Effects of External Physical Assistive Devices on Gait

by

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DEDICATION

For my best friend and partner, Lee Marietta, and parents, Robert and Dorilla Sansom.

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ABSTRACT

Assistive devices (ADs) are often used by rehabilitation professionals to help individuals walk independently. When using ADs, individuals show observable changes in their walking pattern. However, little research exists objectively documenting the acute, real-time changes in gait that occur despite the significant influence these devices exert on an individual's movement. Understanding the acute changes in movement that occur with use of an AD is a very important component in the decision-making process for rehabilitation professionals who recommend and, often, provide ADs to patients in the hospital or clinic to foster patient compliance and thus, safety. This series of research studies examining external AD use are the first to quantitatively and qualitatively report both overt and underlying acute changes in gait for two unique populations: children with myelomeningocele (MMC) and typically developing (TD) infants learning to cruise.

Overall, our results showed that use of ADs caused changes in gait patterns of children with MMC and TD infants. For children with MMC walking with rigid ADs compared to independently, changes were found in not only basic gait characteristics, but muscle activation patterns, and energy consumption. However, for TD infants wearing a flexible AD around the hips and pelvis while cruising, gait adaptations were more subtle as evidenced by minimal to no changes in segmental angle trajectories and classic gait parameters. Despite the lack of change in overt gait parameters for infants while cruising

in the flexible AD, more apparent adaptations were shown in dynamic representations of cruising behavior (e.g., shifts in state space location for phase plane portrait plots).

The studies presented show that these two unique populations, children with MMC and TD infants, have the capacity and flexibility to acutely adapt their motor control strategies, segment coordination, and movement patterns to application of external manipulations. What we don’t know is if, over the long-term, these adaptations will result in decreased or increased dependency on ADs. Therefore, further research is warranted to investigate impact of these devices on both overt movement behaviors and underlying control mechanisms before we can determine if use of ADs help or hinder functional mobility.

Chapter I Effects of Assistive Device use on Gait and Muscle Activity in Children with Myelomeningocele

Abstract

Children with sensorimotor deficits of the legs, such as myelomeningocele (MMC), often use assistive devices (ADs) to safely optimize their mobility. Two commonly used ADs, posterior walkers and forearm crutches, limit arm use. Our goal was to compare the acute effects of AD use on gait kinematics and muscle activation patterns in children with MMC while walking independently, with a posterior walker, forearm crutches, and a novel AD that promotes reciprocal arm activity, walking poles.

We tested 9 children (5 females) with MMC who were able to walk 4-6 steps independently, without ADs or orthoses. Children walked in 3 trials of 4 conditions (independent, crutches, walker, and poles), randomly assigned. We used a 6-camera Peak Motus^{TM} real-time motion capture system, GAITRite pressure-sensitive walkway, and Noraxon 8-channel EMG system to collect data; all were synchronized.

Our results show that while walking with poles, children spent less time in double support. Center of mass displacement in the anterior/posterior, medial/lateral, and vertical directions was greatest while walking independently and least with the posterior walker. Muscle activity profiles showed individual-specific adaptations to the various ADs, primarily in the children with greater gait impairments. Segmental reversal timing was

similar for the thigh and shank, but delayed in the foot. Segmental displacement was decreased, but greater variability in the trajectories of segments was noted while children walked with devices compared to independently.

In summary, acute use of ADs by children with MMC produced changes in kinematic and muscle activity patterns compared to independent walking, over short distances. Each child showed a unique walking pattern, specific to their degree of neuromotor deficit and their responses to devices were dependent upon both their individual constraints and the affordances provided by each AD. Future research should examine the impact of use of ADs over longer distances and with practice.

Introduction

Assistive devices (AD) are prescribed for patients with motor disabilities to increase safety and improve stability. Research on which clinicians base their recommendations of what AD is most appropriate is very limited (Shoemaker, Lenker, Fuhrer, Jutai, Demers, & DeRuyter, 2009). Decisions regarding which AD to prescribe are often based primarily upon safety with functional mobility a secondary consideration even though maintenance of mobility in adult life represents one of the primary goals of patient care (Vankoski, Moore, Statler, Sarnark, & Dias, 1997). Frequently, the safest devices are bulky, rigid, and may, inadvertently, contribute to further movement restrictions and subsequently create associated health problems, such as decreased aerobic fitness, obesity, increased cardiovascular disease risk, and decreased bone mineral density (Apkon, Fenton, & Coll, 2009; Buffart, van den Berg-Emons, Burdorf, Janssen, Stam, & Roebroeck, 2008a; Buffart, van den Berg-Emons, van Wijlen-Hempel, Stam, & Roebroeck, 2008b; Okoro, Hootman, Strine, Balluz, & Mokdad, 2004; Quan, Adams, Ekmark, & Baum, 1998; Valtonen, Goksör, Jonsson, Mellström, Alaranta, & Viikari-Juntura, 2006). Development of health problems can contribute to further physical limitations, leading to a vicious cycle of lessening physical activity and increasing health problems that may contribute to the need for greater assistance with cares, higher health care costs, and greater morbidity risk. Therefore, early and accurate determination of what AD is most optimal (e.g. not only safe, but encourages physical activity) is imperative for the overall health and well-being of these individuals. Better understanding of acute responses to AD use will allow clinicians to tailor their recommendations more specifically to each individual's unique needs and goals while also facilitating patient safety through better compliance with use of the recommended device. Also, importantly, the recommendation for AD use can then be modified more appropriately and specifically to progressively, incrementally challenge individuals to become more functionally independent and better prepared to withstand perturbations during walking.

We must remain keenly aware though that each child's needs are different and the AD that best meets those needs may change with growth and development, environment, and interests. One population with diverse needs because of the unique nature of their neuromotor deficit is children with spina bifida.

Myelomeningocele

Myelomeningocele (MMC), a form of spina bifida, is the most common central nervous system birth defect in the United States (Davis, Daley, Shurtleff, Duguay, Seidel, Loeser, Ellenbogen, 2005) with a reported birth incidence of 3.7 per 10,000 live births from 1999-2001 (Canfield, Honein, Yuskiv, Xing, Mai, Collins, Devine, Petrini, Ramadhani, Hobbs, & Kirby, 2006). Spina bifida primarily affects the lower spine and often results in varying degrees of sensorimotor deficits in the legs (Spina bifida fact sheet, NIH Publication, 2007). In MMC, one or more vertebrae are either incompletely formed or absent and the spinal cord and meninges protrude through the defect. Resultant sensorimotor deficits are related to lesion level, but with significant variations in functional mobility activities, such as walking. Typically, walking onset is delayed and difficulty walking persists throughout their lives (Gutierrez, Bartonek, Haglund-Akerlind, & Saraste, 2003b). By adolescence, ~50% of children with MMC transition to wheelchair use due to the increasing metabolic demands of walking (DeSouza & Carroll, 1976; Ounpuu, Thomson, Davis, & DeLuca, 2000) leading to greater inactivity, health risks, and social isolation.

In order to determine ways to decrease the metabolic demands of walking for children with MMC and promote long-term walking, researchers have explored the use of various external assistive devices. Chang and Ulrich (2008) provided external lateral stabilization at the pelvis to children with MMC as they walked on a treadmill. Participants decreased lateral trunk sway and energy cost as well as showed a smaller step width, longer step length, and reduced center of mass and pelvic motions in the frontal plane. However, the external lateral stabilization device is too cumbersome for normal walking. Development of an AD that can add external lateral stabilization with fewer restrictions of trunk and upper limb movement, as is seen with walkers and crutches may help people with MMC increase trunk stability control at lowered energy cost.

Efforts to Increase Stability

Two frequently prescribed ADs for children with gait-affected disabilities are posterior rolling walkers (Figure 1.1) and Lofstrand forearm crutches (Figure 1.2). These devices increase opportunities for physical activity, social interactions, and community accessibility. However, they also restrict the normal reciprocal arm movement typically used during walking. When walking with a walker or crutches, the arms are rigid, either to guide the walker or support the body. Thus, these ADs contribute to increased safety and decreased fatigue, but at the expense of the normal reciprocal arm movement with the legs.

Some researchers have shown that reciprocal arm swing during normal walking in typically developing individuals enhances gait stability (Ortega, Fehlman, & Farley, 2008), others suggest it promotes angular momentum (Bruijn, Meijer, van Dieen, Kingma, & Lamoth, 2008; Collins, Adamczyk, & Kuo, 2009; Elftman, 1939; Herr & Popovic, 2008; Park, 2008). Collins, Adamczyk, and Kuo (2009) examined how the mechanics and economics of arm usage contributed to gait. They found that arm swinging exploits natural gait mechanics, contributing to gait economy through reduction of ground reaction moments. When arm swinging was restricted or altered, metabolic cost increased. Pontzer et al. (Pontzer, Holloway, Raichlen, & Lieberman, 2009) proposed that the increase in metabolic cost due to restricted arm swinging also contributed to decreased trunk rotation around the vertical axis. Thus, for individuals with MMC who already have higher energy requirements during normal walking due to altered gait mechanics, use of ADs that provide increased stability, but restrict or alter reciprocal arm movement like walkers and crutches, may place a further burden on energy needs.

Characteristics of Two Common Assistive Devices

Walker. Posterior rolling walkers are commonly prescribed for children of all ages who have disabilities and require substantial stability to facilitate upright activities.

In order to maintain hold on the hand rests of a posterior walker, children pronate their forearms and extend their shoulders (Strifling, Lu, Wang, Cao, Ackman, Klein, Schwab, & Harris, 2008). For children with cerebral palsy (CP), this positioning of the upper limbs has been shown to contribute to improved upper extremity and torso balance as well as upright postural positioning when compared to walking with an anterior walker or no walker (Allen, Villandry, Zurlo, & Tsoumas, 1999; Park, Park, & Kim, 2001; Strifling et al., 2008). Even though this positioning of the torso promotes balance and posture, some researchers have found no change in walking velocity, cadence, or step length compared to walking with anterior walkers (Strifling et al., 2008). However, kinematic and kinetic analyses of the lower limbs did show differences (Strifling et al., 2008). While children walked with the posterior walker, the knee joints showed a diminished loading response and delayed flexion peak along with decreased displacement in the ankle joints. Conversely, Park et al. (2001) and Greiner et al. (1993) found that children with CP walked faster and spent more time in single limb support while maintaining a more upright posture (decreased trunk, hip, and knee flexion angles) while walking with a posterior walker. However, the acute effects on gait imparted by the mechanical design of posterior walkers that restricts arm movement and, therefore, any of the potential benefits afforded by reciprocal arm swing has not been examined in children with MMC.

Crutches. Another common assistive device used by children and adults with disabilities, such as MMC, are forearm crutches, also known as Lofstrand crutches. Clinicians often prescribe forearm crutches for older children, adolescents, and adults, but rarely young children, because proper use of forearm crutches requires a higher level of coordination between the upper and lower limbs as well as adequate upper body strength to support body weight. Forearm crutches exaggerate reciprocal arm movement during walking and distribute much of an individual's body weight through their arms and upper torso since the arms are held in a relatively rigid posture within the device.

Research has shown that when children with MMC walked with forearm crutches compared to independently, the exaggerated reciprocal arm movement resulted in a slower walking velocity with peak force production occurring earlier, establishing stability and preventing falls (Slavens, Sturm, Bajourniate, & Harris, 2009). Other researchers have found that children with MMC, despite a slower cadence while walking with forearm crutches, will show gait adaptations such as increased stride length and hip flexion from terminal swing to mid-stance (Vankoski, Moore, Statler, Sarwack, & Dias, 1997). However, because the arms are held relatively rigid during use of forearm crutches, muscle activity patterns in the trunk, hips, and lower limbs that underlie gait adaptations may be altered as well. In fact, typically developing adults, 4 weeks status post total hip arthroplasty, showed decreases in all temporo-spatial measures and underlying muscle activity patterns (gluteus medius, vastus medialis and lateralis, erector spinae), except the biceps femoris, while participants walked with forearm crutches (Sonntag, Uhlenbrock, Bardeleben, Kading, & Hesse, 2000).

How children with MMC will acutely respond to use of a posterior walker and forearm crutches that do not facilitate normal reciprocal arm swing or and decrease the amount of weight users bear through their lower limbs is unknown. Use of walking poles that promote reciprocal arm swing, trunk stability, and functional mobility may be a viable alternative AD to facilitate gait for children with MMC.

Characteristics of Walking Poles - A Novel Assistive Device

Use of walking poles (Figure 1.3) for sport and exercise in typical adults began in Scandinavia, but has since grown to include many sports enthusiasts in the United States. Despite its' popularity and purported benefits of improved cardiorespiratory fitness, balance, knee joint unloading, and muscle force redistribution, little research on the impact of walking pole use, even within typical adults, has been performed.

While walking poles provide less stability than posterior walkers or forearm crutches, they promote reciprocal arm swing. Research has shown significant decreases in average EMG amplitude for leg muscles, but increases for the arms of typical adults while walking with poles (Foissac, Berthollet, Seux, Belli, & Millet, 2008); the biceps brachii and triceps brachii were the most active of the arm muscles measured (Schiffer, Knicker, Montanarella, & Struder, 2011). These changes in EMG amplitude may have been due to increased arm muscle activity occurring during pole advancement, thus promoting forward momentum of the trunk and reducing demand for leg muscle forces (Foissac et al., 2008). Additionally, propulsive forces caused by handgrips on the poles may contribute to greater mechanical constraints on locomotor and respiratory muscles (intercostals, abdominals, diaphragm, pectorals). Therefore, use of walking poles by children with MMC or other lower limb dysfunctions may provide a more efficient way to redistribute the muscular demands of gait than is possible with either a posterior rolling walker or forearm crutches.

Because of the potential for decreased muscular activity in the lower limbs while an individual walks with walking poles, researchers have investigated how pole use impacts the gait patterns of typical adults. Results have found increased maximal knee joint angle, hip joint angle, foot angle, velocity, stride length, and time spent in stance (Hansen, Henriksen, Larsen, & Alkjaer, 2008; Stief, Kleindienst, Wiemeyer, Wedel, Campe, & Krabbe, 2008; Willson, Torry, Decker, Kernozek, & Steadman, 2001). For children with MMC who typically show smaller stride lengths and slower walking velocities, use of walking poles may promote longer step lengths at a faster walking speed, allowing individuals to spend more time in single limb support; all signs of improved balance (Helbostad & Moe-Nilssen, 2003).

Characteristics of Gait Kinematics and Muscle Activity in Children with MMC

Gait kinematics._Children with MMC show adaptive gait strategies that have been associated with muscle weakness related to their level of lesion (Bartonek et al., 2002), contributing to supplementary recruitment of stronger muscle groups for maintenance of independent walking function (Gutierrez, Bartonek, Haglund-Akerlind, & Saraste, 2005). Often, muscle weakness is found in the hip extensors and abductors as well as ankle plantar flexors, resulting in an independent gait pattern that is characterized by exaggerated pelvic rotation and pelvic obliquity, pelvic hike, increased hip abduction and knee flexion during stance, and increased ankle dorsiflexion (Bare, Vankoski, Dias, Danduran, & Boas, 2001; Camoriano, Cama, Conrad, Andaloro, Gremmo, Albertini, & Frigo, 1995; Duffy, Hill, Cosgrove, Corry, Mollan, & Graham, 1996b; Ounpuu et al., 2000; Vankoski, Sarwark, Moore, & Dias, 1995). These changes in the independent gait pattern result in displacement of the center of mass laterally over the hip joints during ambulation. Displacement of the center of mass laterally over the hip joints during independent ambulation allows individuals with MMC to avoid use of the hip abductors, but still achieve forward center of mass progression via momentum of the body and swing limb (Bartonek et al., 2002; Duffy, Hill, Cosgrove, Corry, & Graham, 1996a; Eames, Cosgrove, & Baker, 1999; Gutierrez, Bartonek, Haglund-Akerlind, & Saraste, 2003a; Gutierrez, Bartonek, Haglund-Akerlind, & Saraste, 2003b). How center of mass displacement and gait kinematics are acutely effected by use of various ADs in children with lumbo-sacral level MMC will provide valuable information for rehabilitation professionals when selecting the most appropriate and functionally advantageous AD for their patients.

Muscle activity. While numerous studies have been conducted measuring center of mass displacement and gait kinematics in children with MMC, few studies have measured muscle activity. Although kinematics provide valuable information for quantification of observable movement during independent walking, measurement of EMG activation patterns is necessary for gaining crucial insight into the underlying muscular adaptations responsible for those movements. Park and colleagues (1997) measured lower limb EMG activation patterns in children with sacral level MMC while walking. Their results showed differences in muscle activity patterns of the medial hamstrings (semitendinosus and semimembranosus), gluteus maximus, gluteus medius, and rectus femoris. Both the medial hamstrings and gluteus maximus muscles showed longer than normal activity during the stance phase, possibly to stabilize the pelvis and control excessive anterior pelvic tilt. The premature swing phase activity shown by the gluteus maximus may be one of the strategies used by children with sacral level MMC to promote forward progression of the swing limb. Interestingly, the rectus femoris showed a significantly different pattern of activity in the three phasic bursts of activity than seen in typically developing individuals. The first burst of rectus femoris activity was significantly prolonged during the stance phase, onset of the second burst before swing phase was delayed, and timing of the third burst at the end of swing was close to normal at the end of swing. As kinematic analyses have shown changes in trunk orientation during gait in children with neuromotor disabilities using ADs, measurement of EMG activation patterns during acute use of ADs will provide crucial insight into the underlying adaptive responses to the external constraints imposed by AD use.

Summary

The goal of independent walking and maintenance of that ability for individuals with MMC promotes greater independence, physical activity, and a decrease in disease risks. Identification of an AD strategy, such as walking poles, that affords upright physical activity while promoting reciprocal arm movement will be valuable for children with MMC and may be of similar benefit to other populations of individuals with lower limb motor control disorders. This study will be the first to examine the acute strategies adopted by children with MMC walking with a posterior rolling walker, Lofstrand forearm crutches, and walking poles. These devices have been studied separately, but never collectively, nor with this population.

