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Effects of Ankle Immobilization on Lower Extremity Joint Coupling Variability

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EFFECTS OF ANKLE IMMOBILIZATION ON LOWER EXTREMITY JOINT COUPLING VARIABILITY

by

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ABSTRACT

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Human movement is a complex system that encompasses several factors within the body to create coordination. It is not fully understood how the many degrees of freedom (DOF) of the body organize to create movement. Both internal and external influence how the DOF problem works to produce varying movement goals.

The purpose of this study was to better understand the organization patterns of the body by eliminating a DOF. The ankle joint was immobilized to determine if there were differences at the hip and knee joint couplings. It was hypothesized that there would be an increase in movement variability at the hip and knee joints when the ankle was immobilized compared with normal gait.

Joint kinematics have shown that the AFO effectively immobilized the ankle joint by decreasing plantarflexion throughout the gait cycle and restricting ankle inversion/eversion and internal/external rotation. There were also changes seen at the hip and knee joints. Hip flexion decreased and adduction increased through the gait cycle. Internal rotation of the hip decreased in stance and increased during swing phase. There was a decrease in knee extension and adduction during stance phase and an increase in internal rotation during swing. There were no significant changes in coordination variability in any of the nine couplings compared. Further experiments need to be conducted to better understand human movement patterns.

I would like to dedicate this work to my family. Thank you to my mother, father, sister and brother for your constant support and encouragement.

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

NOMENCLATURE

DOF	Degrees of freedom
VC	Vector coding
CRP	Continuous relative phase
EVE	Eversion
INV	Inversion
PF	Plantar Flexion
DF	Dorsiflexion
IR	Internal rotation
ER	External rotation
FLEX	Flexion
EXT	Extension
RMS	Root mean square
AFO	Ankle-foot-orthotic

Chapter 1: Development of the Problem

Background and Rationale

Coordination is the product of innumerous different degrees of freedom (DOF) working together to create movement. As Bernstein (1967) described, DOF of the body (e.g., joints, muscles, and the nervous system) combine with external forces during movement to produce countless patterns. Movement is not simply regulated by one signal to create one response. The human body benefits in many ways by having a large number of DOF that work together. Self-organization of this highly dimensional system is a questions to be answered in understanding coordination (Turvey, 1990).

From a dynamical systems perspective, movement is not pre-determined; it is the body's way of self-organization to transition between states of internal and environmental constraints. Different tasks are controlled under different biological constraints in motor control. The external environment, an obstacle or perturbation, can influence the order of DOF within motor control. The high number of DOF aid in the function of human movement in response to stimuli. If there is a temporary perturbation, all of the internal DOF readjust immediately to preserve the task goal (Turvey, 1990). The dynamical systems theory seeks to understand how the body can organize such a complex system.

The highly complex system involved in coordination leads to inherent variability. Variability of coordination can be used to understand how the body solves the DOF problem under different variables. Traditionally variability has been used as a means to quantify repeatability. In various research problems variability has been viewed as error or noise, which may suggest signs of pathology or injury (Glazier, Wheat, Pease, & Bartlett, 2006). Through the study of dynamical systems, researchers have also seen that variability in human movement is unavoidable and is seen in even the most elite practiced athletes. The dynamical systems theory uses variability to better understand the organization of the DOF that contribute to movement coordination (Davids, Glazier, Araujo, & Bartlett, 2003).

Increased variability within an individual indicates growing instability, which may lead to a shift to a new behavior. On the other hand, low variability within an individual suggests motor output is not able to adapt as well and the behavior may be not be able to cope with change. In such cases, movement patterns may be repeated in such a way that may result in constant stress on cartilage, tendon, and ligaments that could ultimately cause degenerative changes. The redundancy of the system allows for the use of multiple strategies to accomplish any given task (Turvey, 1990). Thus, dynamical systems theory advances our understanding of transitions between movement behaviors, with variability considered not as error but rather as a source of behavioral change through the process of self-organization (Davids et al., 2003).

Researchers have used many different techniques to study coordination and variability of movement from a dynamical systems perspective. They include discrete relative phase, cross correlation, normalized RMS difference, joint timing, continuous relative phase (CRP) and vector coding (VC) (Chiu & Chou, 2012; DeLeo, Dierks, Ferber, & Davis, 2004; Dierks & Davis, 2007; Hamill, Haddad, & McDermott, 2000). The two methods focused on in this paper are CRP and VC. Although both techniques involve the assessment of coordination by the quantification of phase plane trajectories, the phase planes constructed with these two techniques are fundamentally different (Hamill et al., 2000; Miller, Chang, Baird, Van Emmerik, & Hamill, 2010). It is important to understand the differences between the two methods as well as the output of research involving coordination variability (Miller et al., 2010).

CRP uses angular position and angular velocity to determine joint coupling relationships. This method has been used to determine joint coupling relationships in symptomatic and non-symptomatic individuals. Chiu and Chou (2012) used CRP to determine differences in walking speed and age on joint coupling. Results from this study showed that older adults with similar gait patterns showed less variability compared with the younger group. They concluded that this could be due to a decrease in motor control in older individuals. Hamill, van Emmerik, Heiderscheit, and Li (1999) also used CRP measures to determine if Q angles can predict injury. The results showed no joint coupling or variability differences in low or high Q angles. Authors concluded that Q angles could not be used to determine structural differences and the effect on gait. Hamill et al. (1999) did a second study in this article to determine if there were differences in joint coupling variability in subjects with patellofemoral pain syndrome compared with healthy subjects. They found that joint coupling variability was less in those with injury, similar to the results found in Chiu and Chou (2012), which found the older adults also had less variability compared to a younger population. The constraints experienced by the subjects in each study, although different, both decrease the DOF; Chiu and Chou (2012) suggested a decrease in motor control due to age, while Hamill et al. (1999) suggested the constraint was injury. Both studies show a decrease in joint coupling variability, which supports the dynamical systems theory that variability of joint coupling is representative of normal and impaired subjects.

There are several limitations to CRP. One limitation is the assumption that all data is sinusoidal (Hamill et al., 2000). However, during gait, the ankle and knee joint waveforms violate this assumption, as they do not project simple sine waves. A second limitation of CRP is the process of normalization. Kurz and Stergiou (2002) compared CRP data that was normalized and non-normalized. Position and velocity, two different scaling factors, were used to create a phase plot. Results showed that normalization changed the dynamic qualities of the oscillating segment. Normalization in CRP is used to scale amplitude differences. Researchers from Kurz and Stergiou (2002)state that amplitude differences between segments are accounted for by using the arc tangent function. Therefore, normalization techniques might not be necessary. Normalization methods could alter data and researchers should be aware of these differences when comparing results for future data (DeLeo et al., 2004). The differing views of this technique between researchers provide another limitation of this method for future studies.

Similar to CRP, VC uses phase plots to compare joint coupling relationships. This method uses the vector with respect to the horizontal from two consecutive points in an angle-angle plot (Hamill et al., 2000). VC has been used to determine similar couplings of symptomatic and non-symptomatic subjects that have been seen in CRP. Pohl, Messenger, and Buckley (2007) used VC to determine kinematic coupling between walking and running trials. They found significant differences between joint couplings of walking trials compared with running trials. Running trials demonstrate greater coupling of the rearfoot, forefoot, and shank compared to walking trials. Results from both trials suggest that kinematic coupling of the rearfoot, forefoot, and shank are linked through the ankle complex. In a follow up study Pohl and Buckley (2008) altered foot strike to examine coupling patterns. Joint coupling was in sync for all three foot strike patterns (heel first, forefoot first, and toe first); meaning the joints moved simultaneously in the same direction. This data supports the previous study, that coupling of the lower extremity is controlled through the ankle complex. VC does not require the use of any normalization techniques, which eliminates this problem associated with CRP. However, CRP provides both spatial and temporal data and VC directly provides spatial information (Hamill et al., 2000). Both studies provide interpretation to coordination data. It is important to understand both methods when looking at studies of coordination and variability.

Most of the research involving coordination variability patterns has looked at thigh, shank, and foot relationship. Seay, Van Emmerik, and Hamill (2011) were able to use VC to assess coordination of the trunk in walking and running trials in those with lower back pain. Researchers discovered that those with LBP had a greater ROM and researchers suggest that they are less coordinated in the running trials compared with other groups. Heiderscheit (2002) used joint coupling variability to assess subjects with patellofemoral pain (PFP) with healthy matched controls. VC was used to determine joint coupling variability of the lower extremity including the knee. With the use of VC, researchers were able to determine that variability was only greater in the thigh/leg rotation at heel-strike for both healthy and normal populations. The previous studies have primarily looked at ankle joint couplings. These two studies use VC to look at hip and knee joints in addition to the ankle joint. This offers a wider scope to which joint coupling can be analyzed.

Regardless of the method used to examine joint coupling variability (i.e., CRP, VC, or other) these studies have attempted to determine how physical and/or environmental constraints influence movement coordination. As of recent, there have been no studies that have looked at coordination variability by purposefully freezing a DOF. A common intervention used to control position and motion of the ankle, compensate for weakness, or correct for anatomical deformities are devices such as an ankle-foot orthotic (AFO). AFOs are frequently used in the treatment of disorders affecting muscle function such as stroke, spinal cord injury, muscular dystrophy, cerebral palsy, polio, multiple sclerosis and peripheral neuropathy. Research has shown that AFO use in children with spastic diplegia results in an increase in peak knee extension moments during the stance phase of gait and aids in heel strike and dorsiflexion (Buckon et al., 2004). In subjects with spastic cerebral palsy, two different AFOs were used and compared. Both the standard AFO and the hinged AFO increase stride length and reduce plantar flexion (Lam, Leong, Li, Hu, & Lu, 2005). Nair, Rooney, Kautz, and Behrman (2010) conducted a study with healthy subjects and the use of a unilateral AFO. Results showed the AFO restricts ankle movement and proximal hip joint excursion. Increase in hip flexion and decrease in dorsiflexion limited the smooth transfer of body weight onto the stance limb. Although this study looked at lower body kinematics in healthy subjects, they did not include coordination variability in the assessment. Coordination variability is an important assessment tool that aids in understanding human movement and should be investigated further.

Statement of the Problem

While the influence of AFOs in altering joint kinematics during gait has been well documented, the acute effects of such devices on coordination variability remains unknown in a healthy population. By effectively eliminating a degree of freedom using the AFO, it is reasonable to assume that coordination will change as the individual searches for a new behavior pattern through self-organization. How long it takes an individual to adapt to this constrained system and develop a new behavior pattern remains unknown, however, a first step to understanding coordination variability is to understand the acute effects.

Statement of Purpose

The purpose of this study will be to determine the acute effects of restricted ankle sagittal plane movement on coordination variability of hip and knee joint couplings.

Null Hypotheses

Joint coupling variability will be unchanged when the ankle is restricted in the sagittal plane compared to normal unrestricted movement.

Research Hypotheses

Coordination variability will increase at the hip and knee when the ankle is restricted in the sagittal plane compared to normal unrestricted gait.

Limitations of the Study

- Each participant's physical activity level may vary, thereby affecting the kinematics at the joints of the lower extremity.
- Physiological variations are inherent during movement and manifest in joint coordination variability measures.

- Only a single control parameter will be manipulated (ankle joint) while others (fatigue, environment, etc.) will remain constant.

Delimitations of the Study

- The population of this study will be delimited to 20 recreationally active individuals between the ages of 18-35 years old.
- The order of gait conditions is delimited (immobilized vs non-immobilized).
- Equipment used will be delimited to an 8-camera motion analysis system (Vicon, Centennial, CO) for the collection of marker coordinate data (200 Hz) with an accuracy of < 1mm.
- Data reductions will be delimited to the selection and time normalization of gait phases, calculation of joint angles, CRP and VC measures.
- Analysis of joint coupling combinations will be delimited to knee and hip.

Assumptions of the Study

- All participants will answer all questions honestly on the pretest screening form.
- All participants will closely follow instructions.
- The high-speed cameras will be accurately calibrated for each participant throughout the experiments.
- Skin mounted markers and marker clusters represent the underlying motion of the skeletal system.

Significance of the Study

This study may provide a better understanding of how the motor system adapts to an organismic constraint (AFO) by freezing a distal degree of freedom. This information may provide insight into the flexibility of the motor system to cope with perturbations and dissipate stress at specific points of transition throughout the gait cycle.

Operational Definition of Terms

- Dynamical Systems Theory: Multidisciplinary, systems-led theoretical framework to describe systems that are constantly changing and evolving through mathematical expressions.
- Coordination: The relative timing and magnitude of kinematic variables describing between two or more adjacent or non-adjacent segments.
- Joint Coupling: Simultaneous coordination between two joints throughout a movement cycle (e.g., stride cycle).
- Self-organization: The formation of movement patterns is a function of the cooperation of all the subsystems and their interaction with the environment.
- Phase Plane: Representation of the behavior of the dynamic system. Typically in the form of a 2-dimensional plot of position of the time series versus the first derivative (velocity).
- Phase Angle: Four-quadrant arctangent of a segment or joint phase portrait.
- Continuous relative phase: Method of assessing joint coupling coordination by comparing the phase angle of two segments or joints of interest.
- Vector coding: Method of assessing joint coupling coordination by quantifying the angle-angle plot of two segments or joints of interest.
- Segment angle: The angle of inclination of segment relative to the right horizontal.
- Joint angle: The relative angle formed between two segments of the body.

- Ankle-foot orthosis: Brace (usually plastic) that surrounds the ankle and part of the foot which is used to control the position and motion of the ankle.

Chapter 2: Review of the Literature

Introduction

The area of dynamical systems originated almost a century ago in the mathematical and physical sciences as a means to explain a system that changes over time (Davids et al., 2003). In the last 30 years, this concept of nonlinear dynamics has expanded to include many diverse applications including human movement and neurophysiology (Kugler, 1980). Kugler (1980) proposed a new perception of motor skill performance within movement and laid the foundation for Dynamical Systems Theory (DST), as it is currently applied.

This literature review will provide an overview of DST and the use of this method to investigate kinematic changes in human movement. The various aspects of human movement examined by DST will be presented as well as the techniques used to assess and describe the motion observed.

Coordination

Human movement is complex and involves the interaction of countless components or degrees of freedom (DOF). As Bernstein (1967) described, multiple DOF of the body (e.g., joints, muscles, and the nervous system) combine with external forces during movement to produce countless patterns to accomplish a movement goal. The production of stable, coordinated movements thus requires that the DOF be organized in a sequential fashion by the neuromuscular system. However, the redundant nature of the human neuromuscular system creates an indeterminate problem in which it is possible for multiple combinations of DOF to achieve the same goal and for the same DOF to achieve a different goal (Turvey, 1990). Turvey (1990) defined this process of organization redundant. The formation of coordinative structures requires the collective interaction of single muscles and neuropathways and allows for the DOF to be reduced so that functional movement patterns can be achieved (Bernstein, 1967).

DST is based on the assumption that variations in movement occur because of the neuromuscular system's response to local and global constraints (i.e., biological, biomechanical, or environmental demands) (Stergiou, Jensen, Bates, Scholten, & Tzetzis, 2001). Since human movement patterns are complex in nature they involve the coupling of multiple DOF. When a perturbation occurs within a system, the DOF must be reorganized by the neuromuscular system to achieve a new functional outcome. The inability to adequately reorganize the DOF would thus result in abnormal movement patterns or instabilities (Davids et al., 2003). The principles of this method have been effectively utilized to examine the coordination involved within a variety of different movement patterns such as injury, foot-strike pattern and speed (Burgess-Limerick, Abernethy, & Neal, 1993; Clark & Phillips, 1993; Hamill et al., 2000; Schoner, 1990; Schoner & Kelso, 1988; Stergiou et al., 2001).

