance phase of different tasks. (* indicates significant group difference; # indicates significant task effects)

In the swing phase, CONC showed overall greater DP values in hip-knee and knee-ankle inter-joint coordination compared to NORM ($p < 0.007$; Fig 5-4). Significant task main effects were detected for knee-ankle inter-joint coordination DP values between Level and LOB, Level and TOB, LOB and ATT, and TOB and ATT ($p < 0.009$; Fig 5-4b).

(a)



(b)

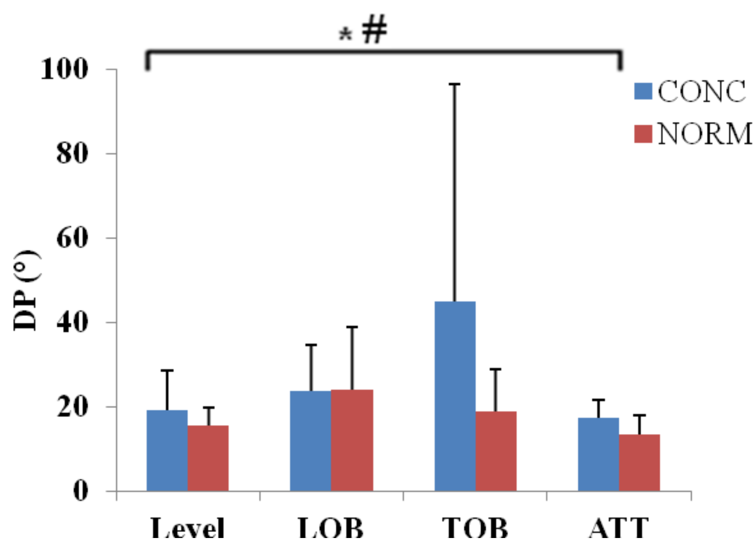


Fig. 5.4. Deviation phase (DP) values of (a) hip-knee and (b) knee-ankle inter-joint coordination for NORM and CONC groups in swing phase of different tasks.
(* indicates significant group difference; # indicates significant task effects)

Table 5.1. Mean and standard deviation of spatial orientation for the leading limb (LOB) and trailing limb (TOB) foot placements during obstacle crossing and percentage of correct answers during walking with a concurrent attention test.

Mean (SD)	CONC	NORM	*p* values
TOB toe obstacle distance	28.3(4.8)	28.2(4.3)	0.961
LOB heel obstacle distance	24.8(6.6)	26.7(4.1)	0.236
TOB toe obstacle clearance	19.9(6.2)	19.1(5.3)	0.633
LOB toe obstacle clearance	15.0(2.9)	15.8(4.3)	0.462
Correct answers (%)	90.9(14.5)	95.4(6.3)	0.174

Table 5.2. Root mean square (RMS) differences (°) and cross correlation coefficients (r) in hip-knee and knee-ankle CRP patterns over a gait cycle.

	CONC			NORM			*p* values
Mean(SD)	LOB-L	TOB-L	ATT-L	LOB-L	TOB-L	ATT-L	
RMS difference							
Hip-Knee	83.4	44.9	18.2	72.1	41.2	16.5	$p_g = 0.003$
	(22.2)	(10.4)	(6.6)	(13.4)	(11.5)	(7.5)	$p_c < 0.001$
Knee-Ankle	114.1	68.7	30.9	106.5	48.8	22.8	$p_g = 0.012$
	(34.8)	(29.6)	(33.8)	(33.8)	(15.6)	(7.5)	$p_c < 0.001$
Cross correlation							
Hip-Knee	0.95	0.89	0.98	0.96	0.91	0.98	$p_c < 0.001$
	(0.02)	(0.07)	(0.22)	(0.01)	(0.06)	(0.01)	
Knee-Ankle	0.42	0.53	0.85	0.38	0.68	0.88	$p_c < 0.001$
	(0.25)	(0.21)	(0.09)	(0.15)	(0.16)	(0.09)	

p_g : group main effect; p_c : task main effect.

Discussion

The pattern and variability of lower limb inter-joint coordination of healthy and concussed young adults during level walking, obstacle crossing, and walking with a concurrent cognitive task were investigated in this study. The goal was to understand how neuromuscular control over the locomotor system would be affected in individuals suffering a concussion. The findings suggested that concussion group had significantly greater values than the control group for: (1) RMS differences between single-task level walking and obstacle crossing or dual-task walking in hip-knee and knee-ankle inter-joint coordination; (2) the variability of knee-ankle inter-joint coordination in the stance phase obstacle crossing trailing limb and dual-task walking; and (3) the variability of hip-knee and knee-ankle inter-joint coordination in the swing phase for both limbs in obstacle crossing and dual-task walking.

When compared to level walking, the pattern changes in hip-knee and knee-ankle inter-joint coordination induced by the obstacle or the concurrent cognitive tasks in concussed individuals were greater than that in healthy controls. This may indicate that concussion influences the ability of modulating inter-joint coordination patterns to accommodate perturbations during locomotion. Such differences may be explained, in part, by the concussion-induced limitations in reallocating the capacity or resources of the central nervous system (CNS) [116]. Since lower limb inter-joint coordination in important for the maintenance of dynamic balance [6,7], the inability to select appropriate coordination patterns accordingly may further contribute to the gait characteristics observed in the concussed individuals.

The greater supporting limb inter-joint coordination variability in the CONC group compared to NORM could be a contributing factor to gait imbalances reported in concussed individuals when negotiating an obstacle or performing a concurrent cognitive task while walking [114,117,120,121,125]. Similarly, the elevated inter-joint coordination variability of the swing limb in concussed individuals may explain their greater variation in the foot-obstacle clearance and high tripping incidences as reported previously [117,120]. Although a certain amount of variability is required for successful movement transitions to accommodate external constrains, excessive or insufficient variability is also considered to be associated with the presence of injuries, and risks of injuries or diseases altering motor control patterns [18]. Since a robust balance control of the supporting limb and a precise trajectory control of the swing limb are necessary for a safe ambulation, the excessive variability of inter-joint coordination observed in concussed subjects may reflect an unstable neuromuscular control during locomotion, which leads to a poor balance control and increased tripping risk. The greater variability of inter-joint coordination in the concussed subjects observed in this current study may also indicate that greater neuromuscular control of the locomotor system is necessary for them to accommodate gait perturbations.

It has been shown previously that movement control strategies utilized during obstacle crossing were different between the leading and trailing limbs [27]. The current study also found that changes in the patterns of hip-knee and knee-ankle inter-joint coordination of the leading limb were greater than that of the trailing limb. A stable support is required during the stance phase of the trailing limb to ensure a safe crossing of the leading limb. This results in low trailing limb inter-joint coordination variability. Because there is a lack of visual feedback while the trailing limb is crossing the obstacle, it is reasonable to expect that the tailing limb demonstrates a greater variability in inter-joint coordination during the swing phase than the leading limb.