Hypotheses

1. Children with MMC will show decreased walking balance (e.g., increased dynamic base of support, decreased walking velocity), but increased stability (e.g., increased step width, increased time in double limb support, decreased stride length) when using all ADs compared to independent walking. Thus, while walking with ADs, children will show:
    a. Decreased step length and velocity.
    b. Increased stride frequency, dynamic base of support, and step width.
    c. More time spent in double limb support
    d. Children will walk faster, with longer, but fewer strides when walking independently due to familiarity and comfort.
2. Children with MMC will adapt their center of mass displacement path, each with their own unique manner, dependent upon the interaction between their unique constraints.
    a. The posterior rolling walker will decrease anterior/posterior, medial/lateral, and vertical COM displacement.
    b. Lofstrand forearm crutches and walking poles will decrease anterior/posterior COM displacement, but increase medial/lateral and vertical COM displacement.
3. Children with MMC will show changes in muscle activity levels and patterns when walking with ADs compared to independent walking. These changes will be

seen more frequently in children with greater gait impairments than children with few or no gait impairments.

   a. The posterior rolling walker will:
      i. Decrease amplitude of the lower trapezius, but prolong muscle activity duration.
      ii. Decrease amplitude and delay onset activation of the hip extensors and rectus femoris.
   b. Lofstrand forearm crutches will:
      i. Decrease the amplitude and show a slight delay for muscle activity onset in the hip extensors (gluteus medius, hamstrings) and rectus femoris.
      ii. Increase the amplitude and show earlier onset for muscle activation in the lower trapezius.
   c. Walking poles will:
      i. Increase the amplitude and show earlier onset for muscle activity in the lower trapezius.
      ii. Decrease the amplitude and delay onset for muscle activity in the hip extensors and rectus femoris.

4. Children with MMC will change the timing of segmental reversals and decrease lower limb segmental angle displacement when using ADs compared to independent walking.
   a. The posterior rolling walker will:

    i. Cause a delay in timing for reversal of the foot segment, but earlier reversal of the thigh.
    ii. Show decreased displacement of thigh, shank, and foot segments.
  b. Lofstrand crutches will:
    i. Cause a delay in timing for reversal of the foot segment, but earlier reversal of the thigh.
    ii. Show decreased displacement of thigh, shank, and foot segments.
  c. Walking poles will:
    i. Cause a delay in timing for reversal of the foot segment, but no change in timing for shank or thigh segments.
    ii. Show increases in thigh, shank, and foot segment displacement.

Method

Participants

We tested 9 children (5 females), 5-12 years old, with MMC in the lower lumbar/sacral region who could walk 8-12 steps independently without an orthosis or assistive device. Participants had had no surgeries over the past year, no current demonstration of neurological compromise (i.e. tethered cord syndrome) or cardiovascular problems, and were able to follow directions appropriately.

Procedures

Upon arrival to our lab for the testing session, we read and reviewed an assent form describing what the testing session for this research study involved and its’ associated risks with participants 5-9 years old; for participants 10-12 years old, a written assent form was provided for them to independently read. All children were asked if they understood what was being asked of them and if they agreed to participate. Their assent was documented and signature obtained, if able to write, by the principal investigator. A parent/legal guardian for all participants read information and signed consent forms. Copies of the assent and consent forms were provided to participant families. All informed consent and assent forms were approved by the Institutional Review Board of the University of Michigan Medical School (IRBMED) that assured this research study followed appropriate ethical treatment and protections for our participants and their families. The principal investigator was readily available for questions from the participant and their parent/legal guardian throughout the assent and consent processes. We asked participants to wear shorts, tank top, or swim suit for the test session, but no shoes or socks. Each participant received a monetary gift for the test session and a t-shirt.

Test Session

Following assent/consent procedures, we familiarized participants with the test area, equipment, and lab personnel. The ADs were a posterior rolling walker (W), Lofstrand forearm crutches (C), and walking poles (P). We adjusted each AD to the proper height for each participant based upon manufacturers' instructions. We taught participants how to use the ADs and allowed them to practice walking with each AD until they performed continuous, alternating steps at a pace comparable to independent walking; up to 15 minutes of practice walking with the various ADs was allotted.

Next, we prepared them for the test session (Figure 1.4). We cleaned skin surfaces with alcohol wipes prior to placement of the electromyography (EMG) electrodes. EMG sensors were placed on muscle bellies, bilaterally for one trunk, one pelvic, and two leg muscles in accordance with the SENIAM project recommendations (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000): lower trapezius (LT), gluteus medius (GM), biceps femoris (BF), and rectus femoris (RF). We collected EMG data with an 8-channel Myosystem 1400A (Noraxon Inc., Scottsdale, AZ) unti at a sampling rate of 1000 Hz.

We attached spherical reflective markers (10mm diameter) to each participant’s body, bilaterally at nine landmarks: dorsal surface of the second metatarsal, lateral ankle malleoli, lateral calcanei, lateral femoral condyles, iliac crests, greater trochanters, acromion processes, elbow epicondyles, wrist styloid processes, and temporomandibular joints. We collected marker position data at 60 Hz with a 6-camera PEAK Motus motion capture system and recorded the test session with a video camera.

Last, we positioned a Polar heart rate monitor across the participants’ chest, below the nipple line, to monitor their exertion level, via heart rate, during testing and ensure that participants were not overexerting themselves. Prior to test onset, we measured each participant’s resting heart rate as they sat quietly. EMG, the motion capture system, and GAITRite walkway were synchronized.

Trials

The order of assignment for walking conditions was random. Each condition received three trials; for each trial, participants walked across a 5.3 meter GAITRite walkway.

Following each trial, participants sat and rested quietly until their heart rate returned to within 5-10 beats of their original resting value. During this time, we asked participants to point to a rating that corresponded to their perceived exertion level during the trial on the Pictorial Children's Effort Rating Table (PCERT) (Roemmich, Barkley, Epstein, Lobarinas, White, & Foster, 2006; Williams, Eston, & Furlong, 1994; Yelling, Lamb, & Swaine, 2002). Units on the scale range from one for "very, very easy" to ten for "so hard I'm going to stop" (Appendix A, Figure A.1).

When participants' heart rates returned to within 5-10 beats of their resting heart rate, they began the next trial. This procedure was repeated until all trials were completed.

After Testing

We removed all reflective markers, EMG electrodes, heart rate monitor and cleaned the participant's skin with alcohol wipes to remove any residue. We asked each participant to fill out a brief questionnaire with their parent/guardian's help focused on their amount and type of typical physical activities, number of siblings, medical history, and use of ADs. Additionally, we measured each participant's weight (kg), standing height (cm), sitting height (cm), foot length (cm), calf length (cm), thigh length (cm), wrist-to-elbow length (cm), elbow to acromion process length (cm), and circumference of the mid-calf, thigh, upper arm, and forearm for each limb (cm). We also measured skinfold thicknesses at seven-sites bilaterally for comparison to norms of typically developed children: triceps, biceps, subscapularis, suprailiac, abdomen, thigh, and calf (Robertson, Cullen, Baranowski, Baranowski, Hu, & de Moor, 1999). Measurements were taken in millimeters three times bilaterally (as participants allowed) for comparison and calculation of the mean and standard deviation.

Data Processing

While all children who participated in this research study were considered community ambulators (Hoffer, Feiwell, Perry, Perry, & Bonnett, 1973), two sub-groups emerged. Five children wore ankle foot orthoses (AFOs) to improve their gait, which we classified as Community Minus (C-). Four did not use orthoses, which we classified as Community Plus (C+). Within participants, data was examined by limb involvement (e.g. more or less). Determination of limb involvement was based on participant’s medical history as reported by their parent. If neither lower limb was identified as ‘more’ or ‘less’ involved by parental report or apparent during observation, the child’s right leg was labeled as ‘more involved’ since most of our participants had a ‘more involved’ right leg.

Three to five steps for each child per trial per condition were included in each analysis based on how many steps each child took within the motion capture area and how clean the marker position and corresponding EMG data were. Marker position data were filtered with a sixth-order Butterworth filter. The kinematic variables of interest included segmental angles for the lower limbs. From the position data for 18 reflective markers, we used custom Matlab programs to calculate center of mass displacement, gait events, and for normalization of individual strides to 100 points. Gait variables normalized according to Hof (1996).

EMG signals were filtered using a fourth-order Butterworth high-pass filter, cut-off at 10 Hz and an 8th order low-pass filter, cut-off at 400 Hz, full-wave rectified to remove high-frequency noise, followed by low-pass filtering at 6 Hz (Hodges and Bui, 1996). EMG trials were then cropped based on gait events (touch down, toe off) and individual strides were normalized to 100 points.

Statistical Analysis

We used one-way ANOVAs to examine the relationships between our dependent variables and the independent variable, AD used during the walking trials. Post hoc pairwise comparisons were performed using a Bonferroni correction. Significance was set at $p<.05$. We did not include Age as a factor because 4 of our 9 participants were 5 years old and all were within the C+ group. No order effect was found within or between conditions. Additionally, due to the small number of participants and the widely divergent walking patterns shown by each participant, we describe, as appropriate, dependent variables by their means and standard deviations. To more closely examine variability, we calculated the coefficient of variation (CV) for the COM displacement values. From the EMG and kinematic data recorded for each participant, we descriptively compared data by condition to baseline, independent walking, condition.

Results

Participants

Anthropometrics. To provide the reader with an idea of the physical characteristics of our sample, please refer to Appendix A (Table A.1). These data show that our participants were shorter 113.94(13.73) cm and heavier 24.42(7.82) kg than children with typical development at the same ages (Kuczmarski, Ogden, Grummer-Strawn, Flegal, Guo, Wei, Mei, Curtin, Roche, & Johnson, 2000).

We compared participants' average skinfold measurements (Appendix A, Table A.2), but did not use equations to calculate body fat since the equations commonly used have been derived for children who are typically developing and, thus, not suitable for use with children with a disability (van den Berg-Emons, van Baak, & Westerterp, 1998). Our participants' had skinfold thicknesses that were higher than what has been reported in the literature for children who are typically developing (McDowell, Fryar, & Ogden, 2009). However, when examined by subgroups, our results show that children in the C+ group tended to have less subcutaneous fat (measured in millimeters) (triceps=12.87(3.97); subscapularis=9.75(4.79); thigh=17.25(6.40); calf=18.50(7.32)) than children in the C- group (triceps=16.52(4.27); subscapularis=9.78(3.34); thigh=24.29(4.11); calf=24.21(3.92)).

Pictorial Children's Effort Rating Table (PCERT)

Our data for the PCERT confirmed that the children in our sample were unable to discriminate, at least in this reporting manner, between levels of exertion across conditions. Children reported that the same amount of effort was required for all conditions, including independent walking.

Gait Characteristics

Center of mass displacement. Figure 1.5 presents the normalized anterior/posterior (A/P) (Figure 1.5a), medial/lateral (M/L) (Figure 1.5b), and vertical

(Figure 1.5b) center of mass (COM) displacements and coefficient of variation (CV) values (Appendix A, Table A.3) across the stride cycle per child per condition. COM displacement was normalized to participants' heights (Bare et al., 2001).

Anterior-Posterior. Figure 1.5a shows that overall, and as a function of subgroup, participants' COM moved forward more with each stride while walking independently than with any of the devices.

To examine variability normalized to each child's own mean, we calculated the coefficient of variation (CV). Overall, CV for the normalized COM displacement along the anterior/posterior (A/P) axis was highest when children walked independently than with any of the devices. Thus, forward motion was generally greatest for the distance per stride in independent walking, but also was the most variable from stride to stride.

When separated by subgroup, children in the C- group showed higher A/P variability when walking independently and with poles than while walking with crutches or walker (Appendix D). However, children in the C+ group showed the least amount of variability when walking independently or with poles, but the highest variability in the A/P directions when walking with crutches or walker.

Medial-Lateral. Overall children showed significantly more medial/lateral (M/L) movement (F(3,24)=3.83, p=.023) while walking with crutches (p=.034) and poles (p=.059) compared to the walker (Figure 1.5b). For the C+ subgroup, children moved M/L less while walking with the walker than independently, but more when walking with crutches and poles (Figure 1.5a,b). However, children in the C- group also moved M/L less while walking with the walker.

Compared to independent walking, the CV for M/L displacement (Appendix A, Table A.3) of the COM was higher when walking with devices. Within subgroups, children in the C+ group showed less variability compared to independent walking while walking with poles and crutches, but higher variability when walking with the walker. However, children in the C- group showed higher variability in the M/L directions while walking with devices.

Vertical. Our data show that overall, children with MMC showed significantly more vertical displacement of their normalized COM (F(3,24)=10.14, p<.001) while walking independently (p=.001), with crutches (p=.029) and poles (p<.001) than the walker (Figure 1.5c). For the C- subgroup, participants COM moved vertically slightly more while walking with poles compared to independent walking, but less when walking with crutches and walker. Children in the C+ group showed similar vertical COM displacement while walking independently, with crutches, and poles, but less with the walker.

When compared to independent walking, overall and the C+ subgroup showed higher CV for vertical COM displacement while walking with devices (Appendix A, Table A.3). However, children in the C- subgroup showed less variability while walking with crutches, but higher variability in the vertical direction when walking with poles and walker.

Spatiotemporal. Appendix A (Table A.4) show the gait characteristics for the overall group and subgroup for each condition, respectively. Overall, participants had a significantly higher normalized walking velocity (F(3,24)=9.42, p<.001), longer normalized steps (F(3,24)=2.73, p=.066, and thus, shorter normalized gait cycle times (F(3,24)=4.77, p=.010), took significantly more steps per minute (F(3,21)=4.91, p=.010) while walking independently than with devices.

Crutches. While walking with crutches, children with MMC showed a trend for walking with wide strides (F(3,24)=2.27, p=.106), at a significantly slower velocity (p=.001) and normalized cycle time (p=.009). When separated by subgroup, children in the C+ group walked with the slowest step rate (p=.008) while using crutches.

Walker. Overall, children walked with short (p=.054), narrow strides, requiring the most time to complete a gait cycle (p=.066), and spending more time in double support (F(3,24)=3.95, p=.020) while walking with the walker. The C- subgroup took the shortest strides (p=.077).

Poles. As a group, children walked at a significantly slower velocity (p=.010), requiring more time to complete a gait cycle (p=.115), but spent the least amount of time in double support while walking with poles (p=.021).

Descriptions of EMG Characteristics

Appendix B (Figure B.1.1, Figure B.1.2, and Figure B.1.3) presents EMG profiles for individual children by condition compared to their independent walking. We calculated ensemble averages, surrounded by one standard deviation envelopes. Due to technical problems, we were unable to compare the EMG patterns for two children, AD and SL, across conditions to independent walking.

Independent. Each child with MMC who participated in this research study showed varying levels of gait impairment. When a healthy adult walks at their self-selected, comfortable pace, they show one major peak for the ipsilateral gluteus medius within the first 15% of the stride cycle (Winter & Yack, 1987) (Figure 1.6a). In general, children with MMC showed similar profiles for the ipsilateral gluteus medius while walking, except for 2 children, CV and JB. Both CV and JB showed a small amplitude peak within the first 15% of the stride cycle while stepping with their less involved legs, but also showed a second, larger amplitude major peak beginning just before toe off that continued into mid-swing, decaying to baseline by the end of the cycle.

Adults show two peaks of activation in the biceps femoris muscle profile (Winter & Yack, 1987) (Figure 1.6b). The first, major peak occurs within the first 25% of the stride cycle, and a second, minor peak occurs from mid-swing to the end of the cycle. The biceps femoris muscle profiles in children with MMC were similar to adults except for two children, CV and JB, whose second peak was delayed in onset, occurring from end stance, peaking at toe off, and decaying to baseline by the end of swing.

In adults, the rectus femoris, on average, shows three occurrences of increased muscle activation across the stride cycle (Winter & Yack, 1987) (Figure 1.6c). The first, major peak occurs within the first 20% of the stride cycle. The second peak's amplitude tends to be smaller, peaking at push off followed by a third increase in muscle activity that begins during mid-swing and continues to the end of the cycle (Winter & Yack, 1987). Children showed similar activation profiles for the rectus femoris, except the second peak appears to have occurred at toe off rather than the middle of end stance as seen in adults. Additionally, two children, MD and CV, showed additional differences from the muscle activation pattern shown by typical adults. MD showed only two peaks;

the first peak occurred later, near toe off rather than end stance, and the second peak was delayed, occurring during mid-swing. CV showed four peaks in muscle activation across the stride cycle. The second peak was notable since it occurred during mid-stance; timing for all other peaks was similar to that of typical adults.

In adults, the ipsilateral lower trapezius muscle shows one major peak during the end of stance (Cappellini, Ivanenko, Poppele, & Lacquaniti, 2006) while the arm is moving into an extended position relative to the trunk (Figure 1.6d). As the arm extends, activation of the lower trapezius muscle promotes stabilization of the scapula and thoracic spine while the contralateral arm is swinging forward, driven primarily by the forward momentum created during walking. Children with MMC showed similar muscle profiles for the lower trapezius, except MD and JB who showed a delay in onset with the single major peak occurring later in the cycle during the end of stance and extending into the beginning of swing.

Crutches. While walking with crutches, children with MMC within the C+ subgroup showed similar muscle activation profiles, timing, and duration, but with a decrease in amplitude for the GM, BF, and LT compared to their independent walking patterns (Appendix B, Figure B.1.1). The RF showed early onset for the first activation peak at the beginning of stance with an overall increase in amplitude. Children in the C- group showed more adaptations in their muscle activation profiles while walking with crutches. For the GM, children showed similar amplitude, but MD and JB did not show the expected peak in activation occurring after touchdown. Instead, they showed only a single, major peak just before toe off. The BF muscle activation pattern showed decreased amplitude. Two children, MD and JB showed a shift from three occurrences of increased muscle activation during independent walking to a single, major peak during the middle to end of stance. Examination of the activation patterns for RF showed similarities, including amplitude, except for MD. She showed an increase in activation frequency for her more involved leg (from two to three) with earlier onset for the first activation, a second burst during end stance, and a delay for the third. In her less involved leg, she showed a shift to a single major activation peak during end stance rather than the two bursts of activation observed during independent ambulation. The LT showed decreased amplitude, but overall similar shape and duration of the muscle activation patterns, except JB and MD who showed earlier onset during stance phase.