Variability

Traditionally, variability is synonymous with noise or error. Davids et al. (2003) explored the dynamical systems approach to variability and suggested that error is neither good nor bad. While there are several combinations used by the body to produce a movement, it is suggested that these combinations are related to amount of coordination variability within a movement. A skillful athlete can exploit the many DOF of the motor system. They are able to freeze or unfreeze DOF in specific movement tasks. Davids et al. (2003) provides an example of a skilled gunman that freezes the DOF of the distal arm while allowing proximal segment to be highly variable. Although skilled athletes may be able to control variability, they cannot eliminate variability from trial-to-trial, as movement patterns cannot be duplicated. As the human body ages the complexity of the motor system decreases. This can be represented by lower coordination variability, as there are not as many motor strategies that can be utilized to accomplish a goal. This phenomenon has also been observed in subjects with injury (Chiu & Chou, 2012; Hamill et al., 1999; Heiderscheit, 2002). Turvey (1990) explains that as one segment is synchronized with another segment, such as one leg synchronized with another leg, the DOF are decreased as the motion of one joint becomes coupled with the other movement. Understanding patterns of variability within all aspects of motion is an important step in understanding human movement. Through the use of equations and models, researchers can further explore the DOF problem.

Measures of Coordination Variability

A key theoretical concept integral to many human movements is the kinetic chain. This phenomenon is defined as a proximal-distal linkage system through which energy and momentum are transferred sequentially (Robertson & Winter, 1980; Winter, Quanbury, & Reimer, 1976). Proper gait biomechanics involves synchronous movements of all of the components of the kinetic chain. The foot's many functions include adaptation to uneven terrain, proprioception for proper position and balance, and leverage for propulsion. During the gait cycle, foot motion facilitates, and can be affected by, compensatory movement of the other bones and joints in the lower extremity. Improper alignment from the lumbar spine and lower limb below can alter mechanics and lead to injury.

To gain a better understanding of how body segments are coordinated, biomechanists should refrain from habitually reducing time-series data to discrete kinematic measurements and their corresponding time histories, as this procedure fails to capture the dynamic nature of the movement (Glazier, Davids, & Bartlett, 2003). Instead, as a precursor to more sophisticated kinetic analyses, segmental interactions could be examined by analyzing sets of time series data obtained from adjacent body segments or joints with the following qualitative and quantitative techniques commonly used by dynamical systems theorists in motor control research (Hamill et al., 2000; Mullineaux, Bartlett, & Bennett, 2001; Sparrow, 2001).

The coupling variability of joints and segments has been evaluated with various measures and in numerous conditions. Continuous relative phase (CRP), and vector coding (VC) are among the many methods included in this review, and are identified as common methods to quantify joint coordination. Although both of these methods have been used to look at joint coupling, they must both be understood individually to understand their differences.

Continuous Relative Phase

CRP is a method commonly used by dynamical systems theorists to evaluate coordination over time. CRP is a continuous measure of coordination between two oscillatory components, such as body segments (Burgess-Limerick et al., 1993; Hamill et al., 1999). A CRP of 0° corresponds to in-phase coupling, meaning the phase angles are identical, and a potentially stable couple pattern exists as they are behaving similarly. As CRP moves away from 0° in either a positive or negative direction, the two motions become more out-of-phase and are behaving in a less similar fashion. CRP is calculated by first generating phase portrait of two segments or two joints, which is a plot of each segment's angular position versus its first derivative (Kurz & Stergiou, 2002). The four-quadrant arctangent phase angle is then calculated for each segment. Finally, CRP can be determined by subtracting the phase angle of one segment from the phase angle of another segment (Chiu & Chou, 2012; DeLeo et al., 2004; Hamill et al., 2000).

When utilizing CRP to assess coordination there is often a concern that one segment may dominate the other, which has stemmed the discussion on the need for normalization (Hamill et al., 2000; Kurz & Stergiou, 2002). The two techniques typically used to normalize phase portraits, angle-angle plots, and often CRP values scale the angular displacement and velocity values to a range of ±1 (Kurz & Stergiou, 2002). In the first method, the maximum and minimum values are normalized to values of +1 and -1, respectively. The zero point of these normalized values represents the midpoint of the given range of motion (van Emmerik & Wagenaar, 1996). In the second method, the absolute values of the maximum and minimum values are normalized to ±1 (Burgess-Limerick et al., 1993). This allows zero angular displacement and zero angular velocity to be maintained at the origin, thus centering the plot at the point rather than elsewhere. The first method better maintains the spatial properties of the movement, while the second better maintains the spatial layout (Kurz & Stergiou, 2002). Either method can be utilized but the second method should be avoided if noticeable outliers are observed within the data. This would cause the graph to stretch, thus distorting the visual representation of coordination (DeLeo et al., 2004). Another limitation of CRP is the assumption that all data is sinusoidal. This can be problematic in variables such as knee flexion (FLEX) and ankle plantar flexion (PF), which do not satisfy this assumption during activities such as gait (DeLeo et al., 2004; Heiderscheit, 2002).

CRP measures have been used to look at the effects of walking speed and age on inter-joint coordination (Chiu & Chou, 2012). A total of twenty subjects were used in this study and split into groups based on age. Young adults had an average age of 24.7 ± 4.1 years and the older adult subjects had an average age of 71.6± 5.2 years. Subjects were asked to walk barefoot at their self-selected preferred walking speed, slow walking speed, and fast walking speed. Five trials of each speed were collected and used for analysis. Researchers looked at joint coordination between the hip/knee and the knee/ankle. Angular velocities were normalized using minimum and maximum angles. These normalized values were plotted and phase angles were calculated. To resolve issues of non-sinusoidal signals, empirical mode decomposition was applied. This method was used to eliminate any riding waves and uneven amplitudes throughout the joint motion. Each group was able to walk at significantly different speeds for slow, and fast. Results showed that young adults had significant changes in hip and knee joint angles at the three different walking speeds. Joint coupling in older adults did not significantly change at the three different walking speeds. They showed similar joint coupling patterns at different speeds. However, joint coupling variability of individual joint motion showed that older adults had greater variability compared to younger adults. Joint coupling variability of gait speed was significant in both groups, specifically at slower speeds. Researchers of this study suggest that slower walking speeds are more challenging and require more balance and support as single-leg support time is increased. Also, young adults use different strategies, as seen in coupling patterns, to accommodate for changing speeds.

Older adults did not elicit this accommodation in response to speed changes. This study is an important piece of the dynamical systems theory that can be used to further interpret and understand qualitative data.

Hamill et al. (1999) used CRP to investigate coupling relationships in the lower extremity in individuals with prevalence of injury. Prior to this study, CRP had not been compared in subjects with injuries. Hamill et al. (1999) presented two separate studies. Study one compared data from subjects with Q-angles greater than 15° and Q angles less than 15°. These subjects were asked to run across a 35.0 m runway with embedded force plates with locomotors speeds between 3.6 m/s and 3.83 m/s. A total of ten trials were collected and used for analysis. The phase angles evaluated were abduction (ABD)/adduction (ADD), thigh FLEX/extension (EXT), tibial rotation, and foot inversion (INV)/EVE and were normalized the maximum and minimum angles. A CRP scale of 0° indicated an in phase relationship and CRP 180° indicated an anti-phase relationship. Variation was calculated as the standard deviation of each point on the ensemble curve and was quantified by calculating the average standard deviation over the complete profile. Hamill et al. (1999) found that all three couplings were anti-phase throughout the entire support phase in individuals with low Q angles. The greatest amount of variability was found at heel contact to foot flat. Subjects with high Q angles also had data suggesting all three couplings were anti-phase at foot strike. However, couplings in those with low Q angles became more in phase throughout the support phase, unlike subjects with high Q angles. Similar to those with low angles, the greatest amount of variability was found at heel contact to foot flat. CRP variability showed no indication of differences between Q angles. CRP and CRP variability do not suggest significant differences in Q-angles in running conditions. A possible explanation for this is that both groups were not representative of any injury. Research suggests variability differences are seen in those with injury (Davids et al., 2003).

The second study reported in Hamill et al. (1999) compared subjects with symptomatic patellofemoral pain (PFP) and asymptomatic patellofemoral subjects. Here they asked subjects to run on a treadmill with three speeds at 2.5m/s, 3.0m/s, and 3.5 m/s. Coordination variability was assessed similarly to the first study by evaluating INT/EXT rotation of the thigh and tibia to address antagonistic rotations involved with PFP. CRP and CRP variability at speed of 3.0 and 3.5 matched those in the previous study. Joint coupling variability in healthy versus PFP subjects was shown to be less in subjects with PFP. This study also showed that healthy individuals have greater variability. This indicates that those with PFP are repeating segment actions, which may cause greater stress and injury. The authors suggest that greater variability is a factor in non-injury as opposed to prevalence of injury.

In the previous Hamill et al. (1999), CRP data was normalized. However, there has been some debate on the necessity of normalization. Therefore, Kurz and Stergiou (2002) conducted a study to determine if normalization had an effect on CRP calculations. One male subject was used in this study for data collection. The subject ran at a self-selected pace with markers placed on the right leg only. The coupling of the leg and thigh were used for analysis from ten consecutive footfalls. Two techniques were used to normalize the study along with a set of un-normalized data as well. The angular displacement and velocity were scaled to a range of ± 1. The first method was normalized to maximum amplitude and the other was normalized with minimum amplitude values.

Results showed that there were differences between normalized and non-normalized data. Normalization of the data graphically changed the dynamic qualities of the oscillating segments. Researchers proposed that normalization might not be a necessary component when calculating CRP. When normalization is not applied, the authors suggest that amplitude differences may not be a problem as previously indicated. Normalization is used to account for the amplitude differences. However the scale is not uniform for velocity and displacement when normalization methods are applied. Not using normalization techniques would eliminate this as a limitation when employing CRP methodology. However, the assumption of sinusoidal data remains an issue. Although some methods have been used to eliminate this assumption, such as empirical mode decomposition (EMD), the normalization of CRP data is not consistent among researchers (Chiu & Chou, 2012).

Vector Coding

A second method of quantifying coordination that is increasing in popularity is vector coding (VC). VC is based on chain-encoding techniques and involves the transformation of the data curve from an angle-angle plot into a chain of digital elements(Glazier et al., 2003). The angle of one joint is represented on one axis, and the angle of another joint is on the other axis. For example from Tepavac and Field-Fote (2001), the hip is represented on the x-axis and the knee is represented on the y-axis. The angle for both joints is plotted for every frame. One advantage of VC is that it does not need to be normalized. However it only provides spatial information. In a study conducted by Pohl et al. (2007), researchers used VC to determine whether the kinematic coupling between the forefoot, rear-foot, and shank differed between walking and running speeds. There were twelve subjects with inclusion criteria consisting of rear-foot strikers, free from injury, and participated in some form of activity at least two hours per week. Four different speeds were determined based from the speed at which participants could not refrain from running defined as the max walking speed. A slow walking speed was defined as 50% of walking max, slow running was defined as the walking max, medium running at 120% and fast running at 140%. Five trials were used for analysis of PF/dorsiflexion (DF), EVE/INV and IR/external rotation (ER). Coupling was more specifically determined in forefoot, rear foot, and shank comparisons of the six variables. Correlation coefficients were determined first to indicate coupling. VC was then used to further assess joint rotation. Data showed that running trials had similar coupling characteristics. Running trials showed greater coupling between the forefoot, rear-foot, and shank compared with walking trials. Coupling of EVE/INV and shank IR/ER were lower in walking trials and high in the running trials. Walking showed a higher correlation between rear-foot EVE/INV and forefoot EVE/INV. The more coordinated patterns in the running trials agree with the dynamical systems theory in which an increase in frequency causes an increase in coupling of segments (Turvey, 1990). Overall, data showed there was greater kinematic coupling in running compare to walking in rear-foot and shank coupling. Data also showed that there was low coupling between rear-foot EVE/INV and shank internal and external rotation. This data suggests that forefoot, rear-foot, and shank are coupled during running and linked through the ankle-complex and mid-foot.

Alterations of foot strike patterns were also quantified using VC in s study conducted by Pohl and Buckley (2008). Twelve subjects participated in this study. Three different barefoot heel strikes were defined: a heel strike condition, forefoot strike, and a toe running condition. A total of five trials were collected for each condition. Researchers then used cross-correlations to determine coupling between the pairs of joint angles. However, cross-correlation only provides temporal information, and is limited to evaluating only linear relationships. VC was then used after cross-correlation to determine if coupled joint rotations were truly similar angular excursions or if one had a greater angular excursion. Overall kinematic coupling for all conditions were good. These findings suggest that segments maintain good coupling mechanics when the mechanics of gait are changed. Researchers also suggest that segments may be coupled through the ankle-complex, and mid-foot joints.

Coupling patterns of runners with normal rearfoot mechanics and those with excessive pronation were compared and used to predict injury by Mcclay (1997). There were eighteen runners total used in the study. Nine subjects were identified as excessive pronators and nine were identified as normal pronators. Using a treadmill, subjects ran at a speed of 3.5m/s and five foot-strikes were collected from each subject. Angle-angle plots were created for rearfoot EVE-knee IR and rearfoot EVE-FLEX. The relationship between rearfoot EVE and knee IR curves suggested a larger time difference between peak rearfoot EVE and knee IR in normal subjects. Data also suggested that the over-pronated group had greater knee FLEX and greater rearfoot EVE, which translates to greater tibial IR. Researchers concluded that those with excessive pronation were more prone to injury. Although there were some differences between timing of the knee and rearfoot angles, they were not significant and more research needs to be conducted.

Most of the previous studies have looked at strictly lower body mechanics. In a study by Seay et al. (2011), VC was used to assess pelvis and trunk range of motion and coordination difference in walking and running between those with low-back pain, recovered from LBP, and those with no history of LBP. Each group had fourteen participants that all completed the same protocol. Subjects were asked to walk at 0.8 m/s and increased the speed every 30 seconds by 0.5 m/s until 3.8 m/s. Data reduction included finding maximum and minimum angular excursions to determine ROM and angle-angle plots to determine pelvis and trunk coordination plots. The LBP group had the greatest ROM and the no history of LBP group had the least amount of ROM. Results of coordination suggested that the LBP group was less in-sync joint couplings in the running trial compared with the other groups.