The pattern changes in hip-knee and knee-ankle inter-joint coordination induced by the obstacle (OB-L) was found to be greater than dual-task walking (ATT-L). Similarly, the variability of hip-knee and knee-ankle inter-joint coordination was greater during obstacle crossing when compared to dual-task walking. The selected cognitive task requires additional brain activity in the prefrontal lobe due to the verbal and cognitive processes for the question and answer tasks (mental demand) [130]. By contrast, obstacle crossing requires not only modifications of hip and knee joint motion (physical demand) [128] but also a visuospatial attention when interacting with the obstacle (mental demand) which engages the posterior parietal lobe [129]. It is reasonable to expect that the movement control of the lower limbs to be altered to a greater extent to respond a physical gait perturbation than a concurrent cognitive task during level walking.

It has been suggested that the CNS selects an efficient control strategy that exploits the lower limb dynamics to accomplish different walking conditions [128], especially the control of proximal joints (hip and knee) [131]. In this study, the patterns of hip-knee and knee-ankle inter-joint coordination were both changed when accommodating different tasks. Tasks differences influenced the variability of hip-knee and knee-ankle inter-joint coordination during the stance phase as well as the variability of knee-ankle inter-joint coordination during the swing phase. Adjustments to accommodate the obstacle crossing and dual-tasks were observed at both the proximal and distal joints during the stance phase and at the distal joints during swing. Although proximal joints play a greater role in balance control and modulating walking speed during walking when compared to distal joints [84,103], the adjustment of distal joints might be also important for accommodating more complex walking tasks and the fine tuning of knee-ankle inter-joint coordination during the swing phase, which could be critical for safe foot lifting and forward progression.

In summary, this study indicated that the ability to select appropriate coordination patterns and steadiness of the neuromuscular control could be affected by a concussion, which may contribute to gait control variability. Examining and re-educating the lower limb inter-joint coordination of concussed individuals during perturbed walking may be an effective strategy to improve their gait performance. While both hip-knee and knee-ankle inter-joint coordination are critical for accommodating obstacle crossing and dual-task walking in the stance phase, the knee-ankle inter-joint coordination seems to be responsible for the fine tuning of neuromuscular control in the swing phase.

Bridge

Chapter V examined a neurological disturbance effects on the neuromuscular control. The effects of increased physical and mental demands during walking were also demonstrated.

Chapter VI was a further investigation on the properties of the neuromuscular control with a difference level of perturbations, a musculoskeletal impairment. Specifically, since the proximal joints seemed to be most critical for balance control and adjusting walking speeds during walking, we examined the inter-joint coordination in patients with total hip arthroplasty to help us identify the effect of a musculoskeletal disturbance on neuromuscular control.

CHAPTER VI

INVESTIGATION OF THE EFFECT OF AN ISOLATED JOINT DYSFUNTION ON INTER-JOINT COORDINATION - A STUDY OF TOTAL HIP ARTHROPLASTY

This chapter is adopted with permission from Chiu, SL.; Lu TW; Chou, LS. Titled: Altered inter-joint coordination during walking in patients with total hip arthroplasty. *Gait and Posture*, 2010; 32:656-660. Copyright 2010, Elsevier. Dr. Li-Shan Chou contributed substantially to this study by providing critiques and reviewed the manuscript before submitting. Dr. Tung-Wu Lu was consulted for the programming problems during data analysis. I was the primary contributor to the data collections, data analysis, implementation of the procedure, and writing.

Introduction

Total hip arthroplasty (THA) is a common surgery performed in patients with hip osteoarthritis to effectively relieve pain and regain joint functions. Although THA relieves pain and improves mobility, studies assessing gait patterns after THA demonstrated that patients walk with a slower walking speed, smaller step length, reduced hip and knee motions and moments when comparing to healthy individuals [132-134]. Bilateral hip motion asymmetry, abnormal pelvic obliquity, and increased lateral trunk displacement during walking were also found in THA patients after surgery [135-

137]. These suggest that hip osteoarthritis induced muscle atrophy and weakness, functional disability, and joint stiffness may continue to persist after surgery and affect movement control [132,138,139].

Previous gait analyses of THA patients were mainly focused on the kinematics and kinetics of individual joints [132-137]. However, human gait is a complex task requiring a precise end-point movement control accomplished by a multi-joint coordination [4]. Inter-joint coordination relates motions of two joints, implying the essential timing and sequencing of neuromuscular control over biomechanical degrees of freedom for a smooth and efficient movement. Studies have shown that a certain amount of variability is required for the transition between movement conditions or gradual adaptation to a new movement pattern [17,32]. However, a reduced variability could be recognized as lack of flexibility [18,32], and an excessive variability may also reflect the presence of injuries, risks of injuries or diseases altering motor control patterns [25,26,34,47]. With the use of an asymmetrical leg loading, Haddad el al. [16] were able to demonstrate modifications in the inter- and intra-limb coordination and suggested the need of examining inter-joint coordination in pathological gait analysis.

Continuous related phase (CRP), the difference in four quadrants arctangent phase angles between two joints or segments, has been used to investigate the inter-joint or inter-segmental coordination pattern and variability in various activities [18,19,25,26,27]. The phase portrait provides the velocity as a function of position for each joint or segment and defines its position and direction of motions across multiple points of a gait cycle. As it has been indicated that the afferent fibers in muscle receptors work most efficiently by sensing joint position and velocity, CRP is recommended as a useful tool to quantify the inter-joint coordination due to its usage of joint position and velocity on constructing a phase portrait [28]. CRP can provide a continuous measurement of the interaction between segments or joints throughout the entire gait cycle to assess the overall profile of coordination during walking [18].

It is unclear that how or whether the unilateral hip joint dysfunction will alter the inter-joint coordination of the surgical and non-surgical limbs of THA patients. Investigating the inter-joint coordination during gait would help us to identify changes in neuromuscular control strategies due to joint dysfunctions. This information could enhance the development of rehabilitation interventions for THA patients through facilitating desired and reliable inter-joint coordination, and CRP could be used as an objective measure to evaluate these training effects [17]. Therefore, the purpose of this study was to examine differences in the pattern and variability of inter-joint coordination during walking between THA patients and matched controls. Due to the hip joint dysfunction, we expected that THA patients would demonstrate different coordination patterns with an excessive variability, and there would be also changes in the inter-joint coordination observed in the non-surgical limb. We hypothesized that, for both surgical and non-surgical limbs, (1) patterns of inter-joint coordination in THA patients would be different from that of matched controls, and the variability of inter-joint coordination in THA patients would be significantly greater than the controls before surgery; (2) the discrepancies in coordination observed between the controls and THA patients would decrease post surgery.