Walker. When children in the C+ group walked with the posterior rolling walker, they showed decreased amplitude in muscle activations compared to their independent walking patterns, but similar shape, duration, and timing for the GM, BF, and RF, except EW, TR, and LS (Appendix B, Figure B.1.2). EW showed a delay in onset of gluteus medius muscle activation from initial to mid-stance phase, TR showed an increase in amplitude for RF muscle activation across the cycle, and LS showed earlier onset of the second peak during end stance. The activation pattern for the LT showed inter-participant variation in amplitude with earlier onset and longer duration of activation across the cycle. When children in the C- group walked with the walker, they showed inter-participant variation in amplitude for all muscles. For the GM, MD and JB showed a delay in onset for the first peak, occurring at the beginning of stance instead of touch down. Additionally, CV showed a delay in onset for the second peak, occurring in mid-swing instead of at toe off. Examination of BF shows alterations in the activation patterns for JB and MD. MD showed low level, prolonged activation duration across the cycle for BF. JB showed a shift from three peaks to a single one with onset during mid-stance instead of toe off. The RF activation pattern was similar to independent walking, except for JB. JB showed delay in activation onset for the first peak in the less involved leg, but an overall increase in activation across the cycle in the more involved leg. For the LT, children in the C- group showed decreased amplitude while walking with crutches. MD and JB both showed two peaks of activation instead of one with earlier onset for the second peak during mid-stance. JB also showed a delay in onset for the first peak during mid-stance.

Poles. While walking with poles, children with MMC in the C+ subgroup showed decreased amplitude for the GM, BF, RF, and LT muscles across the cycle compared to their muscle activation patterns produced while walking independently (Appendix B, Figure B.1.3). The shape, duration, and timing of the GM, BF, and RF muscles was similar, except for EW. In the RF, EW showed a delay in onset from the beginning of stance during independent walking to mid-stance while walking with poles. For the LT, children showed a trend for slightly earlier onset of activation during stance, but shape and duration remained similar to that produced during independent walking. When children in the C- subgroup walked with poles, they also showed decreased amplitude for the GM, BF, and RF, but increased amplitude for the LT compared to independent walking. In the GM, CV showed delay in onset for both peaks. In the EMG pattern for MD’s more involved leg, she showed a delay in activation onset to mid-stance, with a multimodal pattern; in her less involved leg, she showed two activations, instead of one, with the second peak occurring during end stance and showing multimodal activity through the beginning of swing. JB showed a shift from two activations to one that extended from mid to end stance. The LT showed a trend for increased amplitude with prolonged duration of multimodal activations across the stride cycle. However, the LT and BF showed considerable differences between the activation pattern recorded during independent walking and with poles for all children both within and between individuals and legs for amplitude, timing, frequency of activations, duration, and shape. For the RF, children showed decreased amplitude, delay in activation onset from touch down to mid/end stance with prolonged, multimodal activation continuing through swing.

Segmental Angles

In order to determine the impact of combined muscle activity on limb segment displacement, we examined timing of limb reversals and displacements of segmental angles for the thigh, shank, and foot in the more involved and less involved (as applicable) limbs as children walked with ADs compared to independent walking. We calculated ensemble averages with a one standard deviation envelope for segmental trajectories across the stride cycle using 3-5 ‘clean,’ typical strides per child for each condition. For one child, CV, we were unable to calculate segmental angle data for segments that included reflective markers on the greater trochanters due to technical problems tracking markers during trials. Please refer to Appendix C (Figure C.1.1, Figure C.1.2, Figure C.1.3) for lower limb segmental angles of individual participants.

Timing of limb reversals. Overall, children showed similar timing for the reversal of thigh and shank segments when walking with devices compared to independently. However, most children in both groups showed a delay in timing for reversal of the foot segment while walking with devices compared to independent walking.

For the thigh, shank, and foot segments, children in the C+ group showed a trend for no change or a delay in reversal timing while walking with devices compared to independent walking. However, children in the C- group showed a trend for either similar or earlier timing in reversal of the thigh segment while walking with devices compared to independent walking. Reversal timing of the shank and foot segments remained unchanged for many of the C- children. When walking with crutches and walker, children in the C- group showed more frequent, earlier thigh segment reversals of the more involved leg, and delayed foot segment reversals in the less involved leg. While walking with poles, children in the C- group showed no change in thigh or foot segment reversal timing.

Segmental angle displacement. Refer to Appendix A (Table A.5) for maximal, minimum, and displacement data for thigh, shank, and foot segments by limb involvement for the overall group and subgroups. Children with MMC showed decreased shank and foot displacement while walking with devices compared to independent walking. When children walked with poles, they showed increased thigh segment displacement, but decreased thigh segment displacement while walking with the walker. While walking with crutches, children showed increased displacement in the less involved thigh, but less in the more involved thigh. When separated by subgroup, children in the C+ group showed slight increases in thigh segment displacement while walking with poles and crutches than either independently or with the walker. For the C- group, children showed pronounced asymmetries between limbs, dependent on limb involvement, in displacement of the shank and ankle segments. Thigh segment displacement showed greater symmetry between lower limbs during walking with poles and independently. The more involved limb's thigh segment showed decreased displacement while children walked with crutches and the walker than poles or independently.

Variability in segmental angles across the gait cycle. In general, while children walked with ADs, they showed more variability in trajectories for limb segments when compared to independent walking. However, no trend was noted for effect of a specific AD on the pattern shown. Overall, thigh and shank segments showed less variability than foot segments. Children showed the greatest increase in variability in consistency as they prepared for, and during, the swing phase, but typically showed a decrease in variability while preparing for touch down.

Children in the C+ group showed decreased variability in the trajectory of the thigh segment across the gait cycle while using devices compared to independently. For children in the C- group, less variability was shown in trajectories of shank segments. Interestingly, three children in the C- group, CV, JB, and MD showed no change in trajectory of their more involved shank segment while walking with devices compared to independently.

Discussion

Our purpose in this research study was to compare the effects of two commonly used assistive devices and a novel one, walking poles, on the kinematic and muscle activity patterns of children with MMC as they walked at their self-selected, comfortable pace. Even though the children tested showed heterogeneous gait impairments, all were independent walkers, showing less variability and characteristics of a more balanced gait pattern while walking independently than with any of the ADs tested in this research study. The results of this study are the first to provide an overview and comparison of the acute gait adaptations that occur with use of ADs by children with MMC. By identifying how children with MMC responded acutely to the use of various ADs, we provide insights into how other children with similar motor control disabilities during gait may respond to short-term use (most AD use is short-term). This information will facilitate acute clinician decision-making to determine the most optimal AD for a patient's individual needs for both short and long-term usage, if appropriate. Additionally, examination of acute effects enable improved patient safety and compliance with use of devices because determination of the AD most appropriate will allow clinicians to tailor their recommendations to the specific needs and goals of each individual and thus, may require modification as the individual's status changes.

For our first hypothesis, we proposed that children with MMC would show decreased walking balance, but increased stability while using ADs compared to independent walking. Our hypothesis was partially supported. Children with MMC did show increased walking stability while using ADs, but each AD afforded different benefits for individual children, dependent upon the constraints imposed by their neuromotor system deficits. The increase in stability while walking with ADs was shown by participant's walking slower, taking more steps per minute that were shorter and wider, and taking longer to complete the gait cycle.

Taken together, results for the gait characteristics seem to show that children with MMC showed less variability walking independently, regardless of level of gait disturbance than while walking with devices. When children with more gait impairments (C- group) used the AD that provided maximal stability (posterior walker), they responded by narrowing their gait width, possibly indicating an increase in balance. However, they also showed decreased walking velocity, stride length, and increased time spent in double support, all characteristics of increased stability (Helbostad & Moe-Nilssen, 2003). While these gait adaptations facilitate safety, stability, and may promote energy efficiency, use of a walker may be less socially acceptable, especially as children grow into adolescence and young adulthood. However, when children with MMC walked with poles, they also showed decreased walking velocity, but spent less time in double support than any of the other conditions, possibly indicating improved balance. This finding of decreased time in double support hints at the possibility of walking poles being an alternative AD that may promote maintenance of functional walking ability while being less bulky and more socially acceptable for users.

In our second hypothesis, we contended that children with MMC would adapt the path of progression for their center of mass dependent upon the interaction between each individual's unique constraints and the specific affordances facilitated by each AD. While use of ADs constrained center of mass displacement and decreased variability in the anterior/posterior direction, it also resulted in increased medial/lateral and vertical displacements, especially when children used crutches and poles. Research has shown that during independent walking, control of foot placement in the medial/lateral direction requires greater active neural control than in the anterior/posterior direction for typical adults (Donelan, J. M., Shipman, D.W., Kram, R., & Kuo, A. D., 2004; Kuo, 1999). Thus, it may be that while using crutches and poles, children with MMC showed decreased control or may have been using different control strategies. Determination of which control strategy is most advantageous for these children may, therefore, be dependent on many factors both internal to the individual and external in the environment.

Our results for children with MMC walking independently show that they were “thrusting” themselves forward with each stride, as reflected by increased anterior/posterior and vertical center of mass displacements. Children with fewer gait impairments (C+ subgroup) showed more anterior/posterior displacement while walking independently, but with the least variability indicating greater stability in this direction. However, children in the C- subgroup also showed greater anterior/posterior displacement while walking independently, but with high variability, possibly indicating instability. For these children, use of ADs helped improve their stability in the anterior/posterior direction. Additionally, children with MMC showed increased vertical displacement while walking with walking poles compared to the other ADs, possibly facilitating increases in ankle joint range of motion. Since many of these children wear ankle foot orthoses that restrict foot movement and contribute to altered gait mechanics, use of walking poles may promote a more symmetrical gait pattern. Overall, children with MMC showed acute adaptations in their path of progression for center of mass displacement in all directions, unique to individual constraints and the affordances provided by each AD.

In our third hypothesis, we contended that children with MMC would show changes in their muscle activation patterns while walking with ADs compared to independent walking and that these changes would be more evident in children with greater gait impairments. Overall, our results support this hypothesis. Children in the C+ subgroup (those with few gait impairments, did not wear AFOs) showed few acute changes in their well-established muscle activation patterns while walking with ADs compared to independent walking. However, children in the C- subgroup (those with more gait impairments, wore AFOs) showed acute adaptations in their muscle synergies in a variety of ways. However, our results do not provide a clear indication of whether these acute adaptations are advantageous or detrimental to maintaining safe, independent functional mobility for children with MMC.

Some of the children we tested, like those tested by Park and colleagues in 1997, also showed prolonged activation of the hip extensor muscles, possibly to stabilize their pelvis and control pelvic tilt while walking during all conditions. Because children in this study were acutely introduced to ADs, they may not have had adequate time to fully adapt their gait pattern and motor control strategies to take advantage of the increased support provided by ADs to facilitate greater pelvic and hip control. Thus, some children continued to show prolonged activation of the hip extensor muscles even while using ADs. Additionally, use of the crutches and walker resulted in a delay in activation onset for the hip extensors occurring during the middle/end of initial stance, instead of at touchdown. This delay may have occurred due to the support provided by the ADs, and, in the case of the walker, also by the posterior bar of the device limiting hip extension during the stride cycle.

Surprisingly, our results showed decreased amplitude in the lower trapezius during use of crutches, and poles for children in the C+ subgroup, possibly due to the manner in which forearm crutches position user's arms, shoulders, scapulae, and trunk. Use of forearm crutches promotes arm internal rotation, rounding and forward positioning of the shoulders, protraction and upward rotation of the scapulae, and kyphosis of the upper trunk. This positioning is maintained throughout each stride cycle and inherently inhibits proper activation of the lower trapezius. Because children in the C+ subgroup did not require additional support during ambulation, they may not have used the poles effectively to promote reciprocal arm swing. However, children in the C- subgroup did show an increase in lower trapezius amplitude while using poles, possibly due to active engagement of the shoulder and scapular stabilizers facilitated by reciprocal arm swing. Also, during use of the walker, children showed prolonged lower trapezius muscle activation across the stride cycle possibly due to absence of arm swing and positioning of the participant's trunk and upper extremities in a highly extended posture, contributing to an isometric contraction of the lower trapezius across the cycle.

Overall, children showed decreased amplitude for the pelvic and leg muscles while walking with devices compared to independent walking, possibly due to the support provided by the devices contributing to the need for less muscular force production for performance of strides during ambulation. During AD use, some children, primarily those in the C- subgroup, showed a shift from multiple bursts of muscle activity across the gait cycle to single bursts of muscle activity. This may have occurred in response to the unique support provided by each AD. Because these children, typically, work harder to maintain their independent functional mobility, they may have more experience and are better able to adapt to various types of assistance. Therefore, they may have been able to quickly and efficiently identify where in the gait cycle activation of specific muscles would result in the greatest benefit for continued gait performance.

Children in the C- subgroup showed considerable asymmetries between activation patterns in their legs, dependent on limb involvement, primarily during the crutch and walker conditions. This may have been because use of these devices, which provide the greatest support, allowed children to explore greater varieties of activation patterns and timings within their muscles. Since we only measured four muscles, we can only speculate, but future research may be able to explore this question more specifically.

In our final hypothesis, we proposed that there would be changes in the timing of segmental reversals and segmental angle displacements in the lower limbs as children walked with ADs compared to independent walking. Our hypothesis was predominantly supported by our results. While children did show delays in timing for reversal of the foot segment when using any AD, they showed similar timing for reversal of thigh and shank segments. However, closer examination of the subgroups shows that children in the C-group showed earlier reversals for the thigh segment while walking with crutches and walker, possibly due to the higher level of support provided by these devices, allowing children to begin the stride cycle sooner, even though EMG activations showed delays. These findings may be because children were supporting more of their body weight through their upper extremities while using the crutches and walker, promoting use of passive pendular mechanics to move their legs forward. During use of poles, which provide less support and do not lend themselves to the support of body weight, but instead engage the upper extremities in reciprocal movement relative to the legs, children showed no change in the timing of reversals for their thigh and shank segments. Thus, use of poles may have facilitated active engagement of lower limb musculature throughout the gait cycle.

While we had hypothesized that use of poles would facilitate increased thigh, shank, and foot displacement based on the extant kinematic literature available for typical adults walking with poles (Hansen et al., 2008; Stief et al., 2008; Willson et al., 2001), our results did show increased thigh, but decreased shank and foot segment displacement. This may have been because the results reported were for typical adults with experience using poles. Instead, our results showed the acute kinematic responses of children with MMC, something not reported previously. However, with practice, our participants may show increases in shank and foot segment displacement similar to those found for adults who were practiced users.

Both crutches and walker caused an overall decrease in the displacement of lower limb segments, possibly as a result of the increased support provided and offloading of the lower limbs facilitated by the increased upper extremity support provided by each AD. However, children in the C+ subgroup showed increased thigh segment displacement while using crutches and walker possibly due to their lack of need for these ADs. Instead, they may have used these ADs as “springboards” to promote movement exploration within their proximal legs and hips, a region that may normally be constrained due to the adaptive control strategies they have developed in response to their lower limb sensorimotor deficits. However, for children in the C- subgroup, use of crutches and walkers resulted in decreased thigh segment displacement, but use of poles resulted in increased symmetry between limbs, possibly due to the facilitation of reciprocal upper extremity arm swinging with the legs.

Examination of the variability shown across the gait cycle while children walked independently compared to with ADs showed that children with MMC were able to adapt the coordination patterns of their limbs in space relative to each AD. These acute adaptations allowed participants to take advantage of the constraints and affordances provided by each device, contributing to increased segmental variability in single limb stance. However, variability decreased as the limbs prepared for touchdown when children, especially those with gait impairments, needed to ensure stable foot contact to prevent loss of balance, or worse, falling.

While our results show the acute kinematic and muscular responses of children with MMC while walking with various ADs, something not reported previously, clinicians need to consider how these acute responses will change with continued practice and whether these changes will help individuals optimally meet their functional mobility goals. For the parameters measured in this research study, we would anticipate that children with MMC who were given an opportunity to practice walking with an AD would show changes in their underlying muscle activation patterns. These changes in the underlying muscle activation patterns would manifest over time and with continued practice in more apparent changes in classic gait parameter measurements (e.g., double limb support, swing phase, step width, stride velocity, etc.) dependent on the device used. Therefore, it is imperative for clinicians to consider what parameters they want to specifically target at each stage of an individual's therapeutic training because different devices will have different acute and chronic effects on gait. Use of different devices may be necessary to meet the specific functional mobility goals for different individuals.

Advantages and Disadvantages of Device Use

The prescription of assistive devices by health care professionals for their patients to aid walking safety, independence, participation in upright physical activities, and increase opportunities for social interactions with family and peers has remained a dogma in rehabilitation for a long time. The results of this research study show that acute use of ADs may have facilitated exploration of new segment movement and muscle activation patterns in children with MMC. This was seen when children walked with walkers that provided maximal stability, allowing participants to practice walking with a narrowed gait pattern. Additionally, while children walked with forearm crutches they adopted strategies to improve their stability, but may also have assumed potentially detrimental postural constraints due to the manner in which use of the device positions user’s trunks and arms. While poles provided the least amount of stability for participants, compelling them to adopt a widened base of support and slower walking speed, pole use also showed promise for promotion of increased balance, allowing children to spend less time in double limb support and facilitate reciprocal, neutral arm movement. However, our results also showed decreases in muscular force production while children walked with ADs which could be of benefit by decreasing muscular fatigue, but may also, with long-term use, result in adaptations that could result in weakening of muscles and eventual loss of functional mobility.