Continuous Relative Phase versus Vector Coding

As evident by the studies reviewed above, both CRP and VC are widely used to understand coupling relationships between segments of the body. The purpose of Dierks and Davis (2007), was to explore several methods used to assess joint coupling relationships of the foot, shank, and thigh. These methods include both CRP and VC. This study included 40 recreational runners that were both male and female. All subjects ran an average of 10-20 miles per week and were free of injury. Subjects were instructed to run along a 25m speedway with an embedded force plate at a given speed of 3.65m/s. A total of five trials with kinematic and ground reaction forces were collected and used for data reduction and analysis. Both VC and CRP were used to evaluate the relationships among rear-foot EVE, tibial IR, knee FLEX, and knee IR. The methods were averaged for each subject's five trials. Coupling angles in VC and CRP phase plots were divided into four periods: Heel strike, to first max impact, half the distance to toe-off, and toe-off. Phase plots of CRP were normalized by placing the origin of the angular position in the middle of the range while normalizing the minimum value to -1 and maximum to 1. VC relationships of rearfoot EVE/INV- tibial EVE/INV, rear-foot EVE/INV- knee IR, and tibial IR- knee IR all had values greater than 45° for all four periods. This indicates that distal segments exhibit greater excursions throughout stance. Also, at mid-stance, 45° of coupling indicates there is little change in joint motions. CRP values showed that phase three was the most in phase and least variable, while phases 1 and 4 were the most anti-phase and had the most variability. These are the transition phases of swing-to-stance and stance-to-swing. Both methods indicate that the more anti-phase couplings also had the greatest variability. Researchers concluded that more studies need to be done in order for the relationships to be better understood.

Movement coordination and coordination variability have been used to identify differences between healthy and injured populations. To accomplish this goal researchers have used both CRP and VC techniques. Heiderscheit (2002) investigated the variability of stride characteristics and joint coordination between subjects with patellofemoral pain and those without. They used VC to quantify joint coordination for eight female subjects diagnosed with unilateral patellofemoral pain were matched with eight female subjects that were asymptomatic. All subjects were asked to walk at a fixed speed of 2.68 m/s or run on a treadmill at an average selected speed between 0.5-8.0 m/s. Kinematic data was collected for a total of twenty seconds. Angle data was linearly interpolated with 0% to 100% at heel-strike and terminal swing, respectively. Within limb couplings were created for thigh IR/leg IR and thigh FLEX/leg FLEX. The average standard deviation for the entire trial was used for each joint coordination components. Researchers focused on midstance, toe-off, swing acceleration, swing deceleration, and heel strike. Results showed that average variability between both the patellofemoral pain and healthy group was similar for the entire stride. Broken down into regions, the patellofemoral pain group showed less variability at heel strike in the preferred running speed.

While CRP and VC have been used almost interchangeably to assess coordination variability, the influence of methodology choice remains relatively unknown. Miller et al. (2010) conducted a study to compare the variability differences using both CRP and VC. This study used a theoretical data set, and two sets of experimental data. The experimental data sets were composed of one walking data collection and the other with varying speeds. Variability was measured during the stance phase of walking. Subjects performed five trials of walking bare-foot at a self-selected pace across a force plate. FLEX/EXT, INV/EVE, and ADD/ABD of the rearfoot and forefoot were used for analysis. The second experimental trial consisted of five healthy males walking on a treadmill at three speeds (0.8, 1.4, and 2.0) and running at five speeds (2.2, 2.7, 3.4, 3.9, and 4.5). The thigh and leg FLEX/EXT angles were used to compare VC and CRP. Variability in VC was greatest during mid stance and variability in CRP was greatest in early and late stance for INV/EVE. VC also showed a greater magnitude of variability than CRP. VC did not show consistent variability for INV/EVE while CRP showed greatest variability in early and late stance. ADD/ABD had similar variability in VC compared with CRP. However, peak variability occurred at different times. The walking trial comparisons of FLEX/EXT and INV/EVE both showed discrepancies of peak variability and timing of stance phase between VC and CRP. The overall magnitude in

VC was greater than CRP. The ABD/ADD comparison showed no significant differences in variability between the two methods. Both experiments in this study show that VC and CRP do not consistently give the same results with the same parameters. They are both valid forms in quantifying data, however, researchers must be aware when comparing the two methods that they are giving different conclusions. They suggest that pathological movement should be looked at as well compared with the two methods since both methods have previously been used.

Ankle Foot Orthotics and Stabilization

The previous studies have looked at several methods of assessing coordination and variability. Researchers have evaluated coordination and variability in altering mechanics (Pohl & Buckley, 2008), speed (Chiu & Chou, 2012; Mcclay, 1997; Pohl et al., 2007), and injured subjects (Hamill et al., 1999; Seay et al., 2011). However, these research studies have not looked at how the body self-organizes when a DOF is purposefully frozen. Ankle-foot-orthotics (AFOs) have been used as an effective means of stabilization. The use of orthotics devices such as AFOs are commonly used in subjects with spastic diplegia, post-stroke hemiplegia, and other gait abnormalities. Researchers have conducted several studies to determine the effect of an AFO on gait patterns in these subjects. This presents an interesting question in how the body self organizes when a DOF (talocrural joint) is unable to move through its ROM.

AFOs have been used to enhance gait in subjects with spastic diplegia which effects lower extremity gait patterns. Buckon et al. (2004) conducted a study of 16 subjects with spastic diplegia included in the study. All subjects completed a baseline test after three months wearing no AFO. Subjects completed a gait analysis using a three-dimensional motion analysis, functional skills, gross motor skills, fine motor skills, functional mobility, and energy expenditure. After the initial test subjects wore each AFO for three months with an assessment to follow up. They found that the only change of proximal segments was in the hinged-AFO that created peak knee extensor moment in early stance. Subjects with spastic diplegia benefit from using an AFO to encourage increased dorsiflexion throughout stance to encourage heel strike. Although there were no significant changes in gait kinematics, there were significant improvements in gross motor function.

Gatti et al. (2012) conducted a study with chronic post-stoke hemiplegia patients. They looked at knee flexion at toe off and peak knee flexion in AFO and non-AFO conditions. This study included ten subjects with chronic post-stroke hemiplegia. Subjects were asked to walk 10m in each condition for six trials. Results showed that gait speed was higher and step length was longer when subjects wore the AFO. They also found an increase in peak knee flexion angle in participants with that wore the AFO. Opposed with the previous study conducted by Buckon et al. (2004), Gatti et al. (2012) found that there were changes in proximal kinematics with use of the AFO.

Nair et al. (2010) conducted a study in which post-incomplete spinal cord injury patients were assessed with and without an AFO. The purpose of this study was to assess if motion during the transition phase in gait was similar to healthy gait function when using an AFO. Subjects were asked to walk on a treadmill with a harness at a speed of 1.2m/s. Once the subject was comfortable, data was collected for 30s. Results show that the AFO caused a decrease in ankle plantar flexion, which reduced the ability of the ankle to contribute to push off. Which is not comparative to healthy gait function. There was no difference in hip extension, however there was an increase in hip flexion. There was also an increase in ankle dorsiflexion. Results showed that subjects were able to maintain walking speed and used other mechanisms to account for the lack in symmetry and loading rate that were changed in the AFO condition.

Lage, White, and Yack (1995) did not use an AFO, however they did look at immobilization of the knee joint. This is commonly done following acute knee trauma to correct malalignment and facilitate healing. The purpose of this study was to identify the effects of knee immobilization on uninvolved lower extremity joints during walking. This study included seven healthy college aged students that were recruited for three days of testing. None of the subjects had any previous injury or had never had to immobilize the knee. Three trials were collected including of normal gait and five trials of braced gait with 0°, 10°, and 20° of knee flexion. Subjects were instructed to walk at 90% of their gait speed for all trials. Results show that the ankle and hip of the involved limb were affected the most. The ROM at the hip was greater than normal for all braces. There were also changes to the contralateral limb, which suggests that this limb compensates for changes that have occurred in the involved limb.

Conclusion

AFOs have been used to promote a more normal gait pattern in those with disability. While the influence of AFOs on joint kinematics during gait is well documented, the influence of such devices on movement coordination remains unknown. Given that AFOs restrict motion at the ankle joint, effectively eliminating a DOF, it is reasonable to assume that movement coordination will change as the individual searches for a new behavior pattern through self-organization. Understanding the relationship between a constrained system and movement coordination may help uncover the complexity of human movement and may provide a better understanding of how the motor system adapts. This information may provide insight into the flexibility of the motor system to cope with perturbations and dissipate stress at specific points of transition throughout the gait cycle.

Chapter 3: Methods

Participants

Sixteen participants, six males and ten females, between the ages of 18 and 35 were recruited for this study. All participants were screened using a validated questionnaire to determine inclusion into the study. To qualify for the study, participants had to participate in at least thirty minutes of activity three days per week. They were excluded if they have had a lower extremity injury or surgery within the last six months or had any neurological abnormality that would impair gait. Prior to data collection, research approval from the Institutional Review Board of Old Dominion University and written informed consent from all participants was obtained.

Instrumentation

Marker coordinate data was collected at 200 Hz using an eight-camera motion analysis system (Vicon, Centennial, CO, USA). Synchronously, three-dimensional ground reaction force data was collected using a Bertec force plate (Columbus, OH, USA) flush mounted with the floor.

Experimental Protocol

Participants were asked to dress in tight athletic shorts, females wore tank tops and males did not wear a top. All participants wore laboratory shoes (Nike Air Max Glide) to prevent potential differences created by the shoe. Subjects then performed five walking trials to determine the average preferred gait speed. Gait speed was monitored using timing gaits (Brower Timing Systems) set one-meter apart.

To track lower extremity kinematics, 48 light reflecting skin markers with a diameter of 12.7 mm were placed bilaterally on the participant (Weinhandl, Joshi, &

O'Connor, 2010). Anatomical calibration markers were placed on the left and right acromioclavicular joints, iliac crests, greater trochanters, medial and lateral epicondyles of the knees, medial and lateral malleoli, and the first and fifth metatarsal heads. Tracking markers were placed on the anterior-superior iliac spines, posterior-superior iliac spines, rigid plates with four markers attached to the thoracic spine, bilateral thighs, shanks and heel of the shoes.

After the markers were properly placed, a three second static calibration trial was collected. After collection of the static calibration, the calibration markers were removed and participants completed five successful walking trials at their preferred walking speed while wearing an AFO on their right foot. A second set of five trials was collected without wearing the AFO. Prior to collection of trials while wearing the AFO, participants were allowed a 10-minute accommodation period. The order of non-AFO and AFO condition were counterbalanced between participants. A successful trial was defined as foot contact in the middle of the force plate and a consistent speed with a 5% deviation from the preferred walking speed.

Data Reduction

Data reduction was implemented with Visual3D (v4.86, C-Motion Inc., Rockville, MD). Raw marker coordinate data for the right leg was filtered using fourth-order, low pass Butterworth filter with a cut off frequency of 12 Hz. Right-handed Cartesian segment coordinates systems defined position and orientation of each segment (Spoor & Veldpaus, 1980). Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system approach (Grood & Suntay, 1983). Joint centers were defined for the hip as 25% of the distance from the ipsilateral condyle to the greater trochanter markers (Weinhandl & O'Connor, 2010). The knee was defined as the midpoint between the medial and lateral epicondyle markers (Grood & Suntay, 1983) and the ankle as the midpoint between the medial and lateral epicondyles (Wu, 2002).

Data Analysis

A reference angle for each of the kinematic variables (hip and knee) were determined from the static trials and were subtracted from the angles recorded for the AFO and non-AFO conditions. For each trial, gait events were determined using the forceplate data. These strides were time normalized 101 frames to represent each percent of stride. Using vector-coding assessment techniques (Tepavac & Field-Fote, 2001), joint coordination variability of the hip and knee couples were determined. An angle-angle plot was constructed from the rotations of interest (i.e., hip flexion and knee flexion). A vector angle and magnitude for each consecutive point on the angle-angle plot was determined. A vector coefficient ranging from 0 (no variability) to 1 (maximum variability) was then determined from the variability in these vector angles and magnitudes across multiple strides (Mullineaux et al., 2001). The average of the vector coding coefficients for all trails was calculated to measure the coupling variability relationship for the hip and knee joints.

Statistical Analysis

To determine differences between the two conditions, a curve analysis using the mean ± standard error (SE) for each of the 101 points was determined. Differences between conditions were determined as at least five consecutive points were SEs did not overlap. Where non-overlapping points were observed, the mean was calculated across the points for each participant. Differences between conditions found on the curve analysis and from the average vector coding coefficient were analyzed using paired t-tests. Significance was set at $p<0.05$. Effect size was calculated based on the mean and pool standard deviation of group differences with corresponding 95% confidence intervals.

Chapter 4: Results

Group means and standard deviations for all joint couplings of interest are located in Table 1. Mean ensemble curves for one complete stride are presented stance leg joint angles (Figures 1-9) and joint couplings of interest (Figures 10-18).

Qualitative inspection of the hip joint angle time series reveals that AFO gait averages resulted in decreased hip flexion and increase hip adduction throughout the stride. In the transverse plane, AFO gait resulted in decreased hip internal rotation during the stance phase and then increased internal rotation during the swing phase. Confidence intervals indicate that there were no significant differences in hip kinematics between AFO and normal conditions.

The knee in extension and adduction did not significantly changes in the AFO condition compared with the normal. The knee showed no overlap in the stance phase of external rotation in the AFO condition compared with normal gait.

Joint angles of the ankle in the sagittal plane showed plantar flexion was restricted at heel-strike and push-off phase in the gait cycle. The AFO restricted ankle inversion/eversion during part of the stance phase where no overlap in confidence interval bands was shown. Internal/external rotation was not significantly different throughout the gait cycle compared with normal gait.

No differences were found in joint coordination variability between normal gait and AFO gait for all joint couplings of interest. For the hip flexion – knee flexion coupling the vector coding coefficient for normal gait was 0.22 (0.10) compared to 0.19 (0.03) for AFO gait ($p=0.301$). For the hip flexion – knee ab/ad coupling the vector coefficient for normal gait was 0.28 (0.10) compared to AFO gait vector coefficient 0.26

(0.04), significance of (p=0.449). The hip flexion-knee rotation vector coefficients for normal compared to AFO gait was 0.31(0.08) and 0.30 (0.04) with a significance of (p=0.493). Hip abd/add-knee flexion vector coding coefficients for normal and AFO gait were .32 (0.11) and 0.29 (0.05) with a significance of (p= 0.280). The vector coefficients for hip abd/add-knee abd/add were 0.39 (0.09) for normal gait and 0.38 (0.04) with a significance of (p=0.685). Hip abd/add- knee rotation had vector coefficient values of 0.47 (0.05) and 0.45 (0.05) for normal and AFO gait respectively with a significance of (p=0.474). Hip rotation- knee flexion had normal and AFO vector coefficient values of 0.32 (0.11) and 0.29 (0.04) with a significance of (p=0.343). Hip-rotation- knee abd/add had vector coefficient values for normal and AFO gait of 0.46 (0.09) and 0.44 (0.04) with a significance of (p=0.427). The hip rotation-knee rotation vector coefficients for normal and AFO gait were 0.48(0.08) and 0.47(0.05) with significance of (p=0.565).

Chapter 5: Discussion

It was hypothesized that coordination variability would increase at the hip and knee when the ankle joint was stabilized. Results of the current study revealed that there were no significant differences in coordination variability for any of the nine joint couplings between the two conditions. Coordination variability differences were determined by examining the vector coding coefficient time series for periods of non-overlap between SE intervals between joint coupling of the two conditions. The average of vector coding coefficients for each joint coupling across the entire stride was also compared between conditions to provide a summary measure. These results may indicate that the body is extremely capable of adapting to changes created by restricting sagittal plane ankle motion and thus freezing a DOF of the lower extremity.