Methods

Two groups of subjects consisted of a total of 30 adults were recruited to participate in this study. Twenty patients underwent unilateral THA (15 male, 5 female, age = 56.5 ± 5.4 yrs, BMI = 32.6 ± 4.0 kg/m^2), and 10 subjects served as age-matched controls (5 male, 5 female, age = 59.9 ± 5.3 yrs, BMI = 26.3 ± 3.9 kg/m^2). Three-dimensional gait analysis was performed to assess gait performance and joint kinematics of THA patients at pre-surgery, 6-week and 16-week post surgery, respectively. Control subjects were tested twice, one month apart, to ensure the test repeatability [134]. Each subject provided a signed consent to the experimental procedure approved by the Institutional Review Board prior to testing.

THA patients were recruited from a local orthopedics hospital. They were scheduled for a unilateral THA due to osteoarthritis and with no prior histories of joint surgery, fracture of lower extremities, or neurological impairment. Their Harris hip scores [140] collected prior to the surgery was 54.7 (±12.4). All patients underwent unilateral THA with either anterior or anterolateral surgical approach and received the same un-cemented Zimmer hip implants. The Zimmer hip implant consisted of an acetabular component with an irradiated polyethylene linen, a femoral stem, and a metal head component (Zimmer Inc., Warsaw, IN). After surgery, all patients followed the same rehabilitation protocols under the supervision of the same physical therapist during the study period [134]. The selection of post-surgery testing times was based on the recommendation of the surgeons. By 6-weeks post surgery, patients could walk independently without the use of assistive devices, and by 16-weeks post surgery, the soft tissues affected by the surgery would be presumable fully healed to allow the patients to resume their daily activities [141,142]. Age-matched controls were recruited by posting flyers in the university and surrounding neighborhood. Control subjects were community-dwelling individuals without lower limb joint surgery, any history of neurologic or musculoskeletal impairment, and stated no incidence of vertigo or arthritis.

During each laboratory visit, subjects were asked to walk along a 10-m walkway with a self-selected pace; and data from three to five successful walking trials were collected. Practice trials were provided to assist subjects in becoming acquainted with the laboratory environment. An eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) was used to collect whole body motion with a total of 29 reflective markers placed on bony landmarks [76]. Maker trajectories were sampled at 60 Hz, reconstructed and filtered using a low-pass, fourth order Butterworth filter with a cutoff frequency of 8 Hz. Gait temporal distance and joint kinematics of bilateral lower extremities were calculated using OrthoTrak kinematic analysis software (Motion Analysis Corp., Santa Rosa, CA).

A custom written Matlab program (version 7.0; The Mathworks Inc., Natick, MA) was used to calculate the CRP. Sagittal plane joint angular velocities were obtained by utilizing the generalized cross-validation spline smoothing (GCVSPL) smoothing and differentiation method [78]. Joint angles and angular velocities were interpolated to 100% of a gait cycle. A phase plot for each joint throughout a gait cycle was generated by plotting the normalized angular positions (θ) along the x-axis and normalized angular velocities (ω) along the y-axis. Angular position and velocity data were normalized to a range of 1 and -1 along both dimensions of the phase plane, as suggested previously to minimize differences in amplitude and frequency [19,52,58]. Phase angles (φ) were calculated as $\varphi = \tan^{-1}(\omega/\theta)$ for each data point of a gait cycle. The CRP, which identifies the coordination between two adjacent joints, was then obtained by subtracting the phase angles of distal joint from that of proximal joint ($\varphi_{hip\text{-}knee}$, $\varphi_{knee\text{-}ankle}$) [28].

Similarity of the CRP patterns and joint angles across the entire gait cycle were quantified with cross-correlation measures and root-mean-square (RMS) differences, which examined differences between the ensemble mean curves of the surgical or non-surgical limb of THA group and the controls. The cross-correlation measures quantified temporal differences in the phase shift, and the RMS differences reported differences in the magnitude and changes in the patterns of relative phases. A high cross-correlation coefficient with a low RMS difference would indicate that the two curves are similar [16]. The variability of inter-joint coordination of each subject was assessed with the average value of all standard deviations calculated for each data point over a gait cycle from all CRP curves, namely the deviation phase (DP), which represented the cycle to cycle variability and compared the systemic inter-joint characteristics within a gait cycle. Therefore, a lower DP indicates a more repeatable coordination between two joints [19].

For all variables examined, neither significant testing time nor limb effects were detected for the control subjects. Therefore, the group averages including both limbs from two visits were obtained and used for comparison with the patient group. A mixed-model analysis of variance with repeated measures was used to detect differences between the surgical and non-surgical limbs of THA group and between THA and control groups, as well as the time effects on gait temporal distance parameters and DP values. All analyses were performed with SPSS 14.0 (SPSS Inc., Chicago, IL).

Results

At pre-surgery and 6-week post surgery, gait velocities of the THA group (1.05 ± 0.25 m/s and 1.09 ± 0.22 m/s, respectively) were significantly slower ($p < 0.01$) than that of the controls (1.29 ± 0.17 m/s). However, no significant differences were detected between THA patients and controls at 16-week post surgery. Significant time effects on gait velocity were detected for THA patients at 16-week post surgery (1.23 ± 0.15 m/s) when compared to pre-surgery and 6-week post surgery ($p \leq 0.005$).

Graphical deviations in hip-knee and knee-ankle CRP patterns between THA patients and controls were most noticeable during the stance phase for the surgical limb (Fig. 6.1). Quantitative assessment of these graphical similarities is demonstrated with RMS differences values (Fig. 6.2). For both surgical and non-surgical limbs, RMS differences between THA patients and controls were greater in hip-knee and knee-ankle CRP values [range: 14.4˚ – 29.4˚] (Fig. 6.2a) than in individual joint angles [range: 0.6˚- 4.9˚] (Fig. 6.2b). Average RMS differences of the hip-knee and knee-ankle CRP values between THA patients and controls decreased gradually after surgery, changing from 29˚ and 28.3˚ at pre-surgery to 21.2˚ and 15˚ at 16-week post surgery for surgical and non-surgical limbs, respectively. The surgical limb was found to maintain greater RMS differences than non-surgical limb at 6- and 16-week post surgery (Fig. 6.2a). Such changes across testing times were not observed in the RMS differences of individual joint angles (Fig. 6.2b). Cross-correlation measures between THA patients and controls were overall strong for CRP patterns and joint angles of both limbs, with r^2 values greater than 0.94 (Fig. 6.3). Average correlation coefficients for the hip-knee and knee-ankle CRP patterns of both limbs were noticed to increase after surgery, changing from 0.95 at pre-surgery to 0.99 at 16-week post surgery (Fig. 6.3a).