Summary

In summary, our goal was to describe the differential effects between independent walking, two commonly used assistive devices, and a novel one that promotes reciprocal, neutral arm movement for children with MMC. We wanted to know how acute use of these ADs affected their walking pattern, muscle activity, trajectories of lower limb segments, and center of mass displacement. Our findings suggest that use of assistive devices by children with MMC does not necessarily imply benefit. Acute differences were found between conditions, indicating that each child with MMC was able to adapt their specific constraints to the unique affordances provided by each particular AD. Further research into the effect of various assistive devices on walking and muscle activation patterns will be beneficial in elucidating the impact of these devices and potential long-term implications of their use.

Limitations

Our study design has possible limitations. There is potential that the introduction and use of each assistive device at the initial testing session may not have provided participants with enough practice for efficient use of the devices. Despite this drawback, some of the assistive devices (i.e. posterior walker, forearm crutches) were already familiar to these participants and not so complicated that extensive training was required. Another potentially confounding factor, may have been participants' lack of familiarity with the lab, equipment, and staff. Due to the age range including very young participants, more time, explanation, and demonstration was occasionally required for them to become acquainted and comfortable with the environment.

An important limitation was the small number of participants recruited. Based on sample size calculations with power set at 0.70 for gait parameters of clinical interest, such as normalized step width and stride length, 40 children with MMC who show gait impairments similar to those of our participants in the C- group will be necessary in future research to detect significance at the $p \leq .05$ level between Conditions. However, due to the relatively rare incidence of myelomeningocele, and limited percentage of children with MMC who are able to walk independently, our sample of 9 participants provided a good foundation for formulating these important clinical questions. Additionally, we only measured four muscles bilaterally while there are hundreds of muscles involved during walking performance. If we had focused on only lower limb, hip, or trunk muscles rather than a combination, we may have found greater consistency in changes generated by the various assistive devices.

Despite the potential limitations, this research study provides an initial comparison of walking with assistive devices for children with myelomeningocele. This information is not only valuable for children with myelomeningocele and their families, but also clinicians to optimize mobility and promote healthy outcomes for this population and potentially others with motor control dysfunctions.

Figure 1.1. Posterior rolling walker.

Figure 1.2. Lofstrand forearm crutches.

Figure 1.3. Walking poles.

Figure 1.4. Five year-old child with MMC prepared for testing.

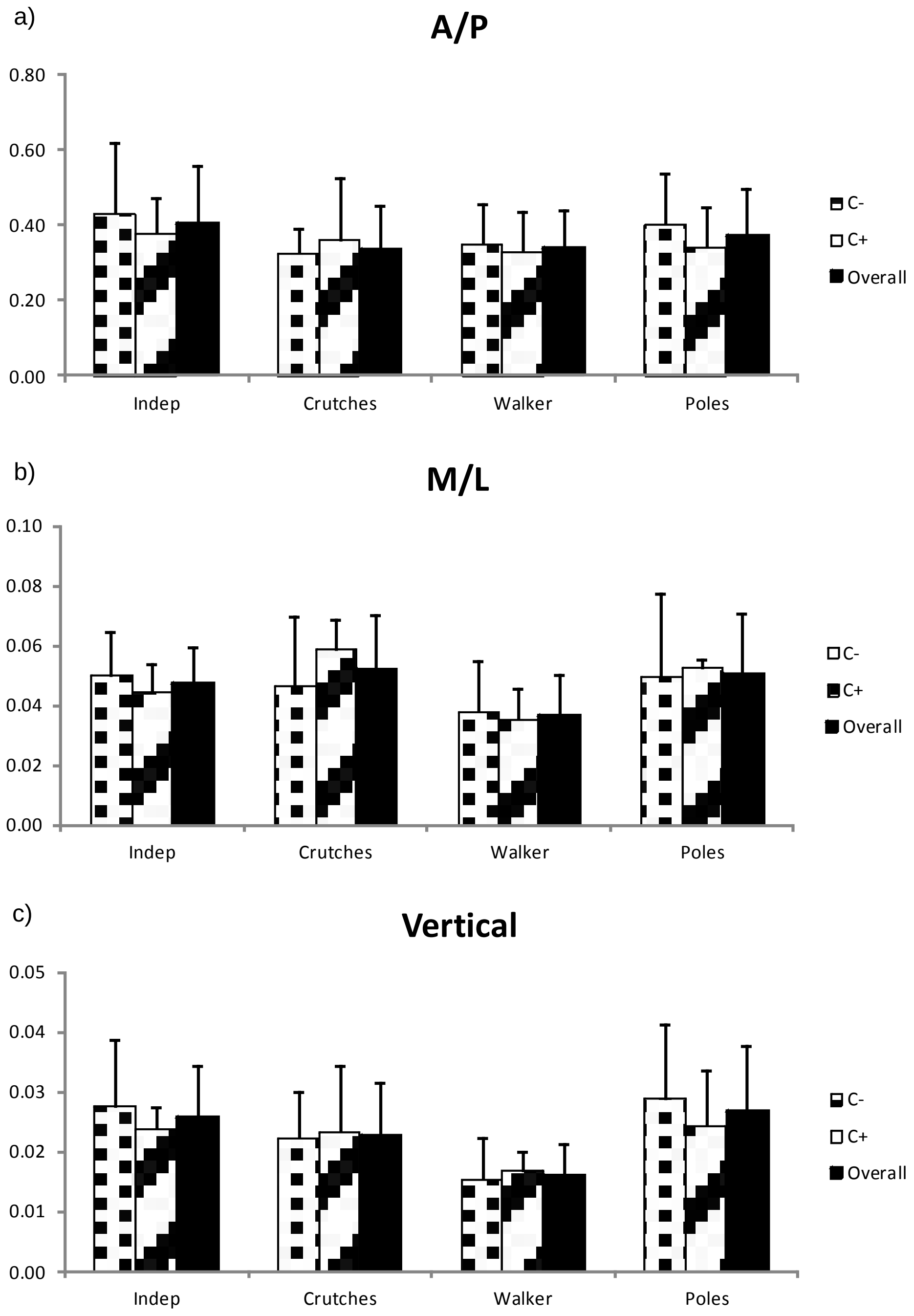

Figure 1.5a,b,c. Normalized center of mass displacements, overall, and by subgroup.

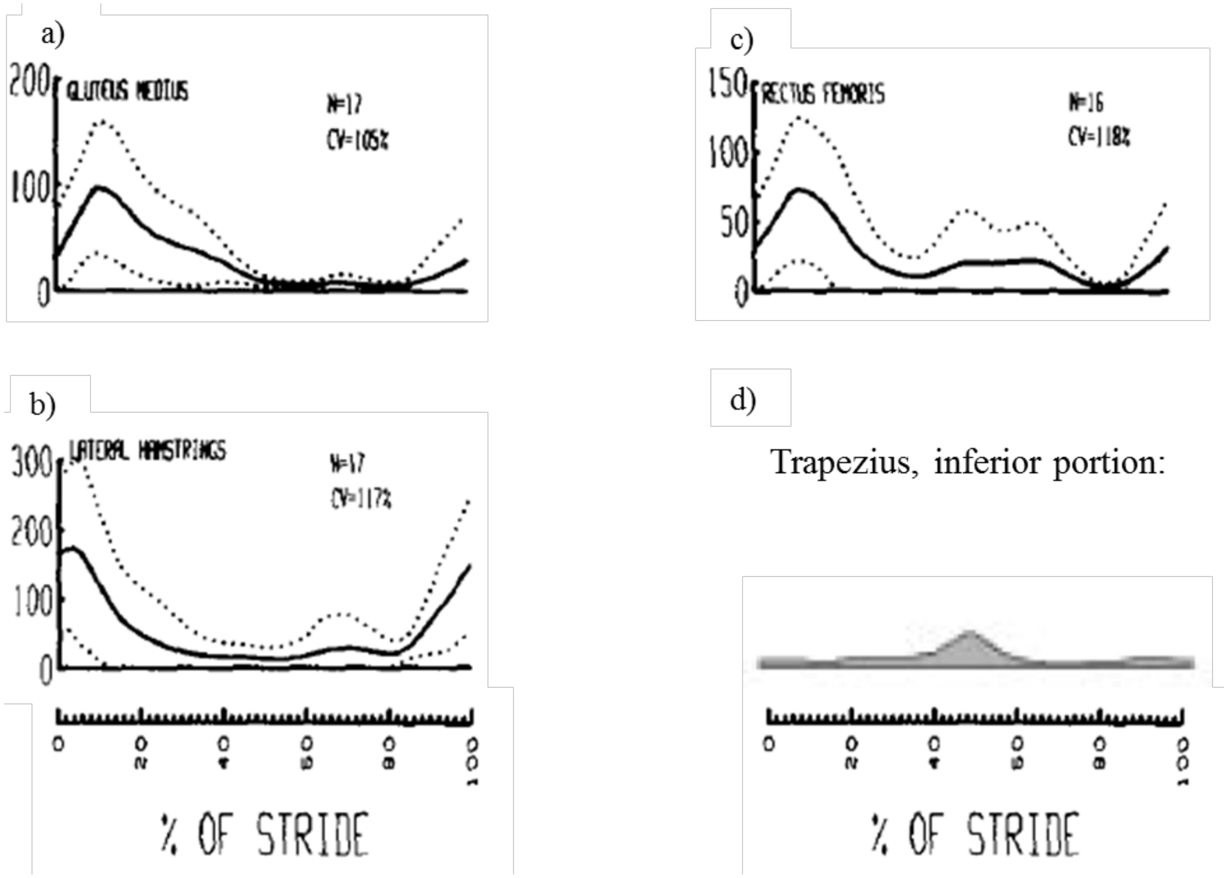

Figure 1.6a,b,c,d. EMG ensemble average profiles during independent walking in typical adults. (Gluteus medius, Lateral hamstrings, and Rectus femoris ensemble averages from: Winter & Yack, 1987; Trapezius, inferior portion ensemble average from: Cappellini et al., 2006).

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Chapter II Energy Consumption and Cost in Children with Myelomeningocele while Walking with Various Assistive Devices

Abstract

As children with myelomeningocele (MMC) enter adolescence and young adulthood, ~50% will transition to use of a wheelchair due to increased metabolic demands. Our goal was to determine if walking poles, a novel assistive device (AD), will increase energy efficiency compared to walkers or crutches in children with MMC.

We tested 8, 5-12 y/o children with MMC in 4 conditions: Independent (I), Walker (W), Crutches (C), Walking Poles (P). They performed 1 trial per condition, randomized, wearing a portable oxygen uptake unit (COSMED K4b^2). All children were considered community ambulators. 4 used ankle foot orthoses (AFOs) (Community Minus = C-); 4 did not use AFOs (Community Plus = C+). Each trial included 3, 5-minute stages: rest, walk, recovery. Children walked at their self-selected pace for all trials.

Overall, children showed increased energy consumption and cost while walking with ADs compared to independently. When separated by subgroups, C+ ambulators had lower net energy consumption (ECS_{net}) while walking I than the C- subgroup. However, the C+ subgroup showed higher ECS_{net} while walking with any AD than the C- subgroup. Coefficient of variation (CV) showed greater ECS_{net} variability across all conditions for

the C- subgroup compared to the C+ subgroup. The highest CV was during I walking and lowest with P. Net energy cost (EC_{net}) showed lower EC_{net} when children in the C+ subgroup walking I than with ADs. Children in both groups had lower EC_{net} when walking I and with P. Calculation of CV showed both groups had lowest EC_{net} variability when walking I.

In summary, our results suggest that for children with MMC, walking with walking poles caused a slight increase in energy cost over independent walking, but reduced cost compared to walking with either a posterior walker or forearm crutches. These results may indicate that walking poles provided ‘just enough’ postural control for these children, but the increased stability provided by the walker and crutches were outweighed by the amount of energy required for their use. Thus, children with MMC may, with practice, remain community ambulators with use of walking poles to facilitate their walking efficiency and stability.

Introduction

Clinicians encourage use of assistive devices (AD) such as walkers and crutches to aid upright mobility, improve balance control, and reduce energy cost for individuals with motor control disabilities. Unfortunately, there is little research to support these proposed benefits. Because of the design and function of ADs, their use alters the normal contribution made by the arms and legs to upright locomotion. Typically, there exists a reciprocal interplay between the upper and lower limbs during ambulation. Reciprocal movement of the arms contributes to walking stability (Ortega, Fehlman, & Farley, 2008), reduces ground reaction moments (Collins, Adamczyk, & Kuo, 2009), and, importantly, reduces energetic costs of walking (Collins et al., 2009; Ortega et al., 2008; Umberger, 2008). Therefore, restriction of contralateral arm movement during use of an AD may, with prolonged use, create walking pattern maladaptations and decrease energy efficiency. These are consequences that children and adults with disabilities cannot afford.

One such population is children and adults with myelomeningocele (MMC), a neural tube defect affecting the integrity of sensorimotor nerves in the lower body, typically resulting in gait impairments. As children with MMC grow, the metabolic demands of walking increase as the upper body continues to develop normally while the lower body decreases in relative size and ability to support their weight (Ounpuu, Thomson, Davis, & DeLuca, 2000). In adolescence and young adulthood, ~50% transition to use of a wheelchair as their primary mode of mobility (Bowman, McClone, Grant, Tomita, & Ho, 2001; Desouza & Carroll, 1976; Thomas, Buckon, Melchionni, Magnusson, & Aiona, 2001). One reason for this is the increasing energy costs associated with maintenance of trunk control (Williams, Anderson, Campbell, Thomas, Feiwell, & Walker, 1983). Unfortunately, use of a wheelchair often leads to greater inactivity as well as increased risk for health problems and social isolation. Health problems may include cardiovascular disease, decreased bone mineral density with associated increase in fracture risk, and obesity (Apkon, Fenton, & Coll, 2009; Buffart, van den Berg-Emons, Burdorf, Janssen, Stam, & Roebroeck, 2008; Buffart, van den Berg-Emons, van Wijlen-Hempel, Stam, & Roebroeck, 2008; Quan, Adams, Ekmark, & Baum, 1998; Valtonen, Goksör, Jonsson, Mellström, Alaranta, & Viikari-Juntura, 2006). Thus, maintaining or improving energy efficient independent ambulation for children with MMC may promote maintenance of independence during social and physical activities while decreasing the incidence and severity of health problems.

Studies of individuals with MMC have shown that the oxygen consumption and cost of independent walking is significantly higher than that of healthy peers (Bare, Vankoski, Dias, Danduran, Boas, 2001; Moore, Nejad, Novak, Dias, 2001). Hypothesized reasons for this difference include: lower levels of physical activity (Schoenmakers, de Groot, Gorter, Hillaert, Helders, & Takken, 2008), adapted gait mechanics, and altered motor performance (Bare, et al., 2001; Gutierrez, Bartonek, Haglund-Akerlind, Saraste, 2003). The adaptations in gait mechanics related to increased energy cost are presumed to result from decreased strength of two primary power generators (hip extensors and ankle plantar flexors) and excessive compensatory pelvic and hip motion in the frontal and sagittal planes (Bare et al., 2001; Gutierrez et al., 2003). Limited or absent muscle activity in the plantarflexor muscles of individuals with lumbar level lesions often leads to a crouched gait pattern and requires excessive isometric activity of the hip and knee extensors to support bodyweight during stance (McDowell, Cosgrove, & Baker, 2002), despite use of ankle foot orthoses (AFOs). Many individuals with MMC wear AFOs to provide additional stability and control of the lower limb through restraint of tibial advancement (Bare et al., 2001). However, without AFOs to control tibial advancement, prolonged recruitment of the knee extensors (Park, Song, Vankoski, Moore, & Dias, 1997) occurs to maintain upright posture, contributing to increases in energy consumption and cost. In addition to AFOs, many individuals with MMC also use ADs to aid functional mobility, safety, upright posture, and facilitate compensation for limited or absent muscle activity.

Little is known about how acute use of an AD affects energy efficiency in children with MMC. Most research studies examining energy expenditure during use of the ADs we tested here (e.g., posterior rolling walker, forearm crutches, walking poles) have involved individuals who had had extensive practice using the AD prior to testing. For this research study, we wanted to know how children with MMC acutely responded to use of these various ADs because if devices are not specifically tailored to each user’s capacity, needs, and goals at the onset of use, the probability of continued compliance is low. For adolescents and young adults with MMC, the trend for transition to wheelchair use due to increasing energy costs has already been documented (Williams et al., 1983). Thus, determination of how children with MMC respond acutely to various devices that may promote decreased energy consumption after only a short duration of practice may facilitate clinician recommendations and thus promote compliance.

Two of the most common ADs used by children with MMC and other individuals with gait disorders are posterior rolling walkers and Lofstrand forearm crutches. For toddlers and children with myelomeningocele, posterior rolling walkers are typically used to help promote walking. These walkers are ‘pulled’ behind the individual, elbows are slightly flexed (approximately 10 degrees), and arms maintained in an extended and internally rotated position while gripping the hand supports (Figure 2.1). Interestingly, this positioning for the upper limbs has been shown, with practice, to enhance upright posture and lower the rate of oxygen consumption in typical children and children with spastic diplegic cerebral palsy more than anterior walkers or no walkers (Park, Park, & Kim, 2001). However, what the acute energy efficiency for children with MMC while walking with a posterior rolling walker is unknown.

Crutches are also prescribed across age groups for individuals with MMC. However, they tend to be used most frequently by older children and adolescents due to the need for both adequate upper body strength to support body weight and ability to coordinate limb movements with two objects-crutches. Lofstrand forearm crutches (the type commonly used) have a cuff that surrounds and provides support to the upper arm in addition to a handgrip. When fit correctly, an individual’s elbow is slightly flexed (10-15 degrees) and the arms typically internally rotated during use (crutches can be moved in parallel or reciprocally) (Figure 2.2). Therefore, the arms do not swing freely, but are instead maintained in a relatively rigid posture throughout the gait cycle. Due to the mechanical constraints imposed by crutches on users and increases in upper arm muscular activation, energy consumption rates for adolescents with MMC using forearm crutches (level of device familiarity not indicated) have been shown to be greater compared to unaided ambulation and typical controls using forearm crutches (Bare et al., 2001; Fisher and Patterson, 1981; Moore et al., 2001).