Although this study did not see changes in coordination variability, Bernstein (1967) suggests there are different stages in organization of movement patterns. The first stage in response to a change in normal movement patterns is to freeze DOF. It is possible that the body could have adapted to the constraint in the first ten minutes of adjusting to the AFO, which would result in no differences in coordination variability. A follow up study may include a collection of continuous gait cycles for several minutes to see if there is an adjustment period after the AFO is put on. This may help understand how movement coordination is organized in the presence of constraints.

In this study, the AFO may not have created enough of an internal change for the body to change the utilization of motor control patterns. In healthy individuals with no previous injury, coordination variability is greater which suggests that they are able to utilize multiple strategies complete a given task. The previous studies of Chiu and Chou

(2012); Hamill et al. (1999); Heiderscheit (2002); Seay et al. (2011) all used coordination variability to examine differences in coordination in gait. These studies compared healthy individuals with injured subjects. The internal constraints of the injury all resulted in lesser coordination variability in all cases. This may be due to the injured subjects having internal impairment compared to external constrains of the body. In the current study, all participants were healthy, with no history of lower extremity injury. As healthy populations have been shown to have greater coordination variability compared to injured populations, it is possible that the participants utilized in the current study were able to adapt to a sagittal plane ankle range of motion constraint with no change in coordination variability between hip and knee joint couplings.

Nair et al. (2010) used healthy subjects to look at the kinematics of normal and AFO gait. They found an increase in hip flexion and a decrease in dorsiflexion in AFO gait compared with normal gait. Visual inspection of the joint angle time series in the current study indicates that while the AFO resulted in decreased dorsiflexion it also resulted in decreased hip flexion, which is in contrast to the findings of Nair et al. (2010). However, even though there was a shift in the hip flexion joint angle, it was consistent throughout the stride and the overall pattern was similar between normal and AFO gait. The consistent pattern of hip and knee kinematics in the presence of restricted ankle sagittal plane range of motion could possibly be accounted for in the kinetics, which were not analyzed. Winter (1984) reported high variability in hip and knee joint moments profiles as a function of gait speed with low variability in the joint kinematics and ground reaction forces. He surmised that this was evidence that a wide range of joint moment patterns at the hip and knee could result in identical joint angle patterns during gait.

Also, our study found that plantar-flexion was restricted, but not dorsiflexion. This may also have contributed to the non-significant of results in this study. Joint angles show that there was movement in the sagittal plane of dorsiflexion. This means that the AFO did not completely immobilize movement, which may have accounted for movement and allowed the hip and knee joints to function normally.

It is unclear how the body self-organizes itself to produce human movement. The AFO did restrict plantar-flexion and mildly alter hip and knee motion, however coordination variability did not change when examining the coupling relationship of knee and hip. The task was successfully completed even with a constraint that restricted ankle motion. Although other studies have seen differences in the coordination variability of injured subjects, it is still unclear how coordination variability changes in the presence of other forms of constraint. Therefore, further studies in healthy individuals may provide a better understanding of this aspect of motor control.

Limitations

The current study used the preferred speed of each participant, which was different for all subjects. A future study may look at increasing and decreasing gait speed to elicit a greater change in coordination variability at the hip and knee joint (Pohl et al., 2007).

The timing of AFO application is another limitation of this study. Subjects may have adapted within the first ten minutes of wearing the AFO, as previously mentioned. However, long-term effects of wearing the AFO may also elicit changes in the gait cycle. A future study might consider investigating the effects of prolonged AFO usage on coordination variability.

Although there were no significant changes in joint coordination variability, there were apparent differences in joint motion, particularly at the ankle. Furthermore, it is unknown if joint moments were altered between the AFO and normal conditions. Future studies should compare magnitude and coordination variability of joint moments between condition.

The AFO device used only restricted planar-flexion movement and did not effectively restrict dorsiflexion. Future studies may see differences at the hip and the knee by using another device that would completely immobilize the ankle joint.

Conclusion

The AFO was successful in restricting sagittal plane ankle motion, most notably, ankle plantar-flexion. However, the restricted ankle motion did not have an effect on coordination variability of the hip and knee joint couplings. While the kinematic patterns appear to change, the body was able to complete the task without altering coordination variability. This result indicates that a healthy human movement system can adapt to eliminating a DOF with seamless efficiency.

Table 1. Vector coding coefficients of the hip and knee joints of normal and AFO gait with standard deviations and significance of each coupling

	Normal	AFO	P value
Hip-Flexion Knee-Flexion	0.22 (0.10)	0.19 (0.03)	0.301
Hip-Flexion Knee-Abd/Add	0.28 (0.09)	0.26 (0.04)	0.449
Hip-Flexion Knee Rotation	0.31 (0.08)	0.30 (0.04)	0.493
Hip-Abd/Add Knee-Flexion	0.32 (0.11)	0.29 (0.05)	0.280
Hip-Abd/Add Knee-Abd/Add	0.39 (0.09)	0.38 (0.04)	0.685
Hip-Abd/Add Knee-Rotation	0.47 (0.05)	0.45 (0.05)	0.474
Hip-Rotation Knee-Flexion	0.32 (0.11)	0.29 (0.04)	0.343
Hip-Rotation Knee-Abd/Add	0.46 (0.09)	0.44 (0.04)	0.427
Hip-Rotation Knee-Rotation	0.48 (0.08)	0.47 (0.05)	0.565

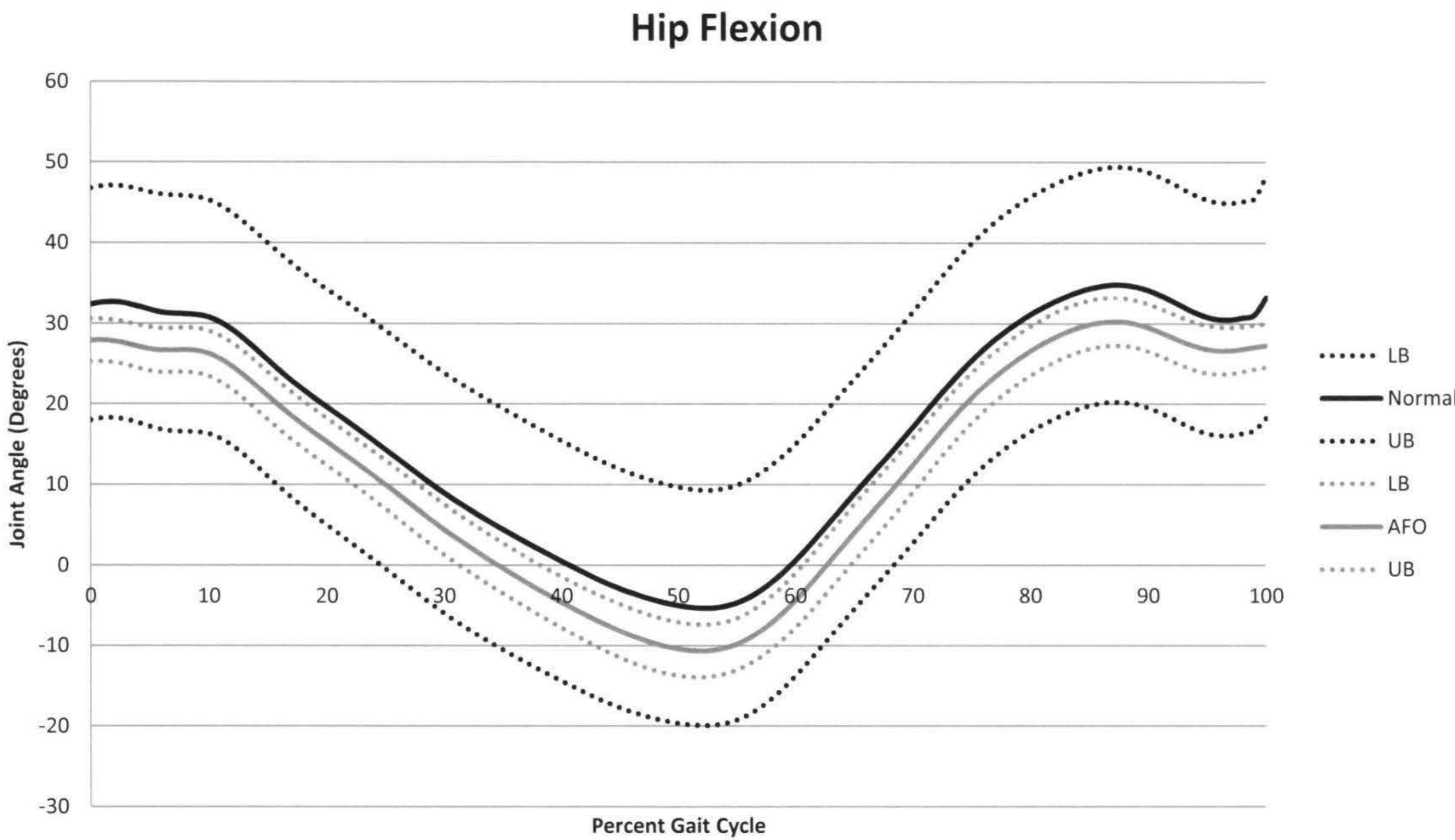

Figure 1. Group mean hip flexion curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent hip flexion, while negative values represent hip extension.

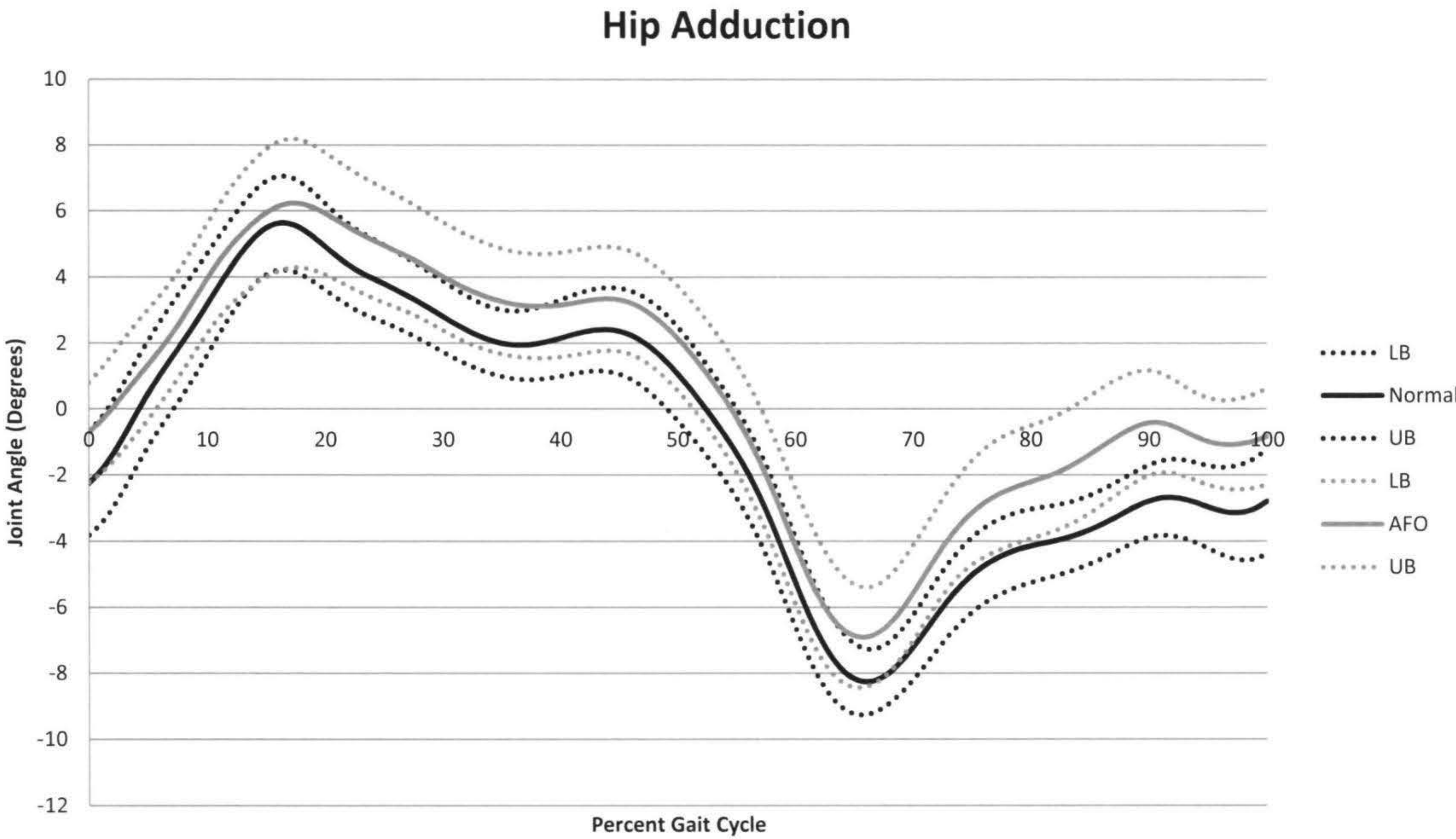

Figure 2. Group mean hip adduction curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent hip adduction, while negative values represent hip abduction.

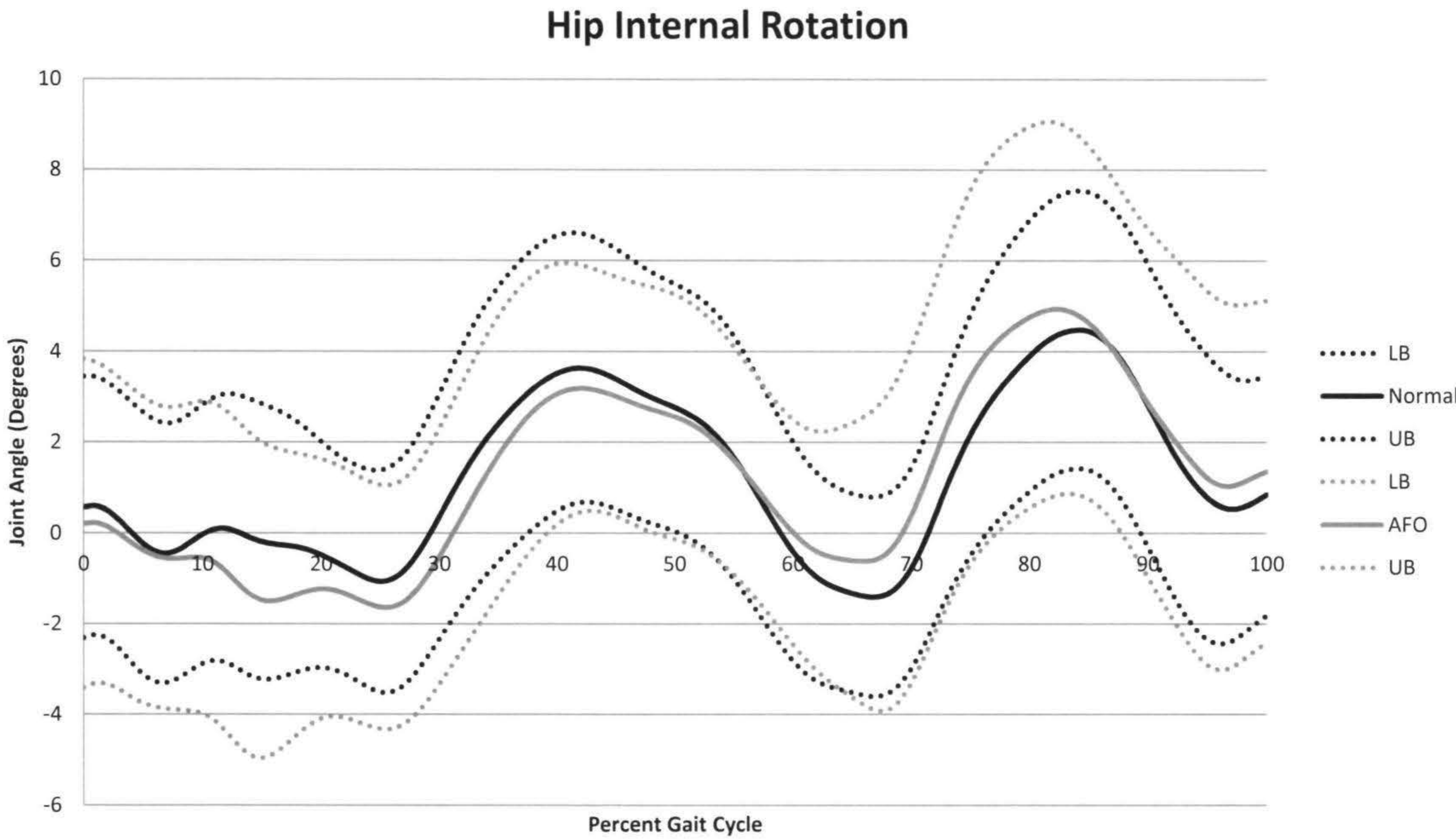

Figure 3. Group mean hip internal rotation curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent hip internal rotation, while negative values represent hip external rotation.