Both limbs of THA group demonstrated higher DP values of the hip-knee and knee-ankle CRP curves when comparing to controls at pre-surgery and 6-week post surgery (Table 6.1). Of the surgical limb, significant differences between THA patients and controls in hip-knee DP values were detected at pre-surgery (p = 0.019) and in knee-ankle DP values at pre-surgery (p = 0.008) and 6-week post surgery (p = 0.036). In addition, the knee-ankle DP values of the surgical limb were found to be greater than that of the non-surgical limb at pre-surgery (p = 0.026). Significant time effects were also detected in DP values of THA patients. Compared to pre-surgery and 6-week post surgery, THA patients demonstrated significantly smaller hip-knee DP values for both limbs and significantly smaller knee-ankle DP values for surgical limb ($p \leq 0.02$) at 16-week post surgery.

Table 6.1. DP values for CRP curves of controls and THA

	Controls	**THA**					
		Non surgical			**Surgical**		
		Pre-surgery	6-week post surgery	16-week post surgery	Pre-surgery	6-week post surgery	16-week post surgery
Hip-Knee	30.6 (7.2)	41.1 (17.9)	40.1 (28.8)	26.9 $^{\dagger\,\#}$ (10.6)	53.4 * (25.8)	45.4 (25.2)	30.9 $^{\dagger\,\#}$ (11.3)
Knee-Ankle	38.5 (6.3)	48.3 (18.9)	47.1 (25.7)	37.8 (21.6)	63.8 *§ (27.2)	53.6 * (21.0)	42.6 $^{\dagger\,\#}$ (15.5)

* significant differences between THA patients and controls;
§significant differences between surgical and non-surgical limbs;
† significant time effects when comparing to pre-surgery;
$^{\#}$ significant time effects when comparing to 6-week post surgery

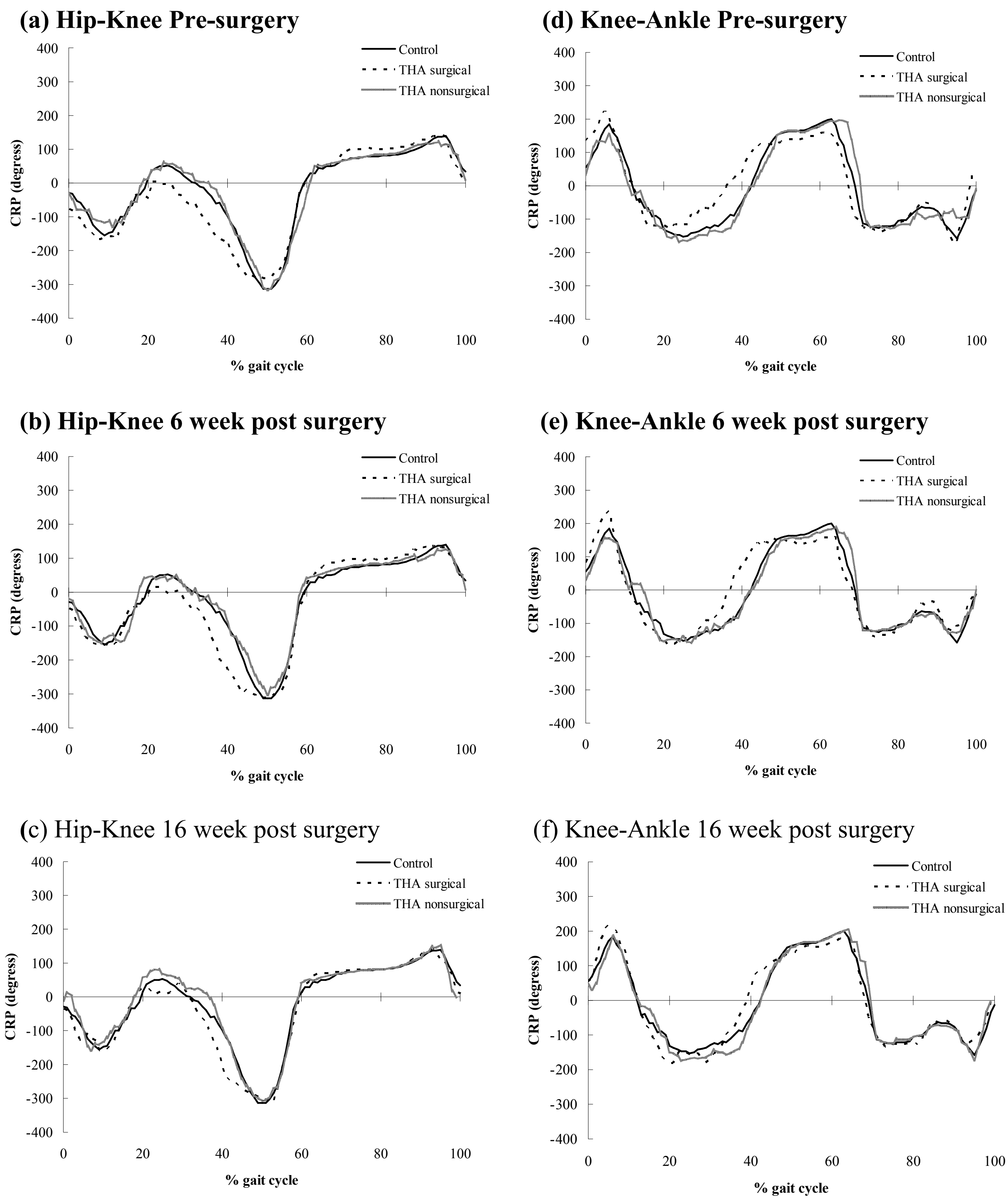


Fig. 6.1. Ensemble mean hip-knee and knee ankle CRP curves of THA and control groups through a gait cycle at pre-surgery [(a) & (d)], 6-week post surgery [(b) & (e)], and 16-week post surgery [(c) & (f)]. The solid black line represents the controls, the gray line represents the non-surgical limb of THA group, and the dash line represents the surgical limb of THA group.

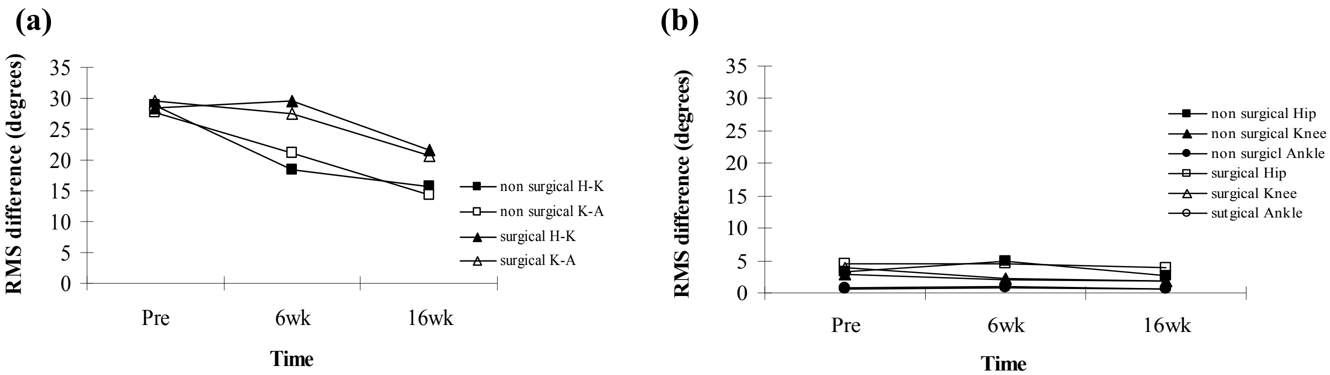


Fig. 6.2. Between-group RMS differences (degrees) in (a) hip-knee and knee-ankle CRP patterns and (b) hip, knee, ankle joint angles of both surgical and non-surgical limbs over a gait cycle.