Because use of posterior rolling walkers and Lofstrand forearm crutches force users to maintain their upper limbs and trunk in a stiffened position, they may not be optimal for upper body muscular recruitment and mechanical advantage. Therefore, identification of a strategy that enables individuals to actively and reciprocally use their arms in relation to their legs may facilitate not only a more typical walking pattern, but also greater energy efficiency. Walking poles may achieve this because they are lightweight, easy to use, socially acceptable, and promote reciprocal use of the arms in relation to the legs during ambulation.

Walking poles have gained popularity for sport and physical activity in typical adults, but, interestingly, are not yet commonly used by disability populations. When walking pole height is adjusted properly, the shoulders are maintained in a neutral position and elbows flexed to approximately ninety degrees (Figure 2.3). While this amount of elbow flexion is not typical during normal walking, pole manufacturers recommend this positioning to optimize the level of support from poles throughout the gait cycle. During walking, pole use promotes upper extremity reciprocal activity which has been hypothesized to facilitate low-level external stabilization of the trunk (Perrey & Fabre, 2008). Thus, walking poles may not only promote a more typical gait pattern and trunk stability, but also increased physical activity for individuals with motor control dysfunctions due to ease of use.

However, studies of walking pole use have only been conducted with typical adults. Results have been variable, impacted by testing environment and surface, grade of incline/decline, and type of poling technique used by participants. When typical adults walk with walking poles on a level track using forceful arm/pole swings, participants show a significant increase in oxygen consumption, heart rate, and caloric expenditure, but no significant change in perceived exertion (Church, Earnest, & Morss, 2002). These outcomes may be due to a re-distribution of muscular demands; arm muscles show increases in average EMG amplitude while leg muscles show decreases during pole use (Foissac, Berthollet, Seux, Belli, & Millet, 2008). Other researchers have shown similar findings of higher oxygen consumption, caloric expenditure, and heart rate, but also increases in rating of perceived exertion (Porcari, Hendrickson, Walter, Terry, & Walsko, 1997). However, when the grade of the walking surface has been increased, and participants instructed to not alter their arm swing in response to changing demands of the walking surface, ratings of perceived exertion decreased. Interestingly, no significant changes in heart rate, minute ventilation, oxygen consumption, or energy expenditure were recorded (Jacobson, Wright, & Dugan, 2000). Conversely, during downhill walking with poles, no significant effect on rating of perceived exertion, heart rate, or preferred walking speed occurred, but an increase in oxygen uptake and energy cost were shown (Perrey & Fabre, 2008). Other researchers have also shown a significant elevation in oxygen uptake and ventilatory efficiency, but did not report any change in rating of perceived exertion despite a significant increase in heart rate (Saunders, Hipp, Wenos, & Deaton, 2008). All of these research studies have involved typical adults. Therefore, how children with myelomeningocele will acutely respond physiologically to reciprocal activity between the arms and legs while using poles during walking is unknown.

Hypotheses

The goal of this study was to determine if walking with a device that promotes reciprocal arm and leg movement, walking poles, will result in improved energy efficiency for children with MMC compared to walking independently, with a posterior rolling walker, or Lofstrand forearm crutches. We hypothesize that acute use of walking poles will:

1. Cause a slight increase in energy consumption and cost for children with MMC who are independent walkers.
2. Result in lower energy consumption and cost than use of a posterior rolling walker for children with MMC.
3. Result in lower energy consumption and cost than use of Lofstrand forearm crutches by children with MMC.

Method

Participants

Participants were 8 children (4 females) with MMC, lumbar-sacral lesion level, aged 5-12 years old, who could walk at least 12 feet without an assistive device. Children were recruited by working with physicians of the University of Michigan Health System and the spina bifida clinic at Sparrow Hospital in Lansing, MI. To be included, participants had no surgeries over the past year, and no current demonstration of neurological compromise (e.g. tethered cord syndrome) or cardiovascular problems. All were community ambulators, but were characterized as C+ if they generally walked without use of an ankle-foot orthosis (AFO) and C- if they typically wore an AFO while walking. We asked that participants not eat for at least 3 hours prior to the oxygen uptake testing. We did not ask participants to fast more than 3 hours because some of them were small children and had a limited ability to refrain from eating for long periods of time (Thomas, Buckon, Schwartz, Russman, Sussman, & Aiona, 2009).

Children wore comfortable clothing they brought with them for the testing session, but walked without shoes, socks, or orthoses. Upon arrival to the lab for testing, we reviewed the purpose and method with each child and their parent to assure agreement with participation (children and their respective parent) provided assent and consent, respectively, during a previous testing session). Once the child positively indicated (verbal endorsement, physically with a nod of their head or thumbs up, written if able) assent to participate, we proceeded with the testing session. Participants were provided with a monetary gift for their participation.

Description of Testing Area

All oxygen uptake testing occurred at the Physical Activity and Exercise Intervention Research (PAIER) lab within the Physical Medicine and Rehabilitation Department at the University of Michigan. The testing area was a large, open room, approximately 60 feet long by 40 feet wide. This space enabled our participants to walk continuously in an 80 foot oval circuit.

Calibration of Oxygen Uptake Equipment

A COSMED K4b^2 portable oxygen uptake unit (COSMED, Rome, Italy) was used to monitor participant’s pulmonary gas exchange concentrations while walking during each condition. Prior to each participant’s testing session, we performed both gas and room air calibrations with the COSMED K4b^2 portable oxygen unit per the manufacturer's instructions as follows:

2) For the gas calibration, we confirmed, with a tank of oxygen and carbon (5% carbon dioxide and 16% oxygen) that the COSMED K4b^2 was able to accurately identify known gas concentrations.
3) Following successful gas calibration, our next step was to calibrate the COSMED K4b^2 relative to environmental room air concentrations (per the manufacturer: 0.05% carbon dioxide and 20.05% carbon dioxide).
4) Lastly, we entered participant's information for age, gender, and mass as well as room humidity and temperature into the COSMED K4b^2 computer software.

Test Procedures

We introduced participants and their parents to lab personnel, the lab space, lab equipment, and each of the assistive devices. Assistive devices included a posterior rolling walker (W), Lofstrand forearm crutches (C), and walking poles (P). We appropriately sized and reviewed how each AD is used with the participant and, as necessary, allowed them time to practice walking with the device until they were comfortable and demonstrated proper use of each AD (determined by the principal investigator). Next, we measured participants' height and weight and asked if they had eaten or drank any items to confirm they had not consumed any calories for the past 3 hours.

We first positioned a POLAR heart rate monitor on the participant's sternum, just below the nipple line, with the strap snugly secured around their trunk. The heart rate monitor was used to monitor the participant's exertion level throughout trials. We adjusted the COSMED K4b^2 harness containing the portable collection unit and battery pack to fit snugly around the participant's trunk (Figure 2.4). A size-appropriate latex-free face mask covered their mouth and nose and was held in place by adjustable straps connected to a headcap. A sampling line connected the face mask with the portable collection unit that was secured on the anterior aspect of the participant's trunk. The portable collection unit was connected to a battery unit secured on the posterior aspect of the participant's trunk (Figure 2.5). Both the portable collection unit and battery pack had antennae that transmitted information, via telemetry, to a laptop computer. The computer recorded and displayed all data on a breath-by-breath basis. Several studies have demonstrated the validity and reliability of the Cosmed K4 telemetric oxygen uptake system for use with children (Boyd, Fatone, Rodda, Olesch, Starr, Cullis, Gallagher, Carlin, Nattress, & Graham, 1999; Corry, Duffy, Cosgrove, & Graham, 1996; Faina, Pistelli, Giulia, Petrelli, & Dal Monte, 1996; Hauswirth, Bigard, & Le Chevalier, 1997; Plasschaert, Matthews, & Forward, 1999). The entire portable COSMED K4b^2 system weighed less than 800 g. Once the COSMED K4b^2 was properly positioned and before testing began, we measured resting heart rate as participants sat quietly.

Testing

Participants performed one 15-minute trial with each assistive device; trial order was randomized. Each trial began with 5 minutes of seated rest followed by 3-5 minutes of walking barefoot with or without the appropriate AD. Participants walked barefoot in order to eliminate the impact of AFO use on muscle activity levels in the lower limb and ensure any changes in measurements of energy consumption and cost were secondary to use of each AD. Participants were instructed to walk at their self-selected, comfortable pace. They walked within the PAIER lab testing space. A video camera on a tripod was placed at one end of the walking circuit and was positioned to view the whole testing space. A researcher walked along with participants to be readily available if the participant required any assistance, but remained slightly behind and to the side in order to limit the influence of their gait pattern on the participant's self-selected gait speed. Although some participants were unable to tolerate ambulating more than 3 minutes during a trial due to fatigue, research has shown that 2-3 minutes is sufficient to establish 'steady state' oxygen consumption in children with MMC (Corry et al., 1996; Duffy, Hill, Cosgrove, Corry, & Graham, 1996; Duffy, Graham, & Cosgrove, 2000). Thus, we accepted these shorter walk durations, as needed. After walking, participants were repositioned in sitting for 5 minutes of recovery. During recovery, we asked participants to indicate, by pointing, their perceived exertion level for each condition to the corresponding rating on the Pictorial Children's Effort Rating Table (PCERT) (Appendix A, Figure A.1) (Roemmich, Barkley, Epstein, Lobarinas, White, & Foster, 2006; Williams, Eston, & Furlong, 1994; Yelling, Lamb, & Swaine, 2002).

Between trials, we removed the face mask, if the participant wanted, while permitting them to recover for an additional 5 minutes before continuing with the next trial. In all cases, we ensured their heart rates had returned to baseline levels. This procedure was followed for all trials until all conditions were completed.

Data Reduction

In order to calculate net energy consumption and cost, we first needed to determine the two separate steady state episodes that occurred during each trial (Figure 2.6). The first steady state episode was identified as 2 minutes during rest when the child's oxygen consumption (VO_2/min) plateaued (<10% change from average value). The second steady state episode was similarly identified during the exercise stage.

For determination of the amount of oxygen inhaled versus carbon dioxide exhaled in each breath, we calculated the respiratory exchange ratio (RER) as VCO_2/VO_2 during the rest and exercise steady state episodes for each trial. Using VO_2 and RER in the following equation, we calculated energy consumption (ECS) during resting steady state (ECS_{rest}) and gross energy consumption during exercise steady state (ECS_{gross}):

ECS (J/kg/min)=(4.960 x RER during SS + 16.040) x VO_2/kg

(De Groot, Takken, Schoenmakers, Tummers, Vanhees, & Helders, 2010; Garby & Astrup, 1987). Thus, net energy consumption (ECS_{net}) was calculated as the difference between ECS_{gross} and ECS_{rest} and provides an important measure of energy consumption per body mass during a specific amount of time (Brehm, Knol, & Harlaar, 2008).

Another important measure of energy expenditure during activity is energy cost, defined as the energy used per unit of distance covered (Brehm et al., 2008). EC is well accepted as an accurate indicator of walking efficiency in clinical gait analyses because it tends to be more sensitive to changes in an individual's condition (Baker, Hausch, & McDowell, 2001; Bowen, Lennon, Castagno, Miller, & Richards, 1998a,b; Plasscheart et al., 1999). Gross energy cost (EC_{gross}) and net energy cost (EC_{net}) (J/kg/m) were calculated by dividing ECS_{gross} and ECS_{net}, respectively, by average walking velocity (meters/minute) during the exercise stage (De Groot et al., 2010). Net EC has been recommended for reporting EC because it provides a more direct indication of walking efficiency and is more clinically meaningful compared to gross measures (Baker et al., 2001; Brehm, Knol, & Harlaar, 2007; McDowell, McLanghlan, Maguire, & Baker, 2001).

Statistical Analysis

For statistical analyses, we ran one-way ANOVAs with repeated measures on condition. Post hoc pairwise comparisons were performed using a Bonferroni correction; significance was set at $p<.05$. We did not include Age as a factor because 4 of our 9 participants were 5 years old and all were within the C+ group. No order effect was found within or between conditions. A Spearman rank order correlation was run to determine if any relationship existed between condition and how children rated their level of exertion (PCERT score) after each condition. Additionally, we also calculated descriptive statistics due to our small sample size. We calculated the mean and standard deviation values for walking velocity during exercise, steady state ECS_{net}, and steady state EC_{net} overall and by subgroup (e.g. C+, C-) for each condition as well as the coefficient of variation (CV) for ECS_{net} and EC_{net} in order to normalize variability to the mean.

Formula for CV:

CV (%)= (standard deviation/mean) x 100.

Results

Participants

Table 2.1 provides individual participant profiles. Mean age was 6.38(2.07) years with height of 113.34(12.60) cm, mass 22.94(4.39) kg, and body mass index (BMI) of

18.33(2.23) kg/m^2. Overall, children with MMC were shorter and had greater mass than children of the same age who are typically developing (Kuczmarski, Ogden, Grummer-Strawn, Flegal, Guo, Wei, Mei, Curtin, Roche, & Johnson, 2000). Between subgroups, children in the C+ subgroup were younger, shorter, and had a lower mass than children in the C- subgroup.

Velocity

Overall, participants walked significantly faster while walking independently (F(3,10)=3.70, p=.050) and slowest with crutches (p=.050) (Table 3.2). When separated by subgroup, results looked quite similar with the exception that children in the C+ group walked considerably slower with crutches compared to all other conditions.

Steady State Heart Rate

During the two minutes of steady state during exercise, children with MMC showed very similar average heart rates across all conditions (Table 3.2). Children in the C+ subgroup had lower average heart rates than children in the C- group across conditions, except when walking with the posterior rolling walker. Children in the C+ subgroup had the lowest average heart rate when walking with crutches, slightly higher while walking with poles and independently, and highest when walking with the posterior rolling walker. Children in the C- subgroup showed lowest average heart rates while walking with the walker, but no difference among other conditions.

Energy Consumption

Children with MMC consumed significantly less energy (ECS_{net}) (J/kg/min) while walking independently (F(3,18)=6.23, p=.004) than with the walker (p=.003) (Table 3.2).

A significant subgroup by condition interaction was found between the crutch and independent conditions (F(3,18)=5.48, p=.007). Examination of the means for the C+ groups shows the independent condition was lowest at 141.79(70.78), followed by poles at 209.54(36.32), crutches at 232.16(52.59), and highest with the walker at 297.28(17.46). For children in the C- group, the lowest energy consumption occurred with crutches at 198.41(81.48), followed by poles at 204.96(79.30), independent at 217.24(66.15), and highest while walking with the walker at 224.03(102.93).

Variability of Energy Consumption during Ambulation

Overall, coefficient of variability (CV) in ECS_{net} was lowest when walking with P=27.58%, compared to W=30.22%, C=30.66%, or I=41.87% (Table 3.2). Within subgroups, C+ children had higher variability when walking I=49.92% than with devices (C= 22.65%; P= 17.33%; W= 5.87%). Conversely, children in the C- group showed less variability when walking I=30.45%, but more with devices (W= 45.94%; C= 41.07%; P= 38.69%).

Energy Cost

Children had significantly lower EC_{net} when walking independently (F(3,15)=16.99, p<.001) compared to the walker (p<.001) and crutch (p=.001) conditions as well as a trend when walking with poles (p=.078) (Table 3.2). When separated by subgroups, the patterns remained with independent walking showing the lowest net EC, followed by poles, crutches, and highest with the walker.

Variability of Energy Cost during Ambulation

In general, children with MMC had the most variability in EC_{net} when walking independently 67.38%. Variability decreased in order: 1) P=56.65%, 2) W=55.78%, and 3) C=48.79% (Table 3.2). The C+ subgroup had lowest EC_{net} variability when walking with crutches at 50.60%, followed by the walker at 57.92%, poles at 60.62%, and highest independently at 63.36%. Children in the C- subgroup had lowest EC_{net} variability while walking with crutches at 54.80%, followed by poles at P=55.50%, independently at 59.19%, and highest with the walker at 59.19%.

Pictorial Children's Effort Rating Table (PCERT)

In order to estimate the strength and direction of association between PCERT score and condition,   we analyzed our data for our respective groups with a Spearman rank order correlation. We found a low, non-significant correlation (ρ=.102, p=.580) between reported PCERT score and condition for the children with MMC tested. However, we found a significant moderate negative correlation (ρ=-,649, p=.002) between PCERT score and years of walking experience, indicating younger children reported higher PCERT scores when rating their exertion level after each trial (Table 3.3).

Discussion

The goal of this study was to determine, in children with MMC, how walking with a device that promotes reciprocal arm and leg movement (e.g. walking poles) would affect energy efficiency and variability compared to walking independently, with a walker, or with crutches. We hypothesized that walking poles would cause a slight increase in energy consumption and cost for children with MMC who are independent walkers, but that this level of energy consumption and cost would be less than when they walked with either a posterior walker or forearm crutches. Our overall results support this hypothesis.

When we analyzed our results by subgroup, a slightly different and more complex picture emerged. Children in the C+ and C- subgroups all consumed more energy while walking with the walker than when walking independently. For the C+ subgroup, children had the lowest ECS_{net} when walking independently, but consumed more energy when walking with all three devices. However, for C- walkers, the lowest ECS_{net} was while walking with crutches, but highest with the walker. This difference in ECS_{net} response between subgroups may have been because children in the C- subgroup, who typically wore AFOs, benefitted more from the additional support afforded by crutches during barefoot walking, but children in the C+ subgroup did not. Instead, acute device use required greater energy expenditure by children in the C+ subgroup during walking, possibly due to lack of need.