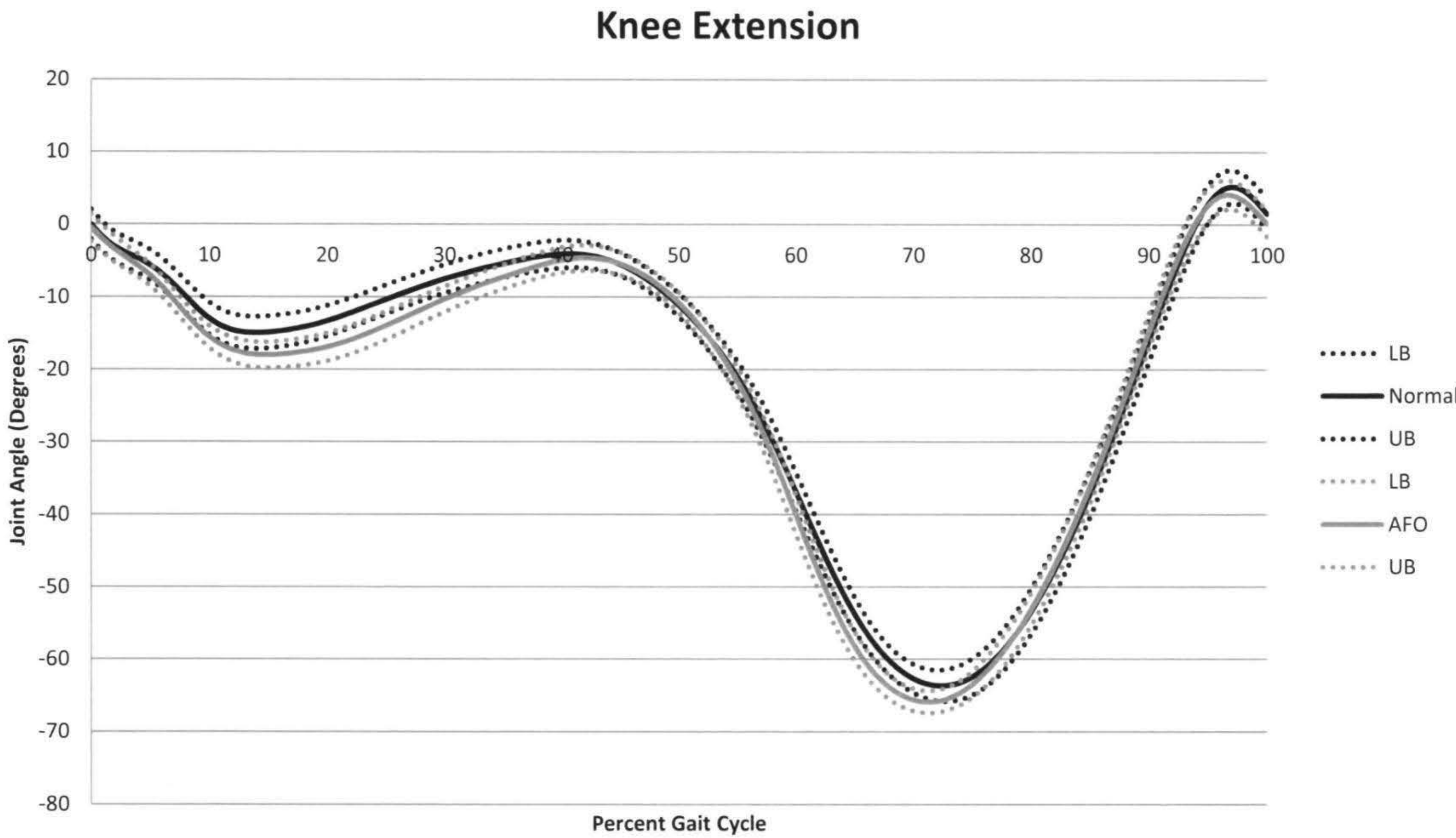

Figure 4. Group mean knee extension curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent knee flexion, while negative values represent knee extension.

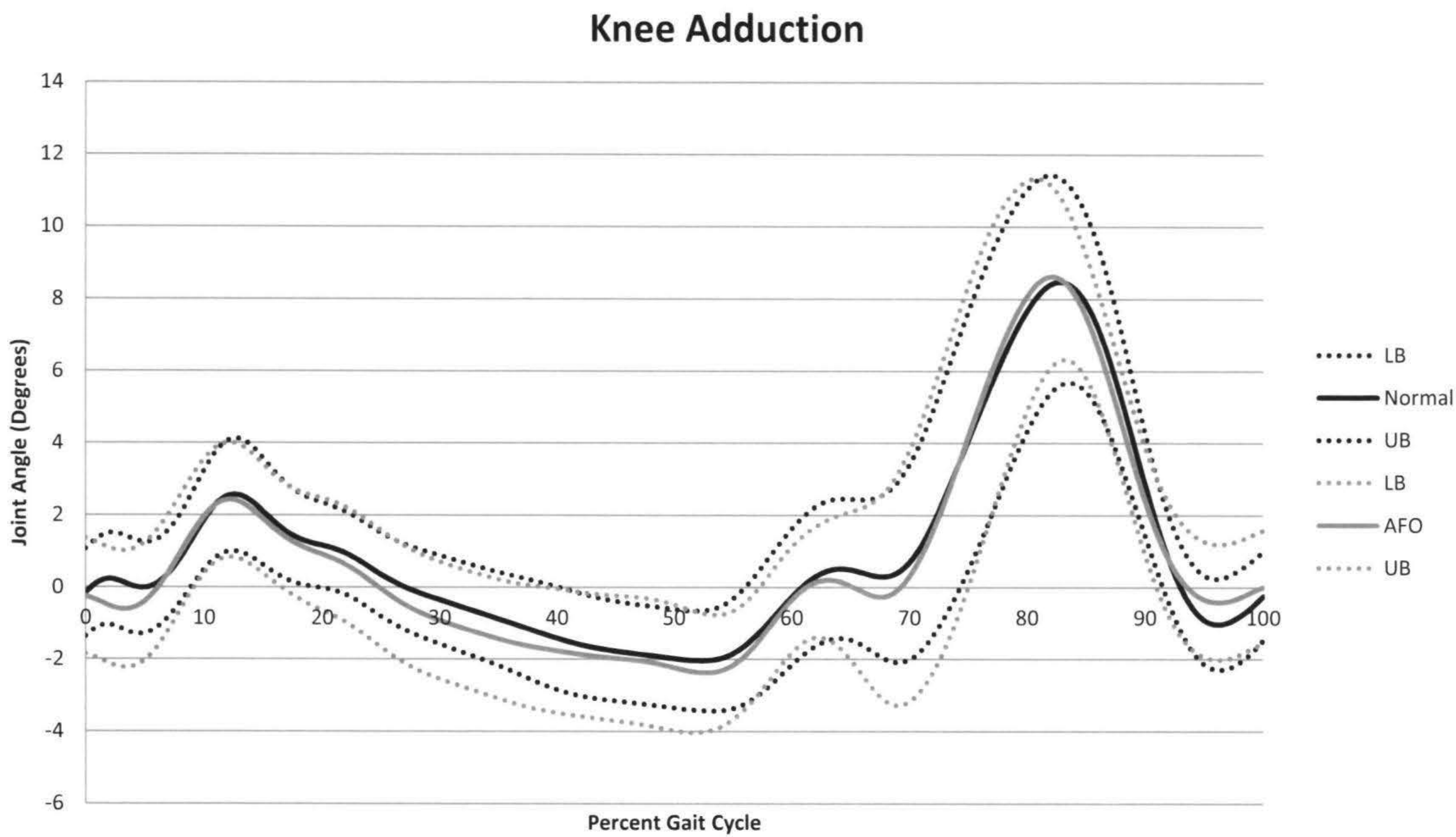

Figure 5. Group mean knee adduction curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent knee adduction, while negative values represent knee abduction.

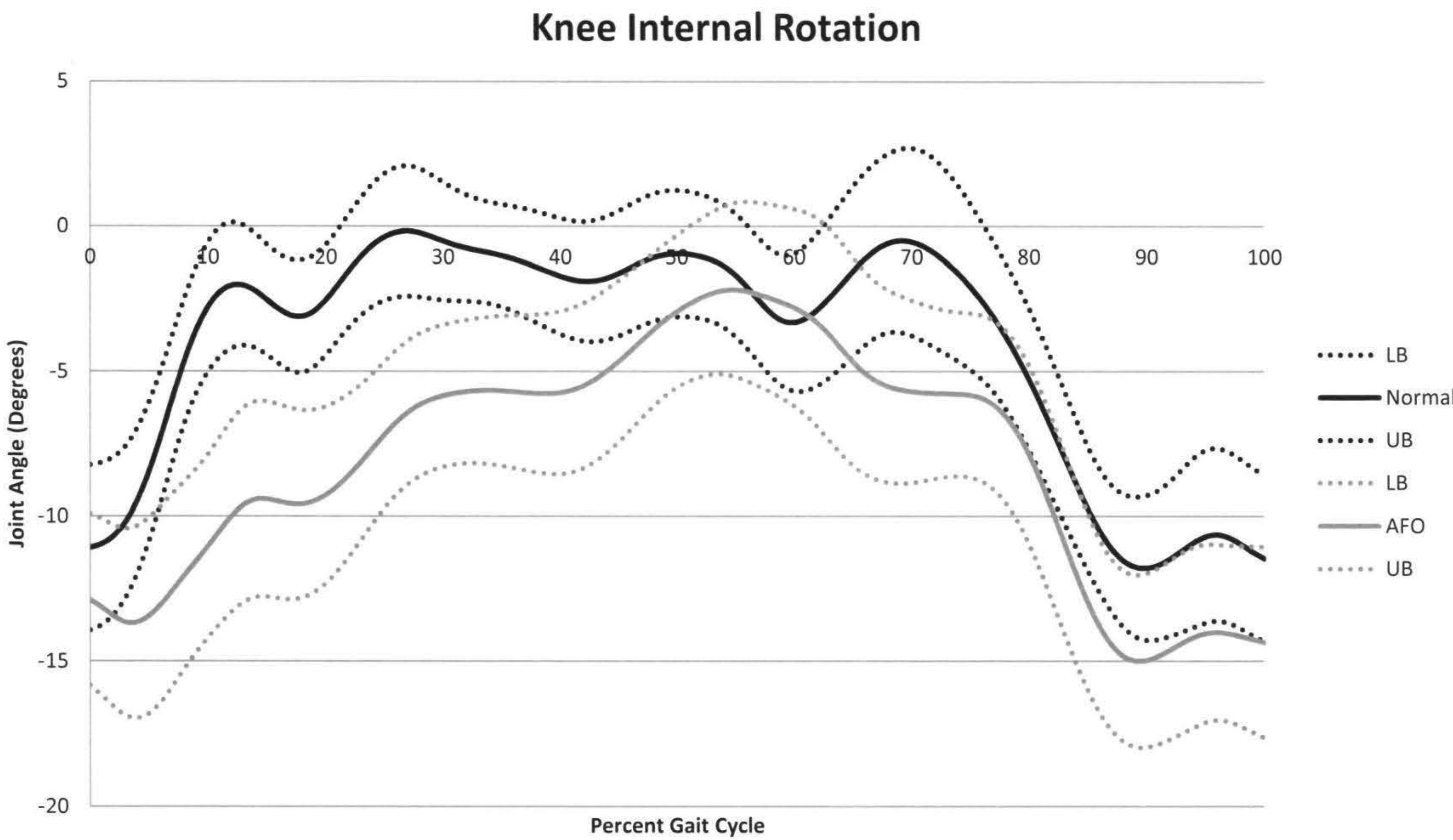

Figure 6. Group mean knee internal rotation curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent knee internal rotation, while negative values represent knee external rotation.

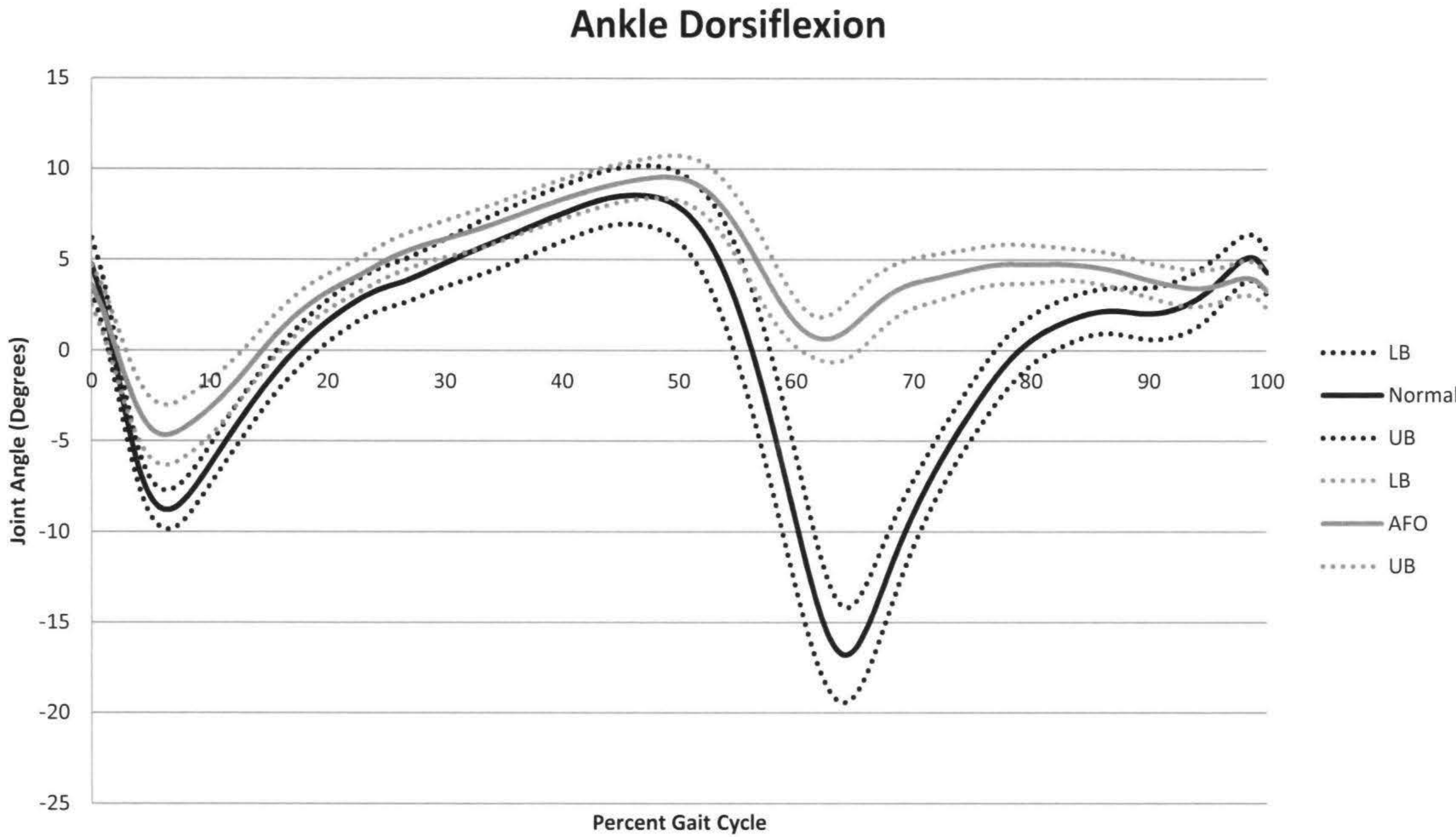

Figure 7. Group mean ankle dorsiflexion curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent ankle dorsiflexion, while negative values represent ankle plantarflexion.