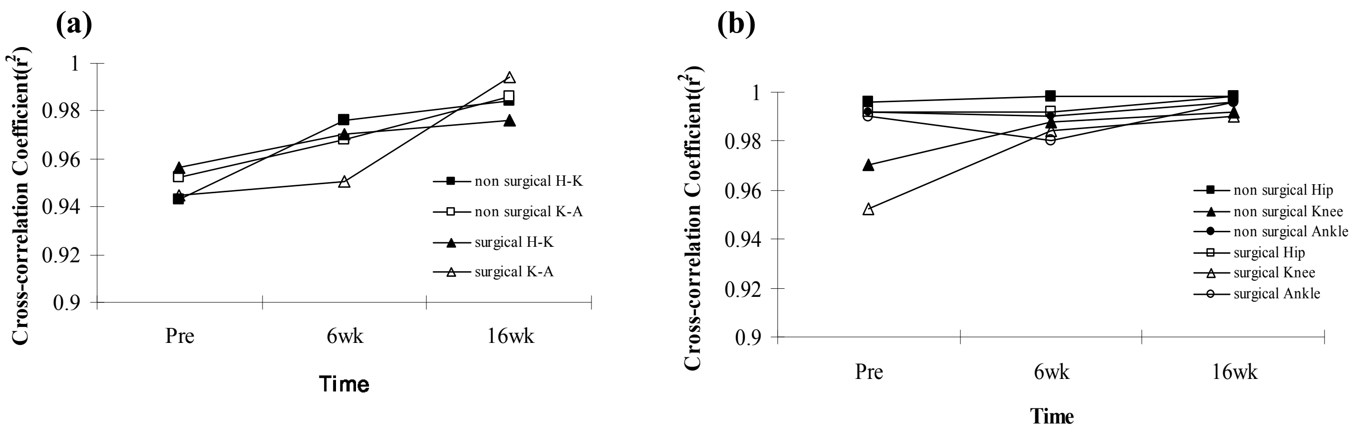


Fig. 6.3. Between-group cross-correlation coefficients (r^2) in (a) hip-knee and knee-ankle CRP patterns and (b) hip, knee, ankle joint angles of both surgical and non-surgical limbs over a gait cycle.

Discussion

In this study we investigated the pattern and variability of inter-joint coordination of surgical and non-surgical limbs of THA patients during walking. We hypothesized that the pattern and variability of inter-joint coordination in THA patients would be different from that of healthy controls. Our findings suggested that adjustments in inter-joint coordination of THA patients occurring at both surgical and non-surgical limbs in responses to a deteriorated hip joint prior to surgery**.** These adjustments were mainly accomplished by changing the relative phase angles between the two adjacent joints (RMS differences), especially for the surgical limb. The cross-correlation measures showed that there were slightly changes in temporal patterns of hip-knee and knee-ankle CRP over a gait cycle. After surgery, the discrepancy in coordination between THA patients and controls had gradually diminished as noted by decreases in RMS differences and increases in cross-correlation coefficients (Figs 6.2 & 6.3). These findings imply that THA patients were adopting a coordination pattern that is closer to that of healthy controls and support our hypothesis that there is a continuous recovery of inter-joint coordination pattern in THA patients after surgery.

Although there were also differences in cross-correlation measures and RMS differences of individual hip, knee and ankle joint motion, these differences were relatively small comparing to that in CRP patterns. Neither there were trends of time effects observed (Figs 6.2 & 6.3). This suggests that examining individual joint motion along might be insufficient to monitor recovery in gait function of THA patients. Since the inter-joint coordination is essential to the end-point control of the limb and the maintenance of dynamic equilibrium during walking [6], investigating the inter-joint coordination of THA patients may provide more insights on the neuromuscular changes contributing to their gait deviations [132-139].

The variability of inter-joint coordination was significantly affected by THA. Prior to the surgery, both hip-knee and knee-ankle coordination demonstrated a higher variability as an effect of the deteriorated hip joint, especially in the surgical limb. This higher variability in THA group could be explained as a transition or an adaptation of a new movement pattern [18,32]. It could be that the center nervous system was developing a new coordination strategy to accommodate the joint dysfunction, which resulting a greater variability in inter-joint coordination. After surgery, as the joint function had been restored and recovered, these accommodations would gradually diminish. This was indicated by the DP values of THA group are restored close to that of controls at 16-week post surgery.

Our data showed that the coordination of the non-surgical limb was also altered. This could be due to that the dysfunction of the involved hip joint affected the pelvic motion [135-137], which altered the motion control of the non-surgical hip and its coordination with the knee. Furthermore, we found that changes in the CRP pattern and variability for the surgical and non-surgical limbs of THA patients were not the same over time (Fig. 6.2 and Table 6.1). This could support the views of previous studies in demonstrating that there were asymmetric changes in accommodations and variability of bilateral lower extremities coordination when adapting to a unilateral loading or constraint [16,54,143]. These asymmetric changes may contribute to the residual gait problems observed in THA patients.

We observed that the walking speeds of THA patients were slower than that of controls at pre-surgery and 6-week post surgery, as what was reported in previous studies [132-134]. Significant changes in both walking speeds and DP values of THA patients were detected from pre-surgery to post surgery. It is, therefore, reasonable to speculate that changes in DP values could be associated with differences in the walking velocity. To clarify this point, a post-hoc analysis, with the self-selected walking speed as a covariate, was performed to examine the differences between the surgical and non-surgical limbs of THA group and control group as well as the time effects on the DP values. Such analysis revealed the same group and time effects on the DP values, indicating that changes in DP values are not merely associated with changes in the walking velocity.

There are limitations in this study. The pattern and variability of inter-joint coordination were only assessed in the sagittal plane joint motion. Since the primary movement of gait takes place in this plane, the inter-joint coordination could be more robustly controlled and illustrated. Furthermore, changes in inter-joint coordination were monitored only for a short-term post surgery in this study. We noticed that although DP values were similar between both limbs of THA patients and controls at 16-week post surgery, there were still differences in CRP patterns (RMS differences). Studies have shown that gait improvement of THA patients were slow, and they have difficulties in regaining normal walking patterns even one to ten years after surgery [132,133,137]. It still needs to be investigated whether THA patients could fully re-acquire the normal inter-joint coordination patterns.