Upon examination of variability, we found that children with MMC in the C+ subgroup had higher variability for energy consumption and cost during steady state independent walking than when using ADs. Lowest variability for energy consumption occurred while children used the walker and for energy cost while using crutches. Variability for children in the C+ subgroup may have been greatest during independent ambulation because their exploration of degrees of freedom was not constrained by limitations in strength and balance as it may have been for children in the C- subgroup, but their degrees of freedom may have instead been constrained by the ADs. Children in the C- group also showed greater variability in net energy consumption when walking with ADs compared to independent walking, perhaps because children experimented with more variations in coordination dependent upon whether a device facilitated or constrained their movement patterns. Interestingly, children in the C- group showed the least variability in net energy consumption when walking independently possibly due to adaptations they have developed to remain independent ambulators despite their gait impairments necessitating use of AFOs during walking. Thus, the locomotive strategies developed by C- walkers have allowed them to be more efficient during independent walking than while walking with any of the ADs tested over the same distance, thus resulting in lower EC_{net}.

While the overall results confirmed our hypotheses about differences in energy consumption and cost between ADs, we were also interested in how our participants' independent walking results compared to other studies involving children with MMC and typical development. Most previous research studies have compared oxygen expenditure for children with MMC to typical children, but have not reported true energy expenditure in caloric units or Joules (Bare et al., 2001; Duffy et al., 1996; Moore et al., 2001; Park et al., 2001). However, we felt use of this measure provided a clearer indication of walking efficiency and was more meaningful for this clinical population than reporting our results as oxygen expenditure (VO_2) (Brehm, Becher, & Harlaar, 2007; Brehm et al., 2008; De Groot et al., 2010; Schwartz, 2007). Researchers have shown that children with MMC use more oxygen per unit time and distance walked at their self-selected walking velocity than typical children (Bare et al., 2001; Duffy et al., 1996; Moore et al., 2001). While our results for independent walking echo their findings, we cannot directly compare our results to that of Bare et al. (2001), Duffy et al. (1996), or Moore et al. (2001) because our results reflect actual energy expenditure based on utilization of various substrates (e.g. fatty acids versus carbohydrates) instead of only oxygen expenditure (VO_2).

Fortunately, Brehm et al. (2007) has reported energy expenditure values during walking for typical children of ages similar to our sample's. In our study, the overall group's energy consumption was less during independent walking due to a lower walking velocity, but our participants used more energy to cover the same distance as typical children (Brehm et al., 2007). However, when we compared our sample by subgroups to children with typical development, we found that children in our C+ subgroup still consumed less energy, but now used similar amounts of energy to cover the same distance as typical children (Brehm et al., 2007), indicating better energy efficiency. Children in the C- subgroup consumed and used more energy to independently walk the same distance as typical children (Brehm et al., 2007), indicating less efficiency.

In general, our participants walked at a much slower velocity than typical children (Duffy et al., 1996; Waters, Hislop, Thomas, & Campbell, 1983) and participants with MMC in other studies (De Groot et al., 2010). Children in our C+ subgroup had the lowest ECS_{net} and EC_{net} while walking independently. Interestingly, our C+ subgroup showed lower net energy consumption and cost while walking independently than others have reported for children with MMC (De Groot et al., 2010) who also reported higher gait velocities. Children in our C- group also had lower net energy consumption than what was reported for both groups of ambulators (e.g. household and community) in De Groot et al.'s (2010), but our C-participants showed a higher energy cost with slower walking velocity. The difference between these results may emerge from the higher average velocity at which children in the De Groot et al. (2010) study ambulated. De

Groot et al. (2010) may have encouraged speed more than we did and provided verbal encouragements to their participants during testing. In order to simulate typical walking for our participants, we instructed them to walk at their comfortable, self-selected pace during all conditions and refrained from verbal encouragements during testing. Since we asked participants to walk continuously for 3-5 minutes during each of the 4 conditions, our participants may have inadvertently adopted a slower walking velocity that was more efficient for their “unique physiological and musculoskeletal constraints,” thus allowing them to complete all of the trials (Bare et al., 2001; Bartonek, Eriksson, & Saraste, 2002).

Our results show that children with MMC who are independent ambulators, walked fastest when walking independently, consumed the least amount of energy, and had the lowest energy cost, despite high variability, compared to walking with devices. We contend that this finding may have been due to participants’ extensive, daily practice walking independently and/or the difference in average age of participants within each subgroup. Even though we provided participants with instruction and practice time with ADs prior to testing, use of ADs was still relatively unfamiliar, but independent walking remained the well-practiced condition. However, the walking pole condition appeared to show some promise as an alternative AD that may facilitate improved energy efficiency for these children, especially as they enter adolescence and young adulthood when increasing energy costs associated with maintenance of trunk control (Williams et al., 1983) cause many to transition to wheelchair use for energy conservation purposes.

When we examined the group’s overall response to walking with poles, we found that children showed increased energy cost, but less than the crutches or walker, while walking at a slower velocity compared to the independent condition. These results may indicate that walking poles provided ‘just enough’ postural control for this group of children whereas the benefits of increased control provided by crutches and walkers was outweighed by the amount of energy required to use them. Therefore, if given the opportunity to practice and improve walking velocity, the use of poles may result in increased overall energy efficiency for children with MMC.

While our results show the acute energy consumption and cost for children with MMC while walking with various ADs in a laboratory setting, it is important for clinicians to consider how acute responses will change with continued practice (as noted in the previous paragraph for walking poles). Will these changes help individuals optimally meet their functional mobility goals? Based on our findings and those from research involving well-practiced users of ADs, we anticipate that children with MMC who are given an opportunity to practice walking with an AD will eventually show greater energy efficiency while walking with the device than when compared to independent walking. The rate and degree to which this transition occurs will be unique to the individual’s underlying physiology as well as specific to the device used during walking practice. During initial use, the high level of energy expenditure associated with the use of some devices may be outweighed by the anticipated benefits, making the user less likely to remain compliant. Therefore, clinicians must consider not only what overt gait parameters they want to impact, but also how an individual’s energy efficiency will be impacted over the short and long-term use of specific devices.

In summary, our results suggest that for children with MMC, walking with walking poles resulted in a slight increase in energy cost over independent walking, but reduced cost compared to walking with either a posterior walker or forearm crutches. These results may indicate that walking poles provided ‘just enough’ postural control for these children whereas the benefits of increased stability provided by the walker and crutches were outweighed by the amount of energy required to use those devices. Thus, these children may, with practice, remain community ambulators with use of walking poles to facilitate their walking efficiency and stability.

Limitations

One of the primary limitations to this study is that our sample size was small with age as a potential confounding factor and the conclusions that can be drawn from our results correspondingly limited. Because our participants were a heterogeneous group, division into two subgroups based on AFO use and subsequent gait impairments was necessary, but resulted in the C+ group being composed solely of 5 year old children while the C- group participants ranged from 6-12 years old (Table 2.1). The adjustments in energy expenditure shown by participants in each group across conditions may have occurred because of differences in age with children in the C+ group requiring less energy to walk independently than children in the C- group because they were younger with lower body fat and less independent walking experience. It is possible that children in the C+ group may develop increased gait difficulties similar to children in the C- group with increasing age and experience walking independently, but we did not follow our participants longitudinally to determine if or when this occurred. However, our results do provide very interesting preliminary insights into the energy expenditure needs of children with MMC and the acute effects of AD use, especially for children currently in the C- group who have a higher likelihood of AD use as they enter adolescence and young adulthood for energy conservation purposes.

We also encountered difficulties using the PCERT as an effective measure of exertion with our participants. Many of the children seemed to randomly pick a number, sometimes the same for all trials and sometimes dramatically different (e.g. rating their typical independent ambulation 10/10, but use of crutches 1/10) following each trial. This finding of poor validity for the PCERT within our research study may have been due to our participants being mostly 5 year old children with, possibly, lower cognitive maturity levels compared to other studies that have shown the PCERT to be a valid measure for submaximal exercise intensity in older (12-14 years old) children (Yelling et al., 2002; Roemmich et al., 2006). Lastly, not all participants used the ADs as well as others, despite instruction and practice until deemed proficient before testing. These differences in efficacy of AD use between children may have been ameliorated with a longer time to practice prior to testing (e.g. 3-6 weeks). However, this would have confounded one of the purposes for our study: to examine the acute effects of AD use on energy expenditure. Further exploration of our preliminary findings with a larger sample of children similar to children in the C- group with lumbar-sacral level lesions who are community ambulators would be beneficial in determining the underlying biomechanical and physiological components contributing to the differences in variability observed in this research study.

Table 2.1

Participant profiles

Participant	**Group**	**Age**	**Lesion Level**	**Standing Height (cm)**	**Weight (kg)**	**BMI (kg/m^2)**
LS	C+	5	L4/L5	108.50	21.82	18.54
EL	C+	5	L4/L5	95.30	19.20	21.14
EW	C+	5	L5	112.20	18.40	14.62
TR	C+	5	S1	99.30	18.18	18.44
MD	C-	6	L1	112.90	23.18	18.19
AD	C-	7	L2	122.70	25.50	16.94
JB	C-	7	L4	122.80	27.05	17.94
CV	C-	11	L3/L4	133.00	30.20	17.07
SL	C-	12	L3/L4	137.80	42.00	22.12
Overall						
M (SD)		7.00(2.69)		116.06(14.33)	25.06(7.56)	18.33(2.23)
C+						
M(SD)		5.00(0.00)		103.83(7.86)	19.40(1.67)	18.18(2.69)
C-						
M(SD)		8.60(2.70)		125.84(9.76)	29.59(7.39)	18.45(2.12)

Group: C+=Community Plus, did not wear AFOs for community ambulation;
C-=Community Minus, typically wore AFOs.
Lesion Level: Spinal level at which surgical repair performed.
BMI: Body Mass Index in kg/m^2; an estimation of body fat based on participant's mass and height.

Table 2.2

Summary of variables by group, by condition.

Condition		Velocity (m/s)	CV Velocity (%)	Heart rate (bpm)	CV HR (%)	ECSnet (J/kg/min)	CV ECSnet (%)	ECnet (J/kg/m)	CV ECnet (%)
Overall	Independent	40.70	50.64	130.67	9.51	179.52	41.87	4.72	67.38
	Crutches	31.54	50.76	128.99	10.51	215.29	30.66	7.40	48.79
	Walker	34.18	55.54	130.38	11.24	260.65	30.22	8.90	55.78
	Poles	36.67	54.28	130.49	10.09	207.25	27.58	6.47	56.65
C+	Independent	40.76	51.60	128.14	12.68	141.79	49.92	2.86	63.36
	Crutches	28.63	52.15	124.34	14.45	232.16	22.65	7.49	50.60
	Walker	34.16	57.76	132.52	10.89	297.28	5.87	9.02	57.92
	Poles	37.31	57.79	128.74	12.94	209.54	17.33	5.43	60.62
C-	Independent	40.64	57.63	133.20	6.68	217.24	30.45	6.57	59.15
	Crutches	34.44	55.27	133.65	5.22	198.41	41.07	7.32	54.80
	Walker	34.19	58.69	128.24	13.07	224.03	45.94	8.83	59.19
	Poles	36.24	55.93	132.25	8.25	204.96	38.69	7.16	55.50

Velocity: average speed in m/s (meters per second) at which participants walked in each condition.
CV: Coefficient of Variation; ratio of the standard deviation of a variable to its’ mean.
HR: Heart rate in bpm (beats per minute); average heart rate while participants walked in each condition.
ECS$_{net}$: Net Energy Consumption; measure of energy consumed per minute walked.
ECnet: Net Energy Cost; measure of energy used per meter walked.

Table 2.3

PCERT scores for individual participants, by condition.

	PCERT Score			
Participant	**Independent**	**Crutches**	**Walker**	**Poles**
LS	10	10	3	9
EL	3	3	8	1
EW	1	1	5	4
TR	4	3	2	2
MD	10	4	10	10
AD	1	1	1	1
JB	1	3	3	5
CV	1	1	1	1
Overall				
M (SD)	3.88(3.94)	3.25(2.96)	4.12(3.13)	4.12(3.64)
C+				
M (SD)	4.50(3.87)	4.25(3.95)	4.50(2.65)	4.00(3.56)
C-				
M (SD)	3.25(4.50)	2.25(1.50)	3.75(4.27)	4.25(4.27)

PCERT: Pictorial Children’s Effort Rating Table

Figure 2.1. Five year-old child with MMC walking with a posterior rolling walker.

Figure 2.2. Eleven year-old child with MMC walking with Lofstrand forearm crutches.

Figure 2.3. Five year-old child with MMC walking with walking poles.

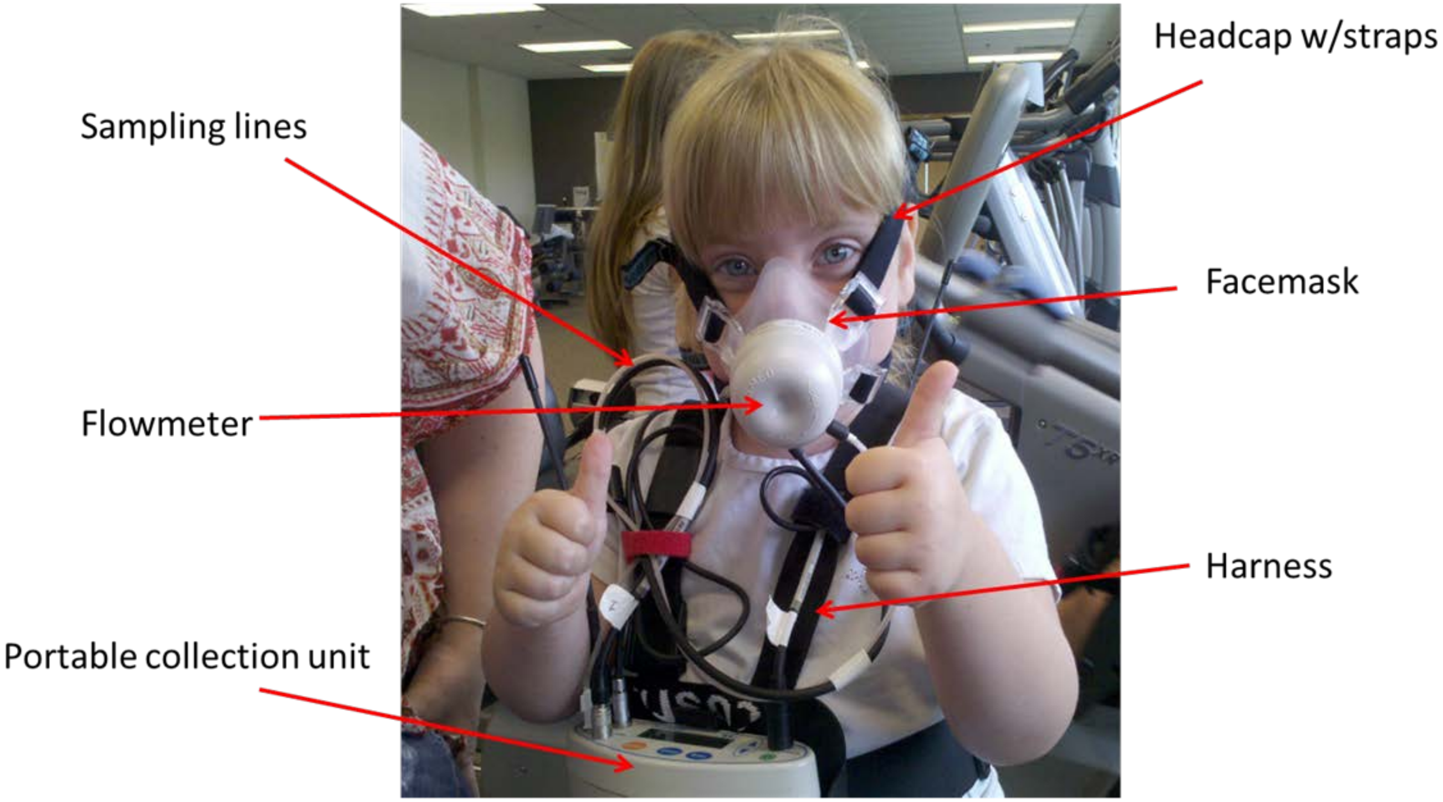

Figure 2.4. Anterior view of COSMED K4b^2 being worn by a 5 year-old child with MMC.

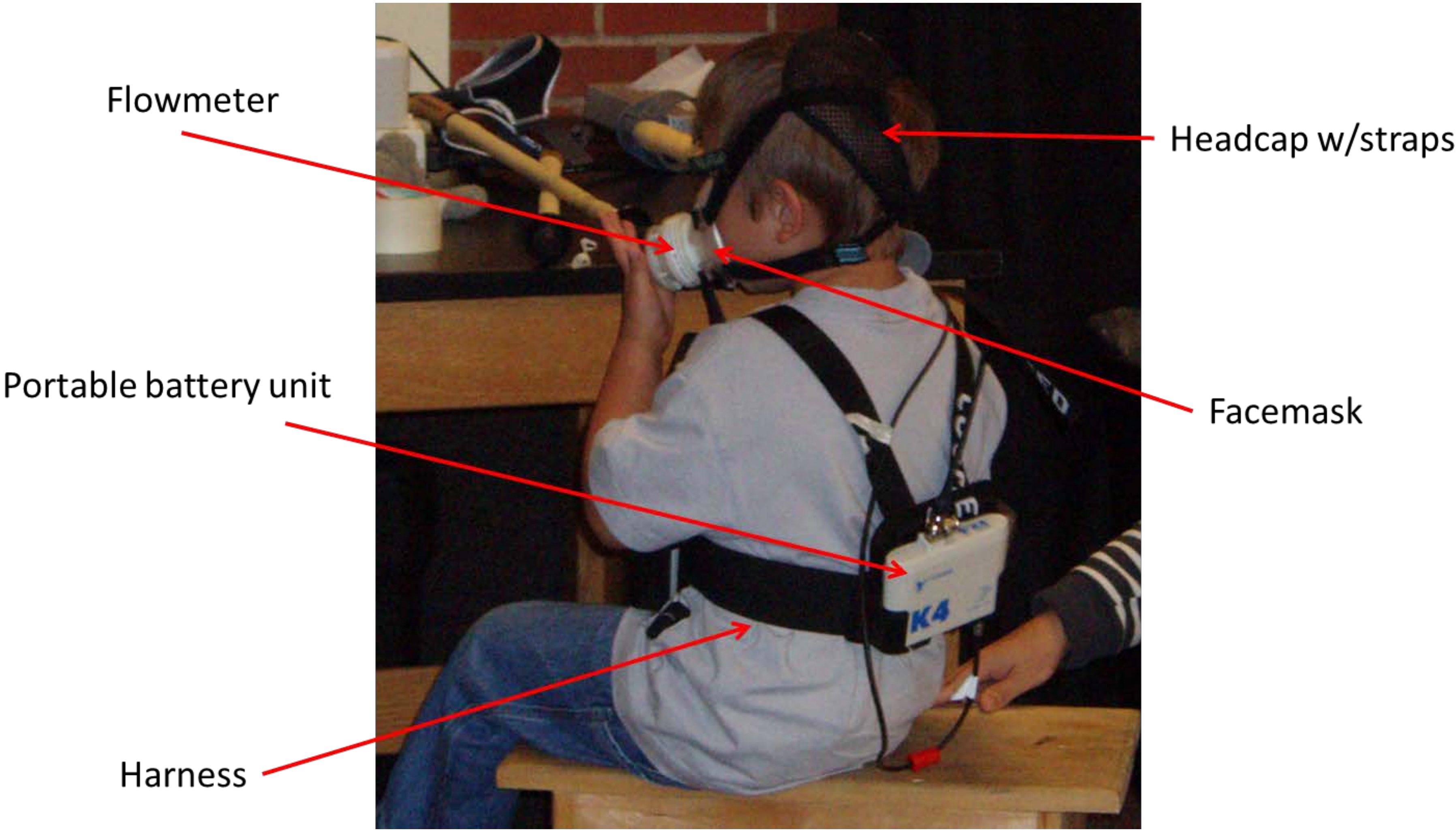

Figure 2.5. Posterior view of COSMED K4b^2 being worn by a 5 year-old with MMC.