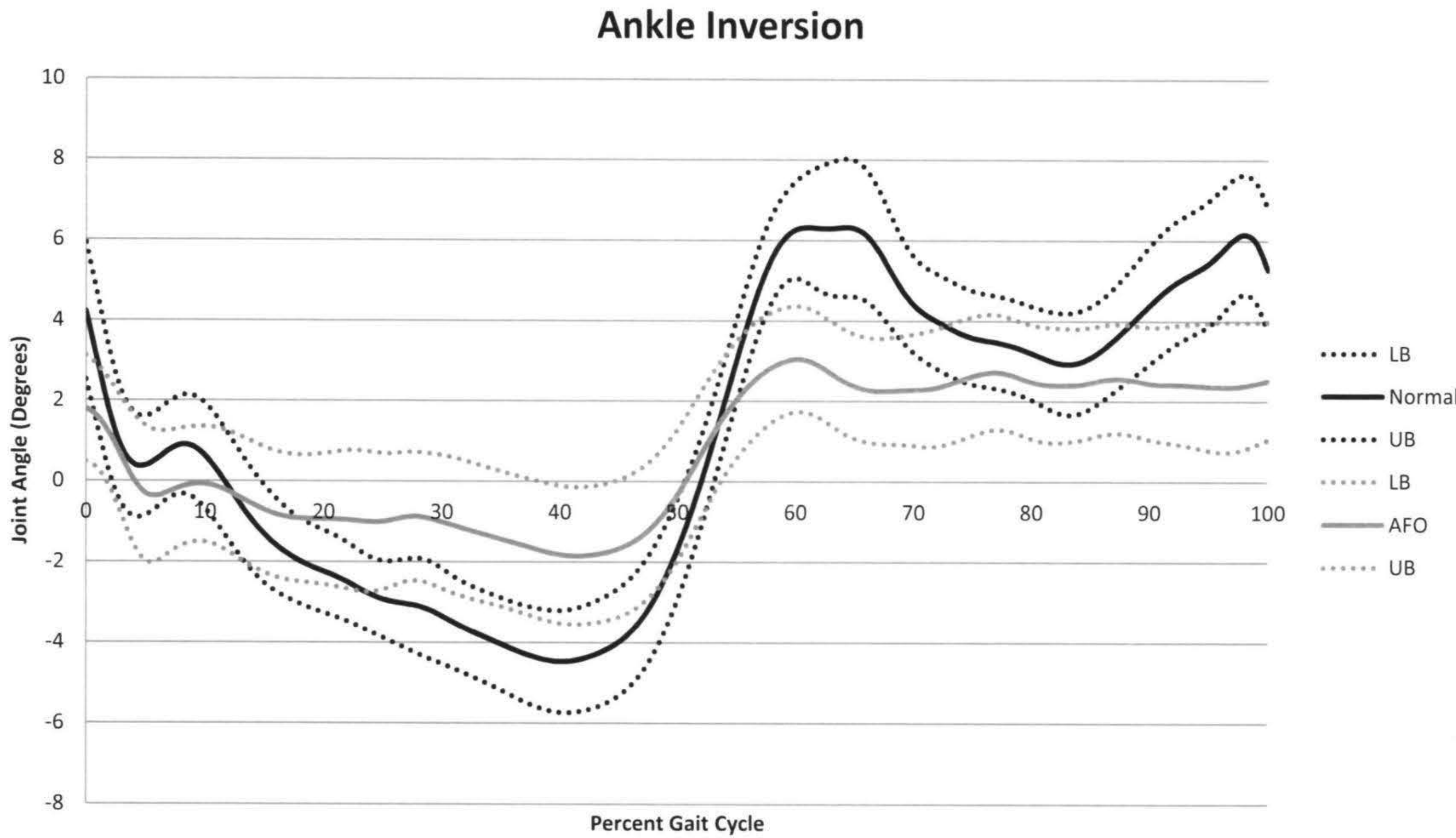

Figure 8. Group mean ankle inversion curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent ankle inversion, while negative values represent ankle eversion.

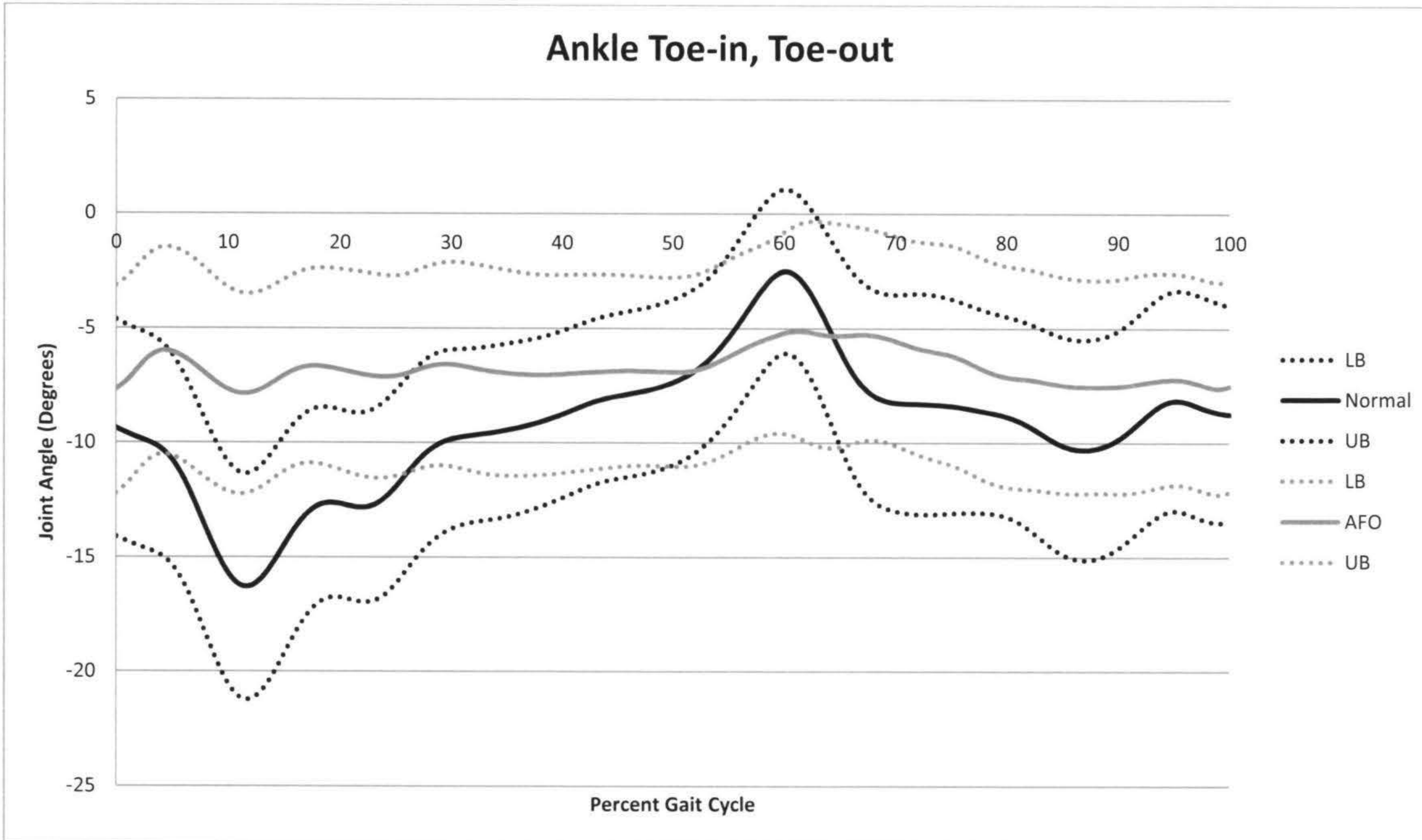

Figure 9. Group mean internal/external rotation of the ankle curves (solid lines) and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Positive angles represent ankle toe-in, while negative values represent ankle toe-out.

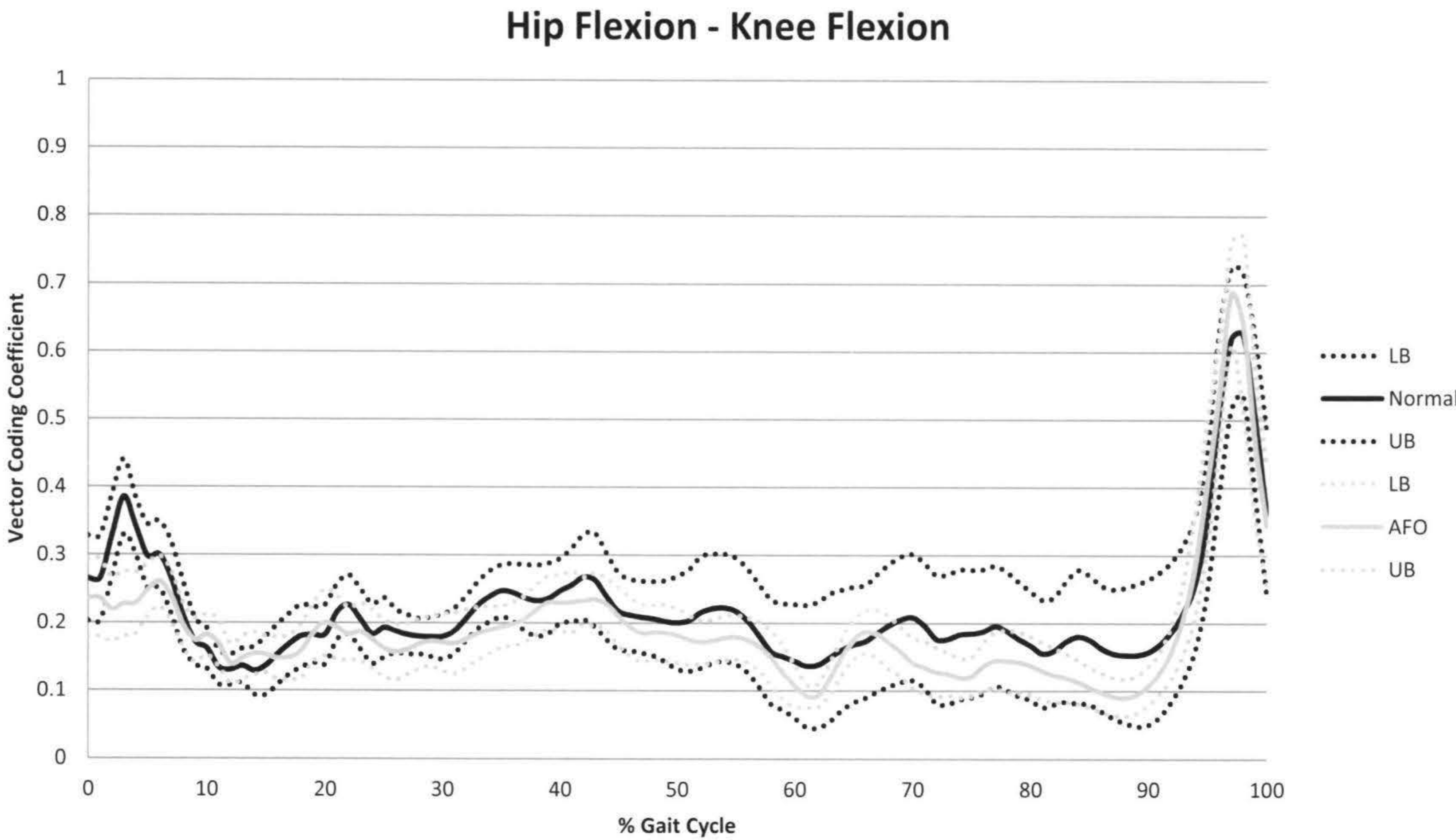

Figure 10. Hip flexion-knee flexion group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