In conclusion, we used cross-correlation measures, RMS differences, and DP values to quantify differences in the CRP pattern and variability and were able to detect the altered inter-joint coordination in patients with hip osteoarthritis and its changes following a THA. Our findings suggested that there is a short-term reformation of inter-joint coordination during gait in the surgical and non-surgical limbs. Assessing the inter-joint coordination as well as facilitating the coordination in clinical treatments may improve the gait performances of THA patients.

Bridge

Chapter VI investigated the effects of a musculoskeletal impairment on the neuromuscular control. With the use of a pre- and post- surgery study design, we were able to identify the short-term reformation of the neuromuscular controls during walking. Chapter VII summarizes the findings and provides a general discussion as well as a conclusion from all the studies in this dissertation.

CHAPTER VII

DISCUSSION AND CONCLUSION

As poor neuromuscular control of lower extremities causes gait deviations, it remains unclear how the CNS accommodates differently to diverse levels of system perturbations. We investigated a general degeneration of the neuromuscular system (aging), a perturbation at the neurological level, and a perturbation at musculoskeletal level. The aim of this dissertation was to reveal the modifications of the neuromuscular control due to different constraints to identify the potential risks during walking. Studies conducted in this dissertation had provided an evidence for a definite property of our neuromuscular control as well as identified its contribution to gait deviations.

Main Findings

In the first study, gait velocity showed significant effects on the pattern and variability of inter-joint coordination. Greater variability in hip-knee inter-joint coordination was found at slower walking speed when compared to faster walking speed, indicating an increased demand of the neuromuscular control over dynamic balance as the single leg support time increases. When asked to walk with different speeds, the proximal joints (φ hip-knee) played a greater role in accommodating walking speed changes in young adult. However, elderly adults seem to use conservative control strategies by maintaining a similar inter-joint coordination. Such differences in modulation and adaptability between elderly and young adults may be associated with their preferred walking speed and age-related mobility impairments.

In the second study, we found that the variability of inter-joint coordination during walking in elderly adults could be better revealed by the Dynamic Gait Index (DGI). The high variability of hip-knee and knee-ankle inter-joint coordination in the stance phase could be associated with declined balance controls in elderly adults. While the variability differences in hip-knee inter-joint coordination could be due to different walking speed between non-fallers and fallers, a higher variability in knee-ankle inter-joint coordination of the stance limb in fallers may indicate a lack of steady control for body weight support, leading to a conservative or a poor adaptive control with low variability for foot trajectories of the swing limb. Similar to our first study, the proximal joints played important roles in maintaining dynamic balance and modulating walking speeds during walking. However, the fine tuning of coordination between distal joints seemed to be a distinguishing feature between fallers and non-fallers.

In the third study, the ability of selecting appropriate coordination patterns and the steadiness of neuromuscular controls were disturbed by a mild traumatic brain injury (concussion). With the use of different walking tasks, we also found that perturbation on musculoskeletal system (obstacle crossing) requires more changes in inter-joint coordination as compared to a concurrent cognitive task (dual-task). Similarly, the knee-ankle inter-joint coordination seemed to be responsible for the fine tuning of neuromuscular control in the swing limb.

In the fourth study, the adjustments in inter-joint coordination for a single joint pathology (hip osteoarthritis & THA) were observed at both surgical and nonsurgical limbs. The hip-knee and knee-ankle coordination demonstrated a higher variability to compromise the impaired hip joint, especially in the surgical limb. A short-term reformation of the neuromuscular control during walking was also identified. Asymmetric changes or recoveries in accommodations and variability of bilateral lower extremities may contribute to the residual gait problems observed in THA patients.

In summary, a systematic degeneration of neuromuscular system (aging) would limit the pattern adaptability of neuromuscular control. Along with clinical balance measures, poor distal joint control ability was predisposed to a poor balance control and an inability to appropriately respond to a trip or slip during walking, resulting in increased fall risks. However, a perturbation at the CNS level and a perturbation at musculoskeletal level would both exaggerate the changes in coordination pattern and variability, indicating that our neuromuscular control was either out of controls or was accommodating/reformatting the movement coordination in response to these constraints.

Additionally, an increase in musculoskeletal demands presents a greater change to the inter-joint coordination as compare to an increased cognitive demand during walking. The neuromuscular controls over the proximal joints coordination (hip-knee) were critical for accommodating different walking conditions, such as walking speeds, walking and crossing an obstacle, or walking with a concurrent attention task. Moreover, the neuromuscular controls over the distal joints coordination (knee-ankle) were responsible for the fine tuning of a secured movement control for the swing limb. Our observation of increased difficulties in the control of the distal joints coordination in concussed individuals and fall-prone subjects suggested that there is a decline in the ability to adaptively control the distal joints. This may be related to the less complex joint kinematic profile during walking and the greater amount of bi-articular muscles of hip and knee joints, as compared to knee and ankle joints. Overall, findings from studies of this dissertation suggested that clinical efforts for improving gait performances and facilitating reliable coordinative movements should be careful designed in individuals with different levels of neuromuscular system deficiency.

Limitations of the Study

The study results provided some interesting findings regarding to the correlations between inter-joint coordination and gait deviations. Although different levels of neuromuscular deficiency were examined, there are some limitations. First, the pattern and variability of inter-joint coordination were only examined in the sagittal plane motion. Since this is the primary plane of motion during gait, the inter-joint coordination might be more robustly controlled and illustrated. Second, we are the first investigators that utilized EMD technique during CRP computation to fit the DST assumptions. EMD is essentially an adaptive nonlinear filter that removes portions of the original signal based on certain criteria. Applying the EMD basically assumes that the signals are nonlinear and non-stationary. However, the advantage of using EMD is that the signal is derived from itself, and it can filter out the lowest frequency component of the signals that is usually considered as the riding waves. The frequency spectrums of the riding waves range from 0 to 0.5 and varied among different subjects in our data. Such approach is similar to a study presented by Bianchi et al. (1998) using a low-pass FIR filter and a 10-harmonics Fourier series analysis.

For the aging studies in Chapters III and IV, we only examined three discrete gait velocities, and all speeds were self-selected. Such examinations may help us to ensure the observed variations in the inter-joint coordination were within a range of self-selected paces and minimize the interferences in the participant's performance. However, a closer monitoring of walking speed in the future may help reduce individual differences. Second, the elderly subjects in these two studies were those who were active in the community and interested in examining their own balance ability. The classification of fallers based on retrospective falling incidences might not be strict enough to differentiate the fallers from non-fallers.