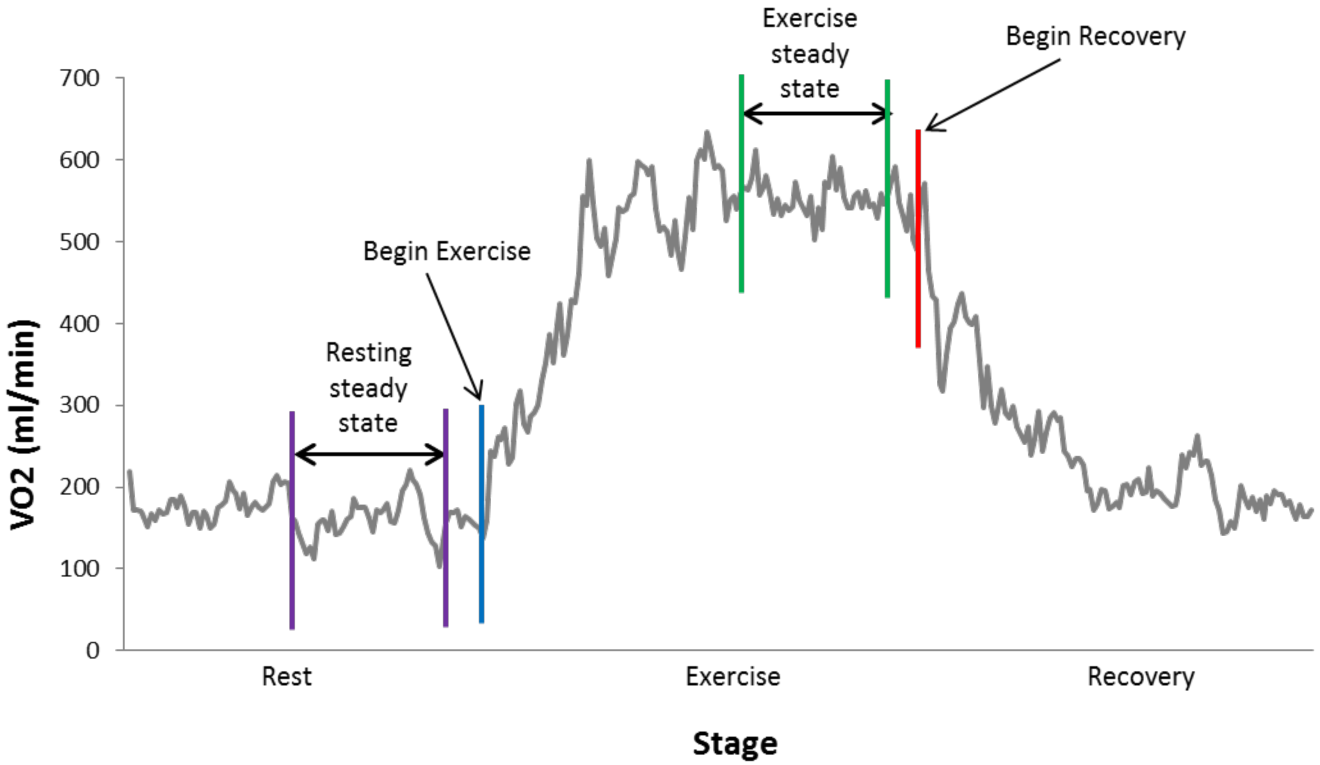

Figure 2.6. Exemplar VO_2 profile for an 11 year-old with MMC during a walking pole trial.

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Chapter III Impact of a Lycra Garment on Locomotion in Infants while Cruising

Abstract

Before infants begin to walk independently, they first learn to cruise (walk with support) while developing limb coordination, control, and balance. However, for infants with motor control disorders, development of these components may be delayed or even inhibited. To promote limb coordination and control for infants with motor control disorders, some clinicians advocate wear of flexible external manipulations, such as lycra garments (LGs), but research supporting use of LGs for infants is lacking. Therefore, we needed to first determine what affect(s) LGs had on the cruising pattern of typically developing (TD) infants.

We tested 9 infants (7 female), 8-11 months old, monthly, from the time they began to cruise until the onset of independent walking. For testing, we placed 22 retro-reflective markers on infants and recorded their cruising performance with an 8-camera Motion Analysis motion capture system. During testing, infants cruised while pushing a custom-made push cart under 2 conditions: diaper-only (control) and while wearing a LG around their pelvis and hips.

Our results show that infants decreased the amount of variability in their step width while wearing the LG compared to only a diaper. Additionally, infants showed more consistency and constraint of segmental motion as well as a shift in location within the state space for leg segments while cruising in a LG across visits.

Overall, infants showed improvements in control of their lower limb segments while wearing a LG when cruising. However, we do not know how infants with motor control disabilities for which the LG is designed and marketed will respond. Therefore, we contend that further research examining use of LGs in infants with motor control disabilities who have greater difficulty learning to control their segments for functional movements is warranted.

Introduction

Learning to walk independently is a complex and dynamic process. The emergence of walking requires adequate strength in the lower limbs and the ability to simultaneously coordinate and control multiple body segments while progressing through space. For infants who are typically developing, the augmentation of these components occurs with engagement in normal, everyday activities and experiences, but for infants and children with motor control disorders, development may be delayed or even inhibited. Thus, of benefit, may be identification of a mechanism, such as wearing of a garment, to facilitate early development of limb coordination and control, critical components necessary for successful independent walking.

Months before infants begin to walk independently, they discover how to cruise, or walk with support (Haehl, Vardaxis, & Ulrich, 2000). This pattern generally follows crawling and precedes independent walking. Research examining this motor skill in infants has been limited, although developmentalists contend that it allows infants to experience, for the first time, upright, self-directed and controlled movement through their environment and promotes repetitive balance practice with variable levels of support from their limbs and surroundings (Adolph, Berger, & Leo, 2010; Haehl et al., 2000). These are purported to be vital elements that build strength and control, contributing to the eventual onset of independent walking.

With development of strength and control during cruising, multiple shifts in the integration of intralimb and interlimb coordination (Vereijken & Waardenberg, 1996) and timing of their actions, reflecting movement experience are shown. When infants begin to cruise, they move only one limb at a time. With experience and practice, they show more complex, overlapping, sequentially timed movements occurring simultaneously in 2-3 limbs (Haehl et al., 2000). Also with increased cruising experience, postural control, as measured by static posture, has been shown to improve (Barela, Jeka, & Clark, 1999; Sveistrup & Woollacott, 1996). However, while cruising, infants use their upper limbs for varying levels of support and stabilization during cruising, resulting in the development of better trunk than pelvic control (Haehl et al., 2000).

While development of trunk control is very important for management of the center of mass during upright and balanced activities (Haehl et al., 2000), pelvic control is also essential to the attainment and maintenance of independent walking. The pelvis adapts to the fluctuations of the upper body, but also links the lower limbs, which both generate forward movement as well as provide the necessary and appropriate level of stability for upright activities. For infants without motor control disorders, pelvic control develops normally and the goal of independent walking becomes a reality. However, for infants and children with motor control disorders, development of pelvic control may never be achieved and independent walking may remain elusive. So, how can rehabilitation professionals facilitate pelvic control during cruising for infants and thus, the development of pelvic stability and onset of independent walking? In order to begin to more closely consider that question, we need to examine the more pervasive research of trunk control development in newly walking infants since research involving infants while cruising is significantly lacking from the extant literature.

To understand how infants' fine tune development of control in their trunk and limbs when newly walking, researchers have used external manipulations, such as loads, to examine the impact of alterations in body mass and proportions on infants' biomechanical parameters. Use of symmetric and asymmetric loads (weighted vests, anklets, wristbands, etc.) have been shown to cause infants to adjust their posture and center of mass position, and therefore, walking stability and pattern (Adolph & Avolio, 2000; Vereijken, Pedersen, & Storksen, 1999). The postural adaptations made by infants to redistribute loads are often accompanied by modifications in their footfall patterns (Chow, Kwok, Au-Yang, Holmes, Chen, Yoa, & Holmes, 2005; LaFiandra, Waggenaar, Holt & Obusek, 2003; Pascoe, Pascoe, Wang, Shim, & Kim, 1997). Limitations in balance control of newly walking infants have been shown to be readily apparent in their characteristic footfall patterns: slow, small, frequent steps with long periods of double limb support (Clark & Phillips, 1987; Ledebt, Bril, & Breniere, 1998), and asymmetric foot rotation (Ledebt, van Wieringen, & Savelsbergh, 2004). Newly walking infants are unable to effectively maintain their balance following the addition of symmetrically distributed loads (15% of their body weight) placed at the shoulders, hips, or ankles, resulting in decreased walking velocity and step length (Garciaguirre, Adolph, & Shrout, 2007). However, after several weeks of independent walking experience, infants will maintain their normal walking patterns despite the addition of symmetrically distributed loads (Vereijken et al., 1999).

While the above studies have focused on the development of locomotor skills and impact of the introduction of loads in newly walking infants, younger infants have also been shown to adapt their movement patterns to the addition of load. When Thelen and colleagues (1987) attached a small weight (185 g) to the legs of 6 week old infants, infants responded with an increase in kick rate, movement amplitude, and velocity of the non-weighted leg. Additionally, 6 month old infants have been shown to be less likely to lean forward to reach when weights are attached to their wrists than when not loaded with weights (Rochat, Goubet, & Senders, 1999). Thus, infants of all ages and for various skills will adapt their motor control strategies, segment coordination, and movement patterns to external manipulations of body and/or limb masses. However, we don't know how infants will respond to an external manipulation, such as a lycra garment, that is intended to realign posture as well as encourage pelvic control and movement in directions favorable to efficient gait.

Lycra garments used by clinicians and rehabilitation researchers in adult and child populations with movement disabilities are believed to provide external, flexible, postural reinforcement promoting support of the hips and pelvis while also facilitating coordination of the trunk and legs (Flanagan, Krzak, Peer, Johnson, & Urban, 2009; Rennie, Attfield, Morton, Polak, & Nicholson, 2000). Unlike the external manipulations (e.g. loads) used in the aforementioned infant studies, a lycra garment with strapping worn around the pelvis and hips will not alter the center of mass, nor the distribution of body mass. Instead, we contend that a lycra garment with strapping worn by infants learning to cruise will alter the biomechanical constraints on standing balance through modification of tissue and joint compression causing increased cutaneous stimulation and joint proprioception (Gracies, Marosszeky, Renton, Sandanam, Gandevia, & Burke, 2000), resulting in greater control of movement in the pelvis and lower limbs. The increase in control of movement in the pelvis and lower limbs will allow infants who are cruising to show a walking pattern that is more advanced than their current experience might predict.

Control of movement is key to the claims made by manufacturers of lycra garments regarding the therapeutic utility of their garments. They propose that elastic properties of the fabric enable immediate and continued improvements in balance, proximal joint stability, postural readiness for movement, inhibition of increased tone, and “inhibition/correction of soft tissue contracture and involuntary movements” in order to facilitate more normal functional capacity for individuals with movement impairments (Blair, Ballantine, Horsman, & Chauval, 1995; Flanagan et al., 2009). One of these manufacturers is TheraTogs and their garments are known by the same name: TheraTogs^{TM}. TheraTogs^{TM} are an orthotic undergarment fabicated from Delta-flex, a lightweight, breathable, and flexible lycra fabric. TheraTogs^{TM} were developed to provide a low-level, passive force to correct imbalance or misalignment by covering the pelvis and upper thighs with a shorts-like garment, on top of which is placed an external strapping system, TogRite^{TM}, to customize the direction and location of force application (Figures 3.1a,b,c).

Many of the therapeutic claims made by TheraTogs have been supported by results from independent research studies involving children and adults with motor control disabilities (e.g. cerebral palsy, Down syndrome, stroke, etc) wearing TheraTogs^{TM}. These research studies have shown that, with practice, participants demonstrated increased proximal stability of the pelvis and distal limb stability (Rennie et al., 2000), increased peak hip extension at terminal stance and increased posterior pelvic tilt (Flanagan et al., 2009), increased gait velocity, cadence, narrowed base of support (Moore, Roth, Killian, & Hornby, 2010), improved postural stability (Fenneman & Ries, 2010), and decreased lateral displacement of the trunk and shoulder girdle with a resultant decrease in incidence of ‘scissoring’ while walking (Rojas, Weiss, & Elbaum, 2008). However, most of these studies involved single subject or small sample sizes with participants wearing their garments 6-8 hours per day for 6-12 weeks. To examine the real-time effects of TheraTogs^{TM} plus strapping wear on muscle activity and temporal-spatial gait parameters, Maguire and colleagues (2009) tested 13 participants following a first unilateral stroke while walking. Results showed significant increase in muscle activity for the pelvic stabilizers as well as improved gait speed and step length symmetry. These studies show that the wear of TheraTogs^{TM} in children and adults with motor control disorders has the potential to be of significant benefit in facilitating functional gait improvements. However, there is no research examining the impact of these garments in infants with developmental disabilities, but in order to lay the foundation for research examining the impact of TheraTogs^{TM} in infants with developmental disabilities, we need to first examine the effects these garments have on infants who are typically developing. For infants who are typically developing, augmentation of the components necessary for independent walking occurs with engagement in normal, everyday activities and experiences, but for infants and children with motor control disorders, development may be delayed or even inhibited. Thus, it could be of considerable benefit to determine whether TheraTogs^{TM} will facilitate perceptible changes in pelvic control and limb coordination during a time when infants who are typically developing are learning an important new skill, cruising, that directly leads to the onset of independent walking.

Our goals in this research project include: 1) to determine if TheraTogs^{TM} significantly change the gait pattern of infants with typical development who are cruising,

2) to determine if cruising experience affects infants' gait adaptations to the wearing of TheraTogs^{TM}, and 3) to determine if wear of TheraTogs^{TM} affects infants' ability to walk independently.

Hypotheses

1. When infants cruise behind a stable push cart while wearing TheraTogs^{TM} compared to wearing only a diaper, they will a.) decrease trunk flexion, b.) increase cruising speed, stride length, time spent in single limb support, and step frequency, c.) show greater symmetry in foot rotation angle, d.) decrease distance between their hands on the push cart handle, d.) increase range of motion of thigh, shank, and foot segments, e.) improve dynamic stability of lower limb segments as evidenced in phase portraits, and increase intersegmental angular coordination as shown in angle-angle plots, and f.) increase intersegmental coordination between the head, shoulders, and trunk in cruising infants as shown by examination of anchoring indices.
2. The impact of TheraTogs^{TM} on cruising patterns will be highest in new cruisers and decrease with cruising experience.
3. TheraTogs^{TM} will provide those infants with the most cruising experience who are closest to walking independently sufficient pelvic stability to take independent steps.

Method

Participants

Participants were 9 infants (7 female) who were tested longitudinally for their gait characteristics while cruising in only a diaper compared to a diaper plus lycra garment. Table 3.1 provides a summary of infant characteristics. Inclusion criteria were no known physical or cognitive disabilities and gestational age ≥37 weeks. Recruitment was conducted through university-sponsored research websites, community flyers, and word of mouth. Parent contact was made before infants began to cruise and we maintained biweekly communication through phone and email correspondence.

We began to observe infant’s motor behavior when parents reported the infant began to pull themselves to stand independently (average age=252 days, range=222-323). Infants began to first cruise laterally with consistent, alternating steps for 6 feet (length of a couch) (average age=273 days, range=236-339) before beginning to cruise forward with consistent, alternating steps 6 feet (average age=298 days, range=250-379) at which time we initiated testing with subsequent testing sessions every 4 weeks until the infant began to walk independently. Infants began to walk independently (defined as 3-5 steps without support) at mean=369 days (range=287-415). Participants were tested an average of four times (including walk onset).

Procedures

All testing occurred in the Developmental Neuromotor Control Laboratory, School of Kinesiology, at the University of Michigan. When families arrived at the laboratory for the first time, we explained procedures and asked parents/legal guardians to sign a consent form approved by the University of Michigan Institutional Review Board. Parents also completed a history survey (e.g., infant’s date of birth, birth weight, birth length, infant's medical history, number of siblings, etc.) and at subsequent visits, provided follow-up information concerning illness, injury, and vacations (factors that may have affected cruising practice).