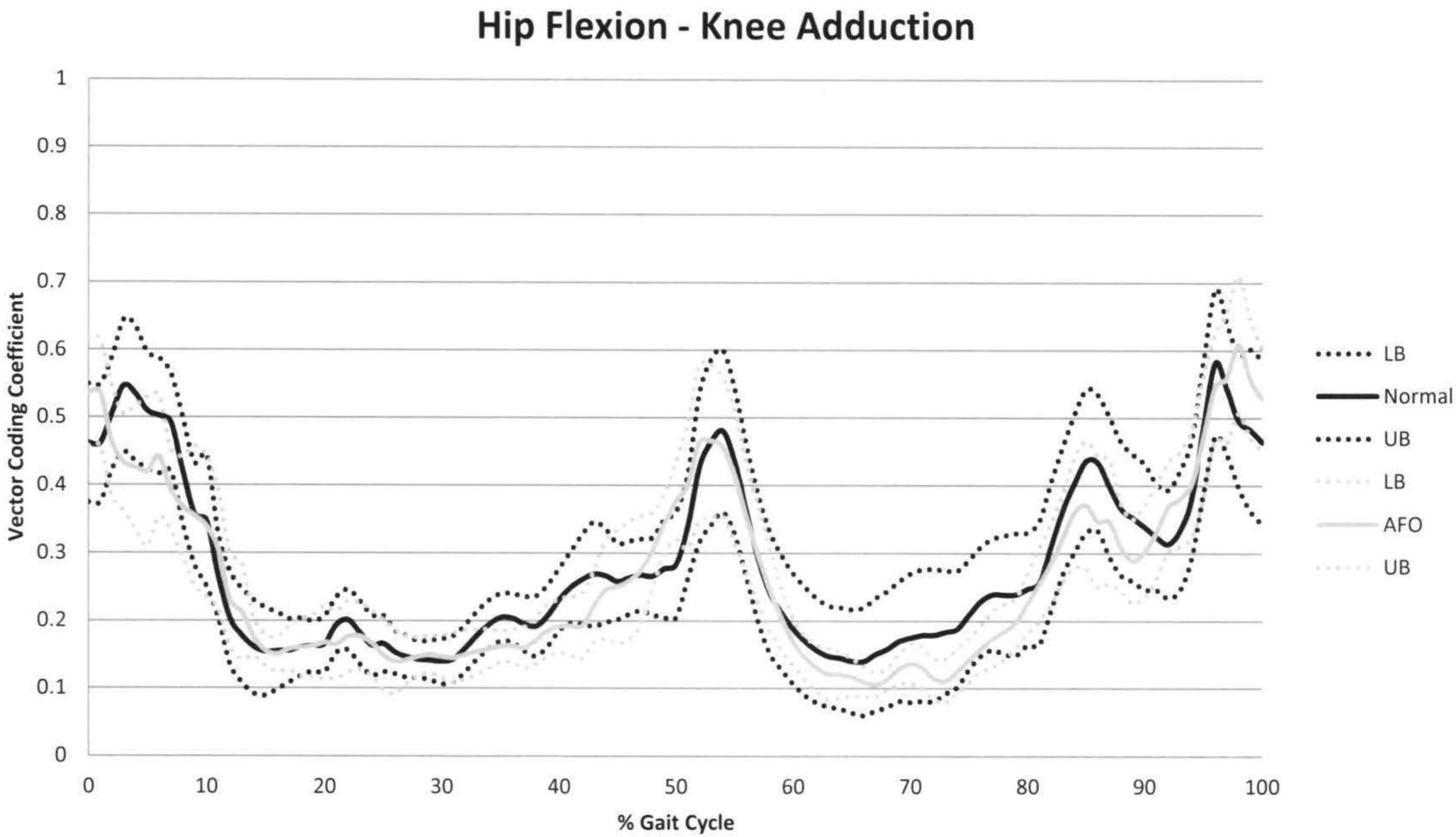

Figure 11. Hip flexion-knee adduction group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

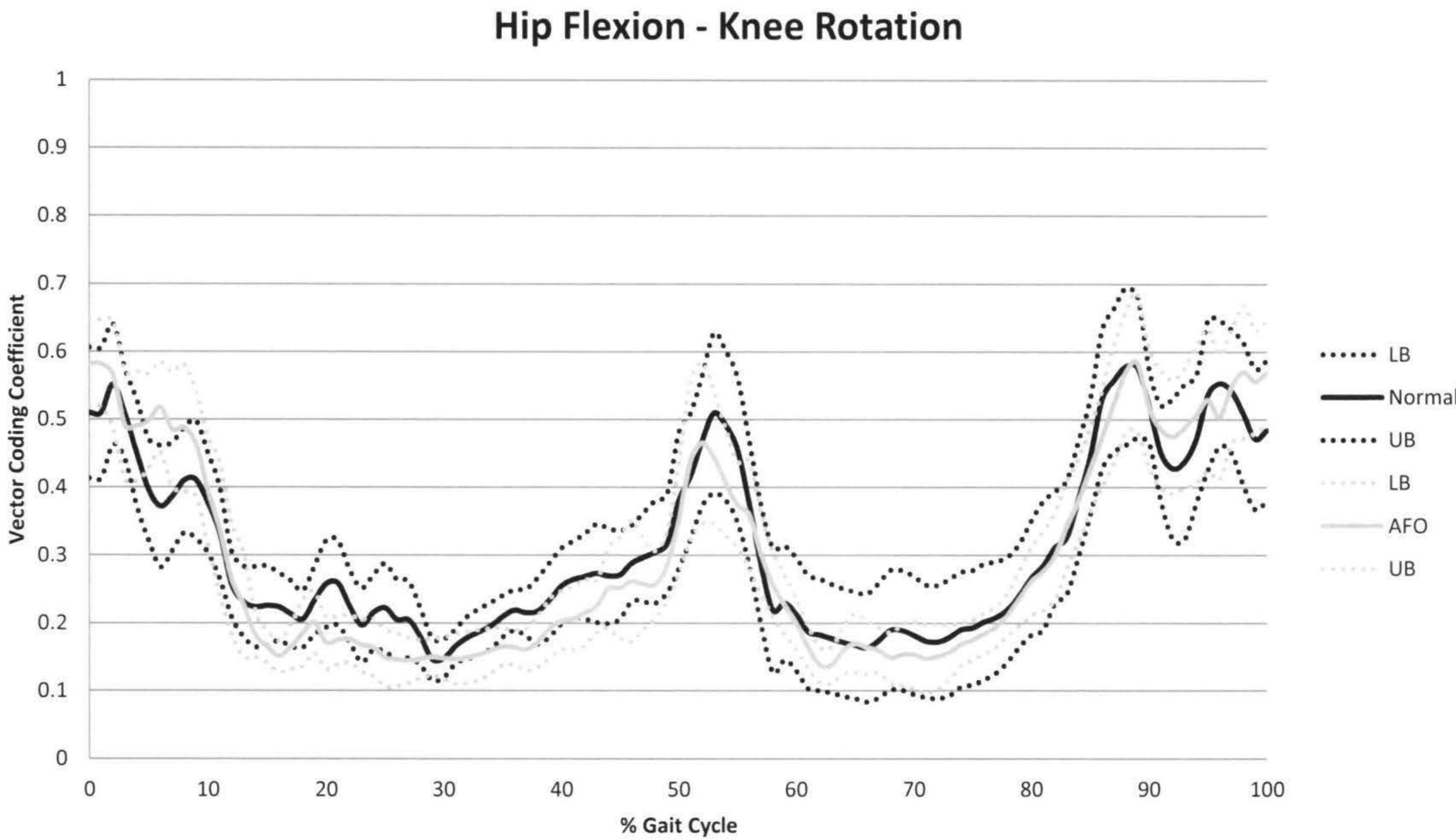

Figure 12. Hip flexion-knee rotation group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

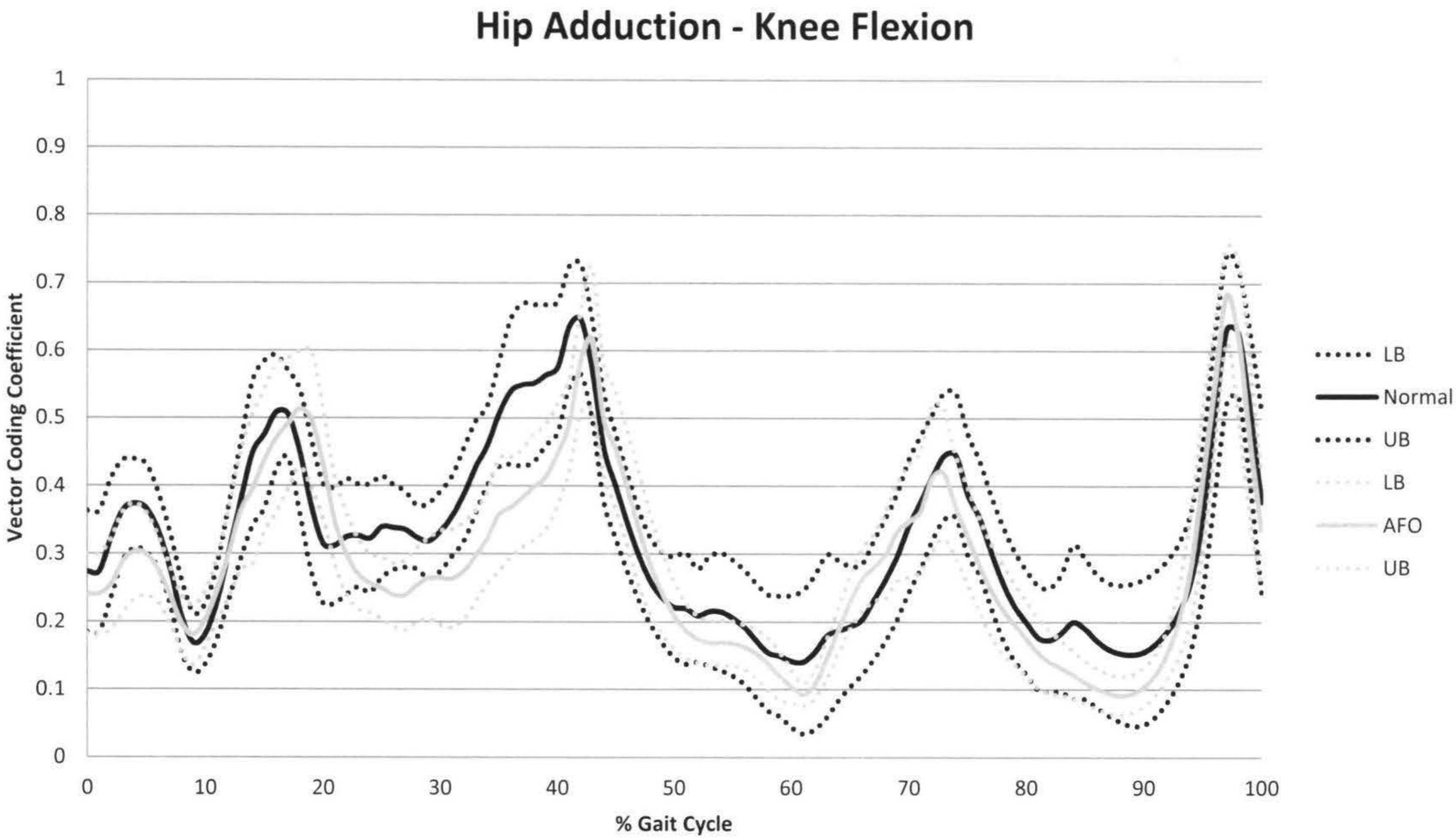

Figure 13. Hip adduction-knee flexion group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

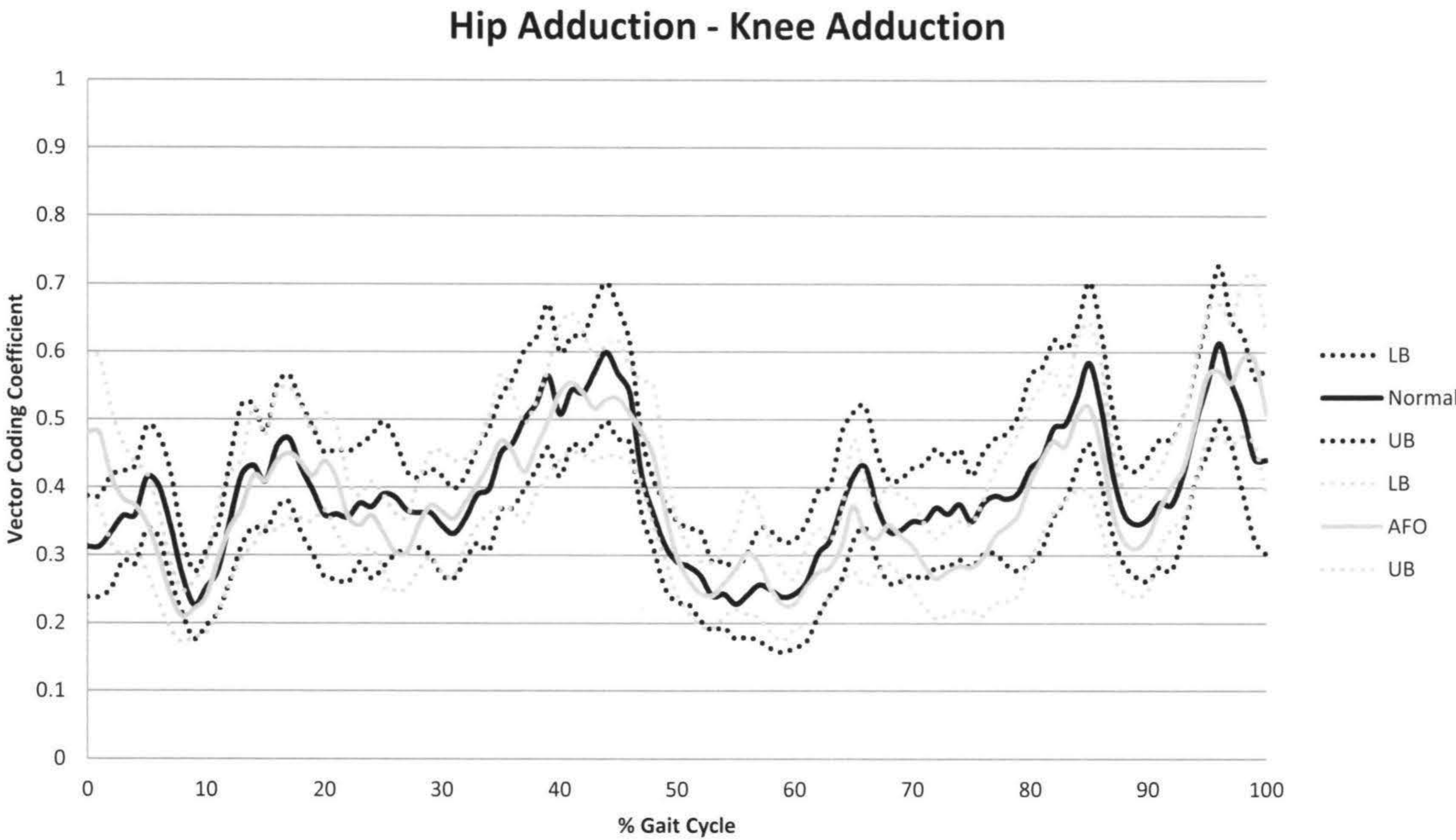

Figure 14. Hip adduction-knee adduction group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

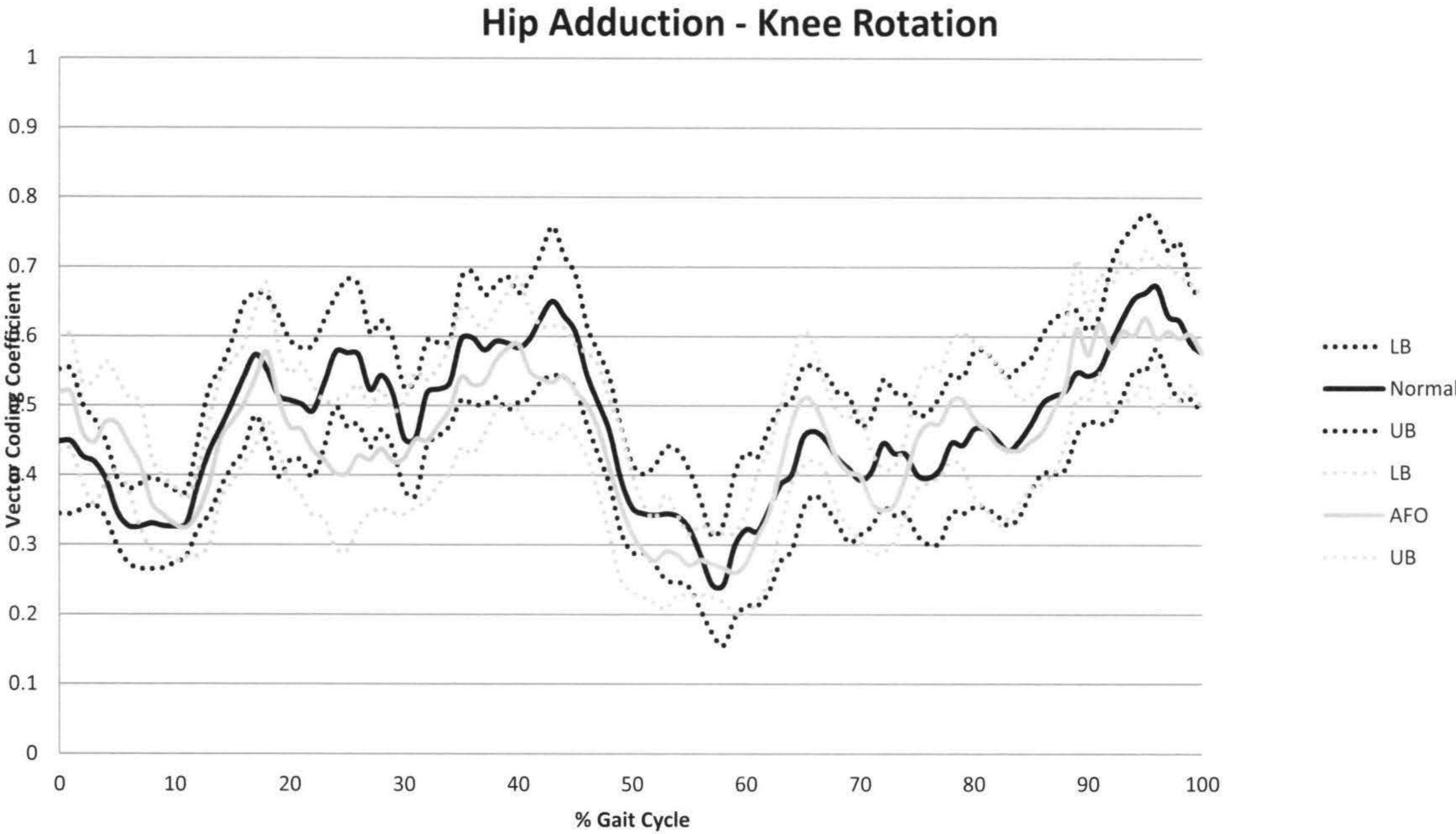

Figure 15. Hip adduction-knee rotation group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

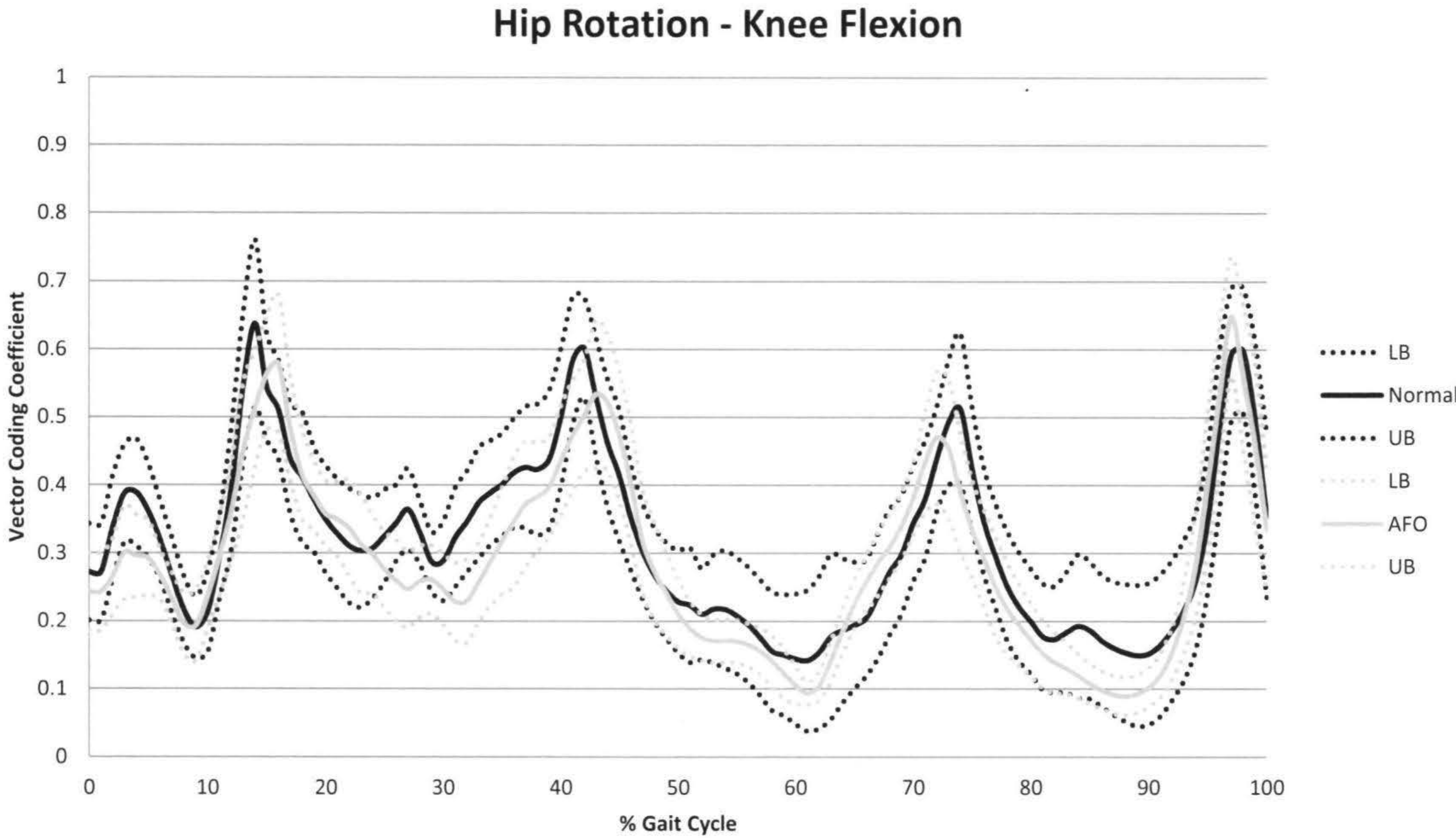

Figure 16. Hip rotation-knee flexion group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

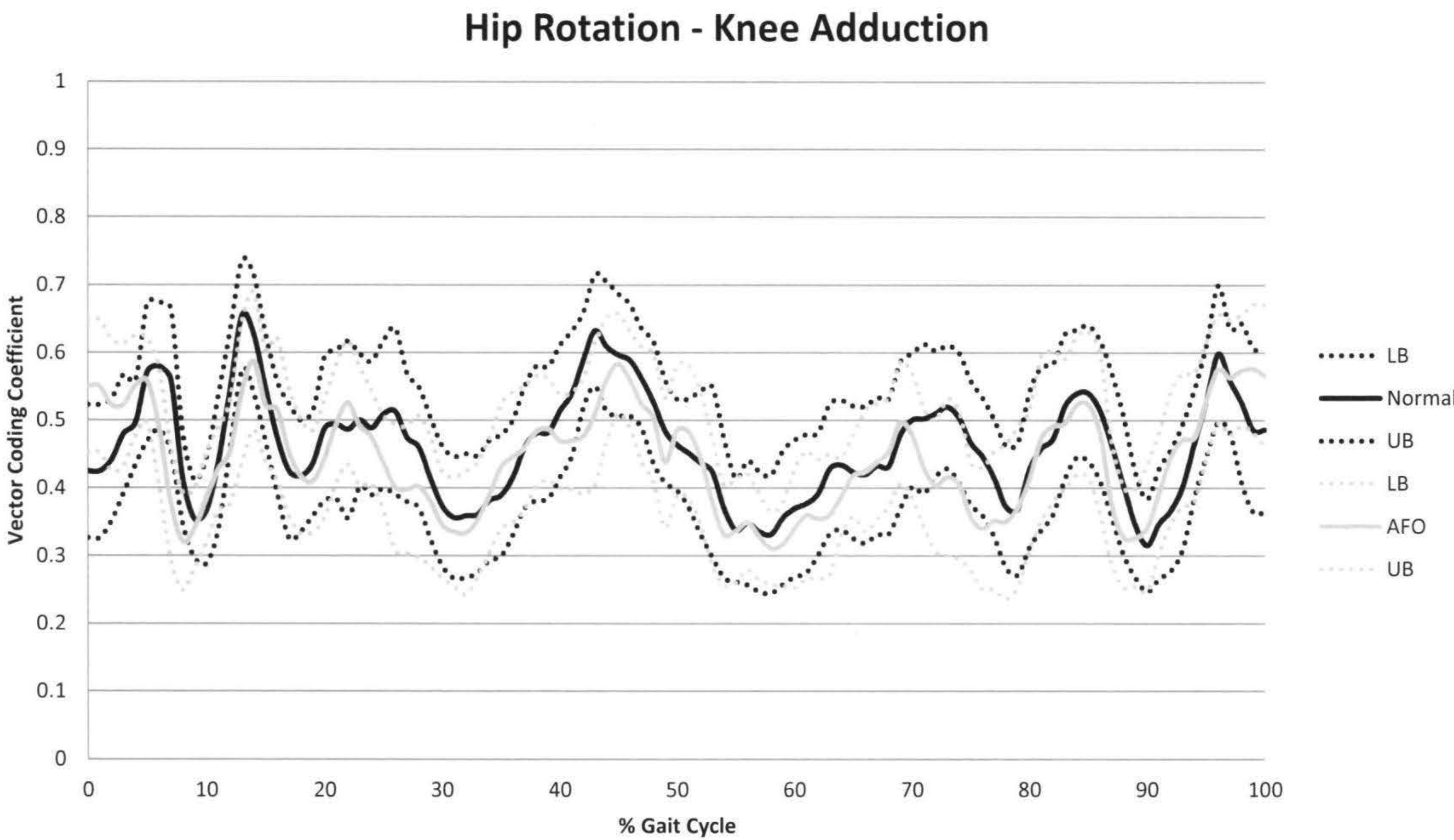

Figure 17. Hip rotation-knee adduction group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

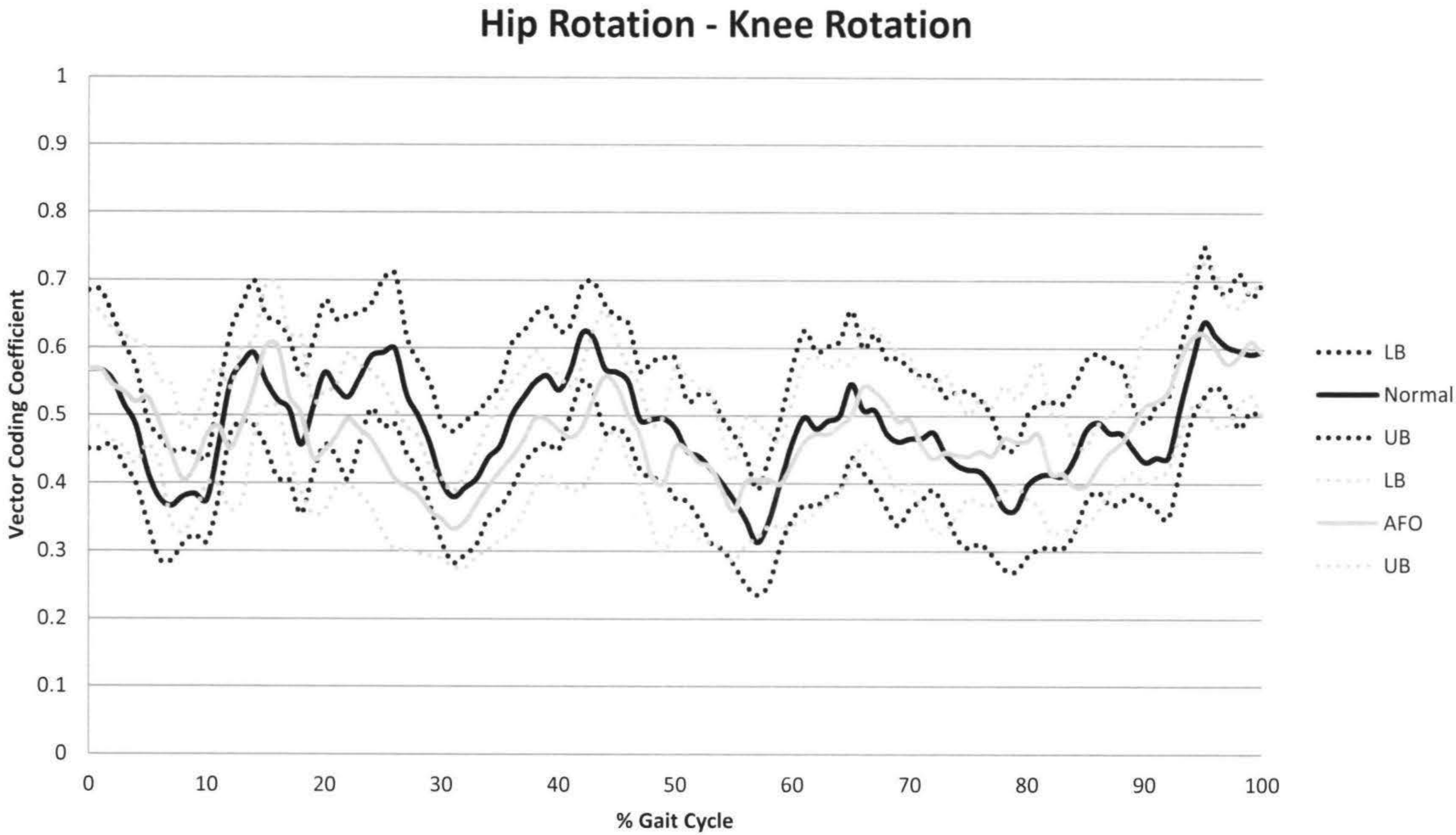

Figure 18. Hip rotation-knee rotation group mean (solid lines) vector coding coefficient curves and confidence interval bands (dotted lines) during one complete stride from normal gait (black lines) and AFO gait (gray lines). Vector coding coefficients closer to 1.0 indicated increased variability.

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VITA

EDUCATION

University of Northern Colorado Bachelor of Science Exercise Science May 2011

Old Dominion University Masters of Exercise and Wellness August 2013

- Thesis: Effects of Ankle Immobilization on Joint Coupling Variability

RELEVANT EXPERIENCE

Old Dominion University Neuromechanics Lab January 2012-August 2012

Biomechanics Research Assistant

- Set-up and collect motion data using Vicon system
- Successfully instruct research participant to perform proper technique for data collection
- Process data in Vicon, OpenSim, and MatLab

The Children's Hospital January 2011-April 2011

Intern in Center for Gait and Movement Analysis

- Successfully set-up specialized equipment for clinical data collection
- Processed data in Vicon and MatLab programs for clinical and research use
- Assist Biomechanist, Physical Therapist, and Pediatric Doctor in gait analysis
- Independently researched for possible future research studies within the Gait Laboratory
- Assist in creating accreditation documents for the laboratory
- Assist with troubleshooting of a variety of technical equipment

COMPUTER SKILLS

ViconNexus Software, MatLab, Microsoft Office Suite, Peoplesoft, Polar TriFit Assessment, Nutritionist Pro, Windows Movie Maker, Audacity