In the concussion study (Chapter V), the walking tasks were not randomly presented to the subjects. The trials were collected in the same orders as Level, OB, and ATT. Ideally, we should consider the accommodations of subjects to our lab environment and marker setting during testing. However, since our primary interests were comparing the group differences (concussion effect on neuromuscular control), the randomization of the task orders was not much of a concern. Additionally, we found significant task and group interactions and tasks main effects in different orders.

Finally, at the musculoskeletal level perturbation, we only investigated the hip joint deficits. Since the hip joint motions are most critical joint for locomotion timing control, we initiated our investigations with the total hip joint pathology. Also, changes in inter-joint coordination were observed only for a short-term post surgery. By the time of 16-week post surgery, we found the inter-joint coordination ability were still different between control and THA patients. A further follow up investigations on whether THA patients could fully re-acquire the appropriate inter-joint coordination ability for safe ambulation may help us identify the potential recovery of neuromuscular control.

Future Research

The studies included in this dissertation have provided some unique information about the characteristics of the neuromuscular control and its interaction with gait deviations or balance control. CRP seemed to be a useful exploratory tool to identify the characteristics of the neuromuscular control. However, future investigations into the limitations discussed and some other research questions to improve our knowledge are necessary.

The major concern is the planar motion we investigated. Frontal plane motions during walking, such as the control of COM and joints, are associated with the gait deviations and balance control ability as well. To our knowledge, the properties of frontal plane inter-joint and inter-segmental coordination have not been well addressed. Future investigations on the frontal plane inter-joint coordination properties may provide an integrated knowledge about our neuromuscular control and balance.

Other than the hip joints, the knee and ankle joints are also important to support and progress our body during walking. Interestingly, we found that the coordination of the proximal joints and distal joints play different roles during walking. Since the distal joint coordination serves as a fine tuner for the neuromuscular control during walking, it would be important to know how our neuromuscular control would regulate the movements accordingly to different joint deficits in the lower extremities to accomplish the walking task, such as total knee replacement or ankle joint fusion.

The boarders for the certain amount of variability required for the transition between movement conditions or the adaptation to a new movement pattern are still unclear. Variability seems to vary for different joint (proximal and distal), different population, and different tasks. Establishing normative values of the inter-joint coordination variability with large cohort of varying population may allow for further improvements in understanding human neuromuscular control and benefit the clinical interventions.

Finally, the further clinical application of the pattern and variability of inter-joint coordination is needed. The enhancement of clinical assessments and the development of a successful individualized intervention or treatment should help improve the gait performances and reduce the fall risks in the future.

APPENDIX A

BERG BALANCE SCALE

Name __________________________________ Date: _______________________

Grading: Please mark the lowest category that applies.

1. Sitting to standing
   Instruction: Ask the patient to please stand up. Try not to use hands for support.
   (4) able to stand, no hands and stabilize independently
   (3) able to stand independently using hands
   (2) able to stand using hands after several tries
   (1) needs minimal assist to stand or to stabilize
   (0) needs moderate or maximal assist to stand ________

2. Standing unsupported
   Instruction: Stand for 2 minutes without holding on to any external support.
   (4) able to stand safely 2 minutes
   (3) able to stand 2 minutes with supervision
   (2) able to stand 30 seconds unsupported
   (1) needs several tries to stand 30 seconds unsupported
   (0) unable to stand 30 seconds unassisted ________

IF SUBJECT IS ABLE TO STAND 2 MINUTES SAFELY, SCORE FULL MARKS FOR SITTING UNSUPPORTED. PROCEED TO POSITION CHANGE STANDING TO SITTING.

3. Sitting unsupported feet on floor
   Instruction: Sit with arms folded for 2 minutes.
   (4) able to sit safely and securely 2 minutes
   (3) able to sit 2 minutes under supervision
   (2) able to sit 30 seconds
   (1) able to sit 10 seconds
   (0) unable to sit without support 10 seconds ________

4. Standing to sitting
   Instruction: Please sit down.
   (4) sits safely with minimal use of hands
   (3) controls descent by using hands
   (2) uses back of legs against chair to control descent
   (1) sits independently but has uncontrolled descent
   (0) needs assistance to sit _______

5. Transfers
   Instruction: Please move from a chair with arm rests to a chair without arm rests and back again.
   (4) able to transfer safely with only minor use of hands
   (3) able to transfer safely with definite need of hands
   (2) able to transfer with verbal cueing and/or supervision
   (1) needs one person to assist
   (0) needs two people to assist or supervise to be safe ________

6. Standing unsupported with eyes closed
   Instruction: Close your eyes and stand still for 10 seconds.
   (4) able to stand 10 seconds safely
   (3) able to stand 10 seconds with supervision
   (2) able to stand 3 seconds
   (1) unable to keep eyes closed 3 seconds but stays steady
   (0) needs help to keep from falling ________

7. Standing unsupported with feet together
   Instruction: Place your feet together and stand without holding on to any external support.
   (4) able to place feet together independently and stand 1 minute safely
   (3) able to place feet together independently and stand 1 minute with supervision
   (2) able to place feet together independently but unable to hold for 30 seconds
   (1) needs help to attain position but able to stand 15 seconds with feet together
   (0) needs help to attain position and unable to hold for 15 seconds ________

THE FOLLOWING ITEMS ARE TO BE PERFORMED WHILE STANDING UNSUPPORTED

8. Reaching forward with outstretched arm
Instruction: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can. Examiner places a ruler at end of fingertips when arm is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance forward that the fingers reach while the subject is in the most forward leaning position.
(4) can reach forward confidently >10 inches
(3) can reach forward >5 inches safely
(2) can reach forward >2 inches safely
(1) reaches forward but needs supervision
(0) needs help to keep from falling ________

9. Pick up object from the floor
Instruction: Pick up the shoe/slipper that is placed in front of your feet
(4) able to pick up slipper safely and easily
(3) able to pick up slipper but need supervision
(2) unable to pick up but reaches 1-2 inches from slipper and keeps balance independently
(1) unable to pick up and needs supervision while trying
(0) unable to try - needs assist to keep from falling ________

10. Turning to look behind over left and right shoulders
Instruction: Turn to look behind you over your left shoulder. Repeat to the right.
(4) looks behind from both sides and weight shifts well
(3) looks behind one side only, other side shows less weight shift
(2) turns sideways only but maintains balance
(1) need supervision when turning
(0) needs assist to keep from falling ________

11. Turn 360 degrees
Instruction: Turn around in a full circle, then turn a full circle in the other direction.
(4) able to turn 360 safely in <4 seconds each side
(3) able to turn 360 safely one side only in <4 seconds
(2) able to turn 360 safely but slowly
(1) needs close supervision or verbal cueing
(0) needs assistance while turning ________

12. Count number of times step stool is touched
    Instruction: Place each foot alternately on the stool. Continue until each foot has touched the stool
    four times for a total of eight steps.
    (4) able to stand independently and safely and complete 8 steps in 20 seconds
    (3) able to stand independently and complete 8 steps in >20 seconds
    (2) able to complete 4 steps without aid with supervision
    (1) able to complete < 2 steps, needs minimal assist
    (0) needs assistance to keep from falling/ unable to try ________