To prepare infants for testing, we removed all clothing, except diaper, and placed 8 reflective markers (8-mm diameter), bilaterally: dorsal surface of the 3rd metatarsal, lateral malleolus, lateral knee joint, 1/3 of the distance from the lateral knee joint to greater trochanter, 10 cm above the iliac crest along the axillary line, acromion process, olecranon process, and radial styloid process as well as markers on the spinous processes of L4 and C7; infants also wore a headband with 4 markers: one above each ear, one in the middle of the forehead, and one in the center of the posterior head. Lastly, we placed 2 markers on the right side of the push cart (1 near the front and 1 near the rear of the cart), 4 inches from the cart's lateral edge, to monitor infant's path of progression through the motion capture space.

We used a custom-made push cart, 12.7 x 66 x 51 cm (height x width x depth) with a wide, adjustable handle, 91 cm (length), (Figure 3.2a,b) that could be raised or lowered dependent on infant height. The cart was made of wood and had rubber wheels to prevent slippage. Each trial was recorded with an 8-camera Motion Analysis motion capture system at 60 Hz. A 60 Hz digital video camera was positioned on the left side of the testing area for verification of gait events during data capture. The video camera and motion capture system were synchronized.

We positioned infants at one end of the motion capture area with the push cart in front and encouraged them to independently reach for the push cart's handle; however, if infants were distracted, fussy, etc., we encouraged use of the cart by placing their hands on the handle. Infants performed 2 sets of 3-6 trials, dependent on infant arousal and continuity of trial (e.g., if infant fell). For 1 set, infants wore a TheraTog^{TM} garment with TogRite^{TM} strapping over their diaper; for the other set, infants wore only their diaper while cruising. During garment trials we first put on them a TheraTog^{TM} hipster garment (Figure 3.1a) and then added TogRite^{TM} strapping (Figures 3.1b,c); we will subsequently refer to the TheraTog^{TM} with TogRite^{TM} strapping collectively as lycra garment (LG). The LG fit snugly around the infant's waist, just above the level of the iliac crests, per the manufacturer's instructions. We adjusted the thigh portions of the garment such that the bottom edge of the garment fell 4 cm above the knee marker. One strap began on the medial aspect of the infant's inner thigh at the bottom portion of the garment and wrapped anteriorly across the thigh, crossing the lateral thigh at the greater trochanter, around the contralateral pelvis, and attached to the anterior aspect of the iliac crest on the ipsilateral side.

To quantify the level of tension brought about by putting on the TheraTog^{TM} garment with TogRite^{TM} strapping to infants' pelvis and hips as well as to ensure consistency in the application of tension from the garment we used 2 metrics. For the first metric, we measured the circumference of infant's waist just below the level of the umbilicus, pelvis at level of greater trochanters, and thigh across garment trials. These measurements were converted into a ratio relative to the same circumferential measurements without the garment. Additionally, we used a 2.5-degree angled, 20 cm plastic shim (Figure 3.3) with proportionately divided increments to check the uniformity in tension between garment layers and strapping at landmarks such as the greater trochanters. The ratio and shim values were used as benchmarks for garment application during all subsequent testing sessions. Table 3.2 shows the average ratio values at each of the landmarks we measured and the level of consistency we were able to achieve across visits; range of acceptable ratio values: 0.90±0.05 and shim values: 3-5 cm.

Set order was randomized with a balanced Latin square so that half began with the LG and half without. We were unable to randomize trials completely because infants did not tolerate frequent switching between wearing and not wearing the garment. For each trial, we encouraged infants, with assistance from parents and toys at the other end of the walkway, to cruise forward while pushing the push cart across our motion capture space, a distance of 12 feet.

If, during testing, infants showed minimal (e.g. intermittent fingertip contact) dependence upon the push cart for support while cruising, we included 1-3 testing trials in which we attempted to elicit independent walking steps with and without the LG.

After all cruising trials were completed, we measured infants' total body weight, standing height, leg length, thigh and shank length, shank circumference, and foot length. Additionally, we measured, both with and without the LG, abdominal circumference inferior to the umbilicus, hip circumference at the level of the greater trochanters, thigh circumference, and passive hip abduction range of motion. We also administered the motor subscale of the Bayley Scales of Infant Development II (Bayley 1993) to assess concurrent levels of functional motor skill.

Data Reduction

For consistency, both within and between infants, we reviewed the motion capture data to determine noise level, infant arousal level, path of progression continuity, performance of 3-5 continuous alternating steps, and lack of falls. Those trials that did not meet our criteria were not included for further data analysis. For each infant, for each test session, we were able to use 3 trials per condition.

Identification of stride events

Gait events were identified based on the identification of continuous, alternating strides by viewing, frame by frame, the recorded digital videos. The frame at which gait events occurred (toe off, touchdown, end of stance) was recorded. Gait events (touch down, toe off) were used to crop the segmental angle data to individual strides.

We attempted to include three to five steps for each child per trial per condition, but some infants, primarily at their first visit, were only able to perform 2-3 consecutive strides within the motion capture area before falling. Position-time data of the body segment and cart markers were digitized with Cortex Motion analysis software (version 3.3.1.1301; Motion Analysis Corporation, Santa Rosa, CA). Marker position data were filtered with a sixth-order Butterworth filter. We quantified kinematic and spatiotemporal gait characteristics in addition to variability (e.g. standard deviation, coefficient of variation (CV), anchoring index) shown within and between participants. Gait variables were normalized according to Hof (1996).

Phase Portraits

To examine the dynamic behavior relation of leg segments as stride cycles unfolded, we created phase plane portraits. Phase plane portraits plot segmental angular velocity against segmental angles for a single segment, providing a graphical representation of the resultant actions for the underlying motor control mechanisms

(Thelen & Smith, 1994; Winstein & Garfinkel, 1989). Thus, phase plane portraits provide cyclic, dynamic representations of segment (e.g., thigh, shank, foot) motion over multiple cycles (DiBerardino, Polk, Rosengren, Spencer-Smith, & Hsaio-Wecksler, 2010). Different underlying motor control mechanisms can be associated with various trajectory shapes. We adopted the geometric descriptions provided by Winstein and Garfinkel (1989) for our analyses of three consecutive stride cycles:

- *Sharp corners:* movement occurs at a constant velocity, preceded or followed by ballistic motion(s).
- *Vertical sides:* ballistic control; control applied at movement extremes and requiring fast acceleration.
- *Round shapes:* smooth rise and fall of velocity, common to passive pendular motion.
- *Inflections:* movement interruptions; movement velocity in an intended direction is abruptly reduced and then resumed or increased.
- *Loops:* reversals of movement within a cycle.

We did not normalize the phase plane portraits for infants while cruising because the trajectories of their gait patterns were not smooth and the quantitative methods that have been developed for normalization of phase plane portraits have all been based on smooth trajectories (DiBerardino et al., 2010).

Segmental Angle Angle Plots

Because cruising is a learned behavior leading to the onset of independent walking, we also investigated the development of intrasegmental coordination within limbs with angle-angle plots. These plots allow us to visualize the dynamic relation between these segments across multiple cycles. Similar to phase plane portraits, angle-angle plots provide additional insight into the underlying control strategies for movement. We adopted the descriptions provided by Winstein and Garfinkel (1989) for our analyses of three consecutive stride cycles:

- *Horizontal/Vertical segments:* while one segment is changing, the other is constant; suggests decoupled coordination between segments.
- Diagonally oriented, straight line:
    - *Positive slope:* segmental angles are coordinated in phase and change at a constant ratio.
    - *Negative slope:* segmental angles are coordinated out of phase and change at a constant ratio.
- *Turning point synchronization:* directional change for the two segments occurs nearly simultaneously indicating similarity in the relative rates of change for the adjacent segments.
- *Rounded trajectory:* large curvature indicating differences between the relative rates of change between the two segments.

Anchoring Index

To investigate the intersegmental coordination among the head, shoulder, and trunk in infants learning to cruise to the onset of independent walking, we calculated the anchoring indices (AI) for each segment. The AI allows characterization of head, shoulder, and trunk stabilization strategies in the frontal and sagittal planes (Assaiante &

Amblard, 1993; Assaiante, Thomachot, & Aurenty, 1993; Assaiante, Thomachot, Aurenty, & Amblard, 1998). This index allows comparison between the stabilization of a given segment relative to external space and the inferior anatomical segment, revealing whether an individual adopts an "en bloc" or inverted-pendulum stabilization strategy (Assaiante & Amblard, 1993). We calculated the following two indices for each trial to determine normalized anchoring indices for the roll rotation axis (movement in the frontal plane):

1. Absolute angular dispersion values relative to the vertical axis for the head, shoulder, trunk, and leg roll angles during each trial. For roll of the head, we used coordinate data from markers above each ear (2); for the shoulder we used markers on the acromion processes (2); for the trunk, we used markers 10 cm above the iliac crests along the axillary line (2) (hip markers could not be used due to positioning of the lycra garment); and for the leg, calcaneal markers were used (2) (Figure 3.4). Standard deviations of the absolute angular dispersions were then calculated for each trial:
    a. σ_a: angular dispersion of body segment
    b. σ_r: standard deviation of the relative angular distribution of the body segment relative to the axes linked to an interior anatomical segment.
2. Second, we calculated the normalized AI using the absolute and relative segment angles. For example, to compare the trunk roll angle relative to the feet axis (Theta h/r, Figure 3.4a) the relative angular distribution was calculated using the following formula:
    a. $\Theta^t_r = \Theta^t_a - \Theta^f_a$

        i. Θ^{t}_{a}: absolute trunk roll angle with respect to the right leg axis
        ii. Θ^{f}_{a}: absolute right foot roll angle with respect to the external axis

3. For the trunk level, standard deviation values for the absolute roll distribution (σ_a) relative roll distribution (σ_r) were calculated. Finally, the normalized AI was calculated with the following formula:
    a. $(\sigma_r - \sigma_a)/(\sigma_r + \sigma_a)$

Therefore, the trunk AI allows us to examine the degree of dependency between trunk and feet vertical movements. Additionally, shoulder and head normalized AIs were calculated by determining shoulder angle relative to trunk and head angle relative to shoulder axes, respectively. AI values are unitless and vary between -1 and +1. Positive values indicate a tendency for stabilization in space rather than on the inferior supporting anatomical level; negative values indicate better stabilization on the inferior anatomical level rather than to the external space.

We also calculated the AI for the head pitch angle relative to the shoulder axis (Figure 3.4b) (Assaiante and Amblard, 1993). For pitch of the head (movement in the anterior-posterior plane), we used coordinate data from a marker above the right ear and a marker in the middle of the infant's forehead. For trunk pitch, the marker at C7 and right acromion process were used. Calculation of the head pitch normalized AI followed the same steps as outlined above for roll.

Statistical Analysis

For our statistical analyses, we used SPSS (version 20.0.0.1; IBM, Somers, NY). We calculated descriptive statistics for all variables and a 2x4 mixed-model analysis of variance (ANOVAs) with repeated measures on Visit and follow-up Bonferroni post hoc analyses. Infant visits were normalized relative to their own walk onset. For example, the visit that occurred 2 months prior to walk onset is referred to as “-2” and walk onset is referred to as “0”. We used Pearson product moment correlations to examine the relationship between trunk position, foot position, wrist position on the push cart, and variability of step length and width. Significance was set at p<.05.

Results

Participants

To provide the reader with an idea of the physical characteristics of our sample, please refer to Appendix D (Table D.1). The mean age of infant’s at entry into the study was 282(40) days, while walk onset occurred at 369(43) days. In general, infant’s ponderal index values decreased across visits (M(SD): -4 visit=29.93(0.30) to 0 visit=25.63(3.19) kg/m^3) indicating the rate at which infants grew exceeded the rate at which they gained weight. Bayley motor subscale scores at entry ranged from 59 to 65 and at walk onset all infants had reached the maximal score of 71 because the last motor skills infants received points for were independent walking.

Stride Characteristics

Spatiotemporal. To provide an overview of the overt behavior as infants cruised, we present a description of stride characteristics and their change over time until walk onset. We used 2x4 mixed-model ANOVAs with repeated measure on Visit to assess the impact of Visit and Condition on the dependent variables: normalized step width, normalized stride length, normalized step frequency, normalized stride velocity, swing phase, double support phase, wrist width, foot rotation symmetry, segmental angular displacement, and segmental angular velocity. Appendix D (Tables D.2, D.3, D.4, and D.5) shows the gait characteristics for individual infants and the overall group mean and standard deviation by Visit and Condition from cruise onset to walk onset.

Overall, participants took longer [F(3,22.64)=7.08,p=.002] (Figure 3.5a), more frequent [F(3,22.15)=11.12, p<.001] (Figure 3.6a), and faster [F(3,19.59)=12.80, p<.001] (Figure 3.7a) strides while spending more time in swing [F(3, 24.45)=5.14, p=.01] (Figure 3.8a), and less time in double support [F(3,25.33)=2.30, p=.10] (Figure 3.8b) as they got closer to walk onset. Additionally, rotational symmetry between infant's feet showed improvements [F(3,21.00)=3.90, p=.02] (Figure 3.9) and infants positioned their hands further apart on the push cart handle [F(3,26.38)=3.49, p=.03] (Figure 3.10) with cruising experience. No main effects for Condition or interaction effects were found.

Additionally, to examine the amount of variability, we calculated the coefficients of variability (CV) for step width, stride length, step frequency, and stride velocity. We used 2x4 mixed-model ANOVAs with repeated measure on Visit to assess the impact of Visit and Condition on these variability measures. Appendix D (Tables D.2, D.3, D.4) shows the CV for gait characteristics of individual infants as well as the overall group mean and standard deviation by Visit and Condition from cruise onset to walk onset. The main effect for Condition for the CV of step width was significant [F(1,24.28)=5.41, p=.03] (Figure 3.11b) with lower average variability shown while infants cruised when wearing the diaper plus garment than in only a diaper across visits. Also, infants showed decreased variability in stride length [F(3,10.04)=3.76, p=.05] (Figure 3.5b) and stride velocity [F(3,18.89)=3.61, p=.03] (Figure 3.7b) as they neared walk onset. No interaction effects were found.

We performed a Pearson product moment correlation to examine the relation between measures more commonly associated with improved stability (e.g., trunk position, variability of stride length, and variability of step width) and two measures that we hypothesized may show associated changes: foot rotation symmetry during the middle of stance and wrist width on the push cart handle. We found a significant correlation between variability of stride length and wrist placement width (ρ=-0.293, p=.031). As variability in stride length decreased across visits, infants increased the space between their wrists when they held onto the push cart handle.

Segmental Angles

To determine the impact of Condition on body segments' movement through space, we examined timing of segmental reversals and displacements across stride cycles for trunk, thigh, shank, and foot segmental angles over time (Visits). We calculated ensemble averages with a one standard deviation envelope for segmental trajectories across the stride cycle using 3-5 'clean', continuous, alternating strides per child for 3 trials per condition for each visit (i.e., total of 9-15 strides per child per condition per visit). Figure 3.12 provides an exemplar from one infant for the segmental angle trajectories across visits between conditions. Appendix E (Figure E.1) shows segmental angles for individual participants.

Timing of segmental reversals. Overall, infants showed little impact of condition on the timing for a reversal of thigh, shank, and trunk segments when cruising. However, infants showed a trend for an earlier reversal of the foot segment during swing phase while wearing the garment at the visit preceding walk onset (-1).

Segmental angle displacement. Infants showed similar displacement trajectories for the thigh, shank, and trunk segments when cruising while wearing the garment compared to diaper. At initial visits, the foot segment showed relatively small displacement in swing, but this displacement became more pronounced by walk onset.

Variability in segmental angles across the gait cycle. In general, infants showed similar amounts of variability in limb segment trajectories between conditions. Trends were shown for increased segment variability during swing and end stance for the foot and shank segments as well as initial to mid stance for the thigh. These trends became more apparent with increased cruising experience. Decreased variability was found during midstance for foot and shank segments as well as end stance and initial swing for the thigh across conditions.

Phase Portraits

Before addressing phase portrait plots, we describe basic changes in mean segmental angular velocities. Only the main effect for Visit was significant for thigh [F(3,23.73)=6.21, p=.003] and shank [F(3,24.13)=13.23, p<.001], showing increases in angular velocity with cruising experience. However, to examine the dynamic relation between segmental angular velocity and segment position, we created phase plane portraits for each lower limb segment (e.g., thigh, shank, foot) across three stride cycles as infants cruised (Appendix F, Figure F.1). Figure 3.13 provides an exemplar from one infant for the phase plane portrait plots across visits between conditions. In general, infants showed greater consistency and constraint of the phase portrait plots with fewer inflections and loops for the thigh, shank, and foot while cruising in a diaper plus LG than in only a diaper across visits. A shift in state space was seen in 6 of 9 infants tested for the shank and foot segments at various visits; shift in the thigh segment was rare while cruising in a diaper plus LG. Also, those infants who took longer to reach walk onset showed less consistency in shape, timing of gait events, and increased presence of inflections and loops across visits than infants who progressed, relatively, rapidly from cruising to independent walking. Additionally, the thigh segment appeared to stabilize its' shape earlier than either the shank or foot segments.

Thigh

Diaper. In general, the shape of the thigh segment had vertical sides, flat bottom (stance), and round top (swing). This shape changed slightly as infants gained cruising experience.

During early visits, overall shape for the thigh segment trajectory during cruising had vertical sides, flat bottom (stance), and round top (swing). At the beginning of swing, toe off tended to occur close to zero velocity for the thigh segment, near the base of the left vertical side, as infants were attempting to provide focused, ballistic control to lift their lower limb up. In the middle of swing, trajectories became rounded as infants took advantage of pendular forces to advance the segment. Close to the end of swing, infants showed an increase in forcing or ballistic control (e.g., relatively vertical right side of figure) to bring their thigh segment down for touch down. Throughout stance, shape is relatively flattened, but with a considerable number of inflections and loops, indicating attempts to actively control the thigh segment's extension, but with limited control. From the end of stance to toe off, we see relatively vertical lines (left side of figure), suggesting increased active forcing or ballistic control