13. Standing unsupported, one foot in front
    Instruction: (Demonstrate) Place one foot directly in front of the other. If you feel that you can't place your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot.
    (4) able to place foot tandem independently and hold 30 seconds
    (3) able to place foot ahead of other independently and hold 30 seconds
    (2) able to take small step independently and hold 30 seconds
    (1) needs help to step but can hold 15 seconds
    (0) loses balance while stepping or standing ________

14. Standing on one leg
    Instruction: Stand on one leg as long as you can without holding on to an external support.
    (4) able to lift leg independently and hold >10 seconds
    (3) able to lift leg independently and hold 5-10 seconds
    (2) able to lift leg independently and hold up to 3 seconds
    (1) tries to lift leg, unable to hold 3 seconds, but remains standing independently
    (0) unable to try or needs assist to prevent fall ________

TOTAL SCORE _______/56

APPENDIX B

DYNAMIC GAIT INDEX

Description:
Developed to assess the likelihood of falling in older adults. Designed to test eight facets of gait.

Equipment needed: Box (Shoebox), Cones (2), Stairs, 20' walkway, 15" wide

Completion:

Time: 15 minutes

Scoring: A four-point ordinal scale, ranging from 0-3. "0" indicates the lowest level of function and "3" the highest level of function. Total Score = 24

Interpretation: **$\leq$ 19/24 = predictive of falls in the elderly**
> 22/24 = safe ambulators

1. Gait level surface _____
Instructions: Walk at your normal speed from here to the next mark (20')
Grading: Mark the lowest category that applies.
(3) Normal: Walks 20', no assistive devices, good sped, no evidence for imbalance, normal gait pattern
(2) Mild Impairment: Walks 20', uses assistive devices, slower speed, mild gait deviations.
(1) Moderate Impairment: Walks 20', slow speed, abnormal gait pattern, evidence for imbalance.
(0) Severe Impairment: Cannot walk 20' without assistance, severe gait deviations or imbalance.

2. Change in gait speed _____
Instructions: Begin walking at your normal pace (for 5'), when I tell you "go," walk as fast as you can (for 5'). When I tell you "slow," walk as slowly as you can (for 5').
Grading: Mark the lowest category that applies.
(3) Normal: Able to smoothly change walking speed without loss of balance or gait deviation. Shows a significant difference in walking speeds between normal, fast and slow speeds.
(2) Mild Impairment: Is able to change speed but demonstrates mild gait deviations, or not gait deviations but unable to achieve a significant change in velocity, or uses an assistive device.
(1) Moderate Impairment: Makes only minor adjustments to walking speed, or accomplishes a change in speed with significant gait deviations, or changes speed but has significant gait deviations, or changes speed but loses balance but is able to recover and continue walking.
(0) Severe Impairment: Cannot change speeds, or loses balance and has to reach for wall or be caught.

3. Gait with horizontal head turns _____

Instructions: Begin walking at your normal pace. When I tell you to "look right," keep walking straight, but turn your head to the right. Keep looking to the right until I tell you, "look left," then keep walking straight and turn your head to the left. Keep your head to the left until I tell you "look straight,“ then keep walking straight, but return your head to the center.

Grading: Mark the lowest category that applies.

(3) Normal: Performs head turns smoothly with no change in gait.

(2) Mild Impairment: Performs head turns smoothly with slight change in gait velocity, i.e., minor disruption to smooth gait path or uses walking aid.

(1) Moderate Impairment: Performs head turns with moderate change in gait velocity, slows down, staggers but recovers, can continue to walk.

(0) Severe Impairment: Performs task with severe disruption of gait, i.e., staggers outside 15" path, loses balance, stops, reaches for wall.

4. Gait with vertical head turns _____

Instructions: Begin walking at your normal pace. When I tell you to "look up," keep walking straight, but tip your head up. Keep looking up until I tell you, "look down," then keep walking straight and tip your head down. Keep your head down until I tell you "look straight,“ then keep walking straight, but return your head to the center.

Grading: Mark the lowest category that applies.

(3) Normal: Performs head turns smoothly with no change in gait.

(2) Mild Impairment: Performs head turns smoothly with slight change in gait velocity, i.e., minor disruption to smooth gait path or uses walking aid.

(1) Moderate Impairment: Performs head turns with moderate change in gait velocity, slows down, staggers but recovers, can continue to walk.

(0) Severe Impairment: Performs task with severe disruption of gait, i.e., staggers outside 15" path, loses balance, stops, reaches for wall.

5. Gait and pivot turn _____

Instructions: Begin walking at your normal pace. When I tell you, "turn and stop," turn as quickly as you can to face the opposite direction and stop.

Grading: Mark the lowest category that applies.

(3) Normal: Pivot turns safely within 3 seconds and stops quickly with no loss of balance.

(2) Mild Impairment: Pivot turns safely in > 3 seconds and stops with no loss of balance.

(1) Moderate Impairment: Turns slowly, requires verbal cueing, requires several small steps to catch balance following turn and stop.

(0) Severe Impairment: Cannot turn safely, requires assistance to turn and stop.

6. Step over obstacle ____
Instructions: Begin walking at your normal speed. When you come to the shoebox, step over it, not around it, and keep walking.
Grading: Mark the lowest category that applies.
(3) Normal: Is able to step over the box without changing gait speed, no evidence of imbalance.
(2) Mild Impairment: Is able to step over box, but must slow down and adjust steps to clear box safely.
(1) Moderate Impairment: Is able to step over box but must stop, then step over. May require verbal cueing.
(0) Severe Impairment: Cannot perform without assistance.

7. Step around obstacles _____
Instructions: Begin walking at normal speed. When you come to the first cone (about 6' away), walk around the right side of it. When you come to the second cone (6' past first cone), walk around it to the left.
Grading: Mark the lowest category that applies.
(3) Normal: Is able to walk around cones safely without changing gait speed; no evidence of imbalance.
(2) Mild Impairment: Is able to step around both cones, but must slow down and adjust steps to clear cones.
(1) Moderate Impairment: Is able to clear cones but must significantly slow, speed to accomplish task, or requires verbal cueing.
(0) Severe Impairment: Unable to clear cones, walks into one or both cones, or requires physical assistance.

8. Steps _____
Instructions: Walk up these stairs as you would at home, i.e., using the railing if necessary. At the top, turn around and walk down.
Grading: Mark the lowest category that applies.
(3) Normal: Alternating feet, no rail.
(2) Mild Impairment: Alternating feet, must use rail.
(1) Moderate Impairment: Two feet to a stair, must use rail.
(0) Severe Impairment: Cannot do safely.

TOTAL SCORE: ___ / 24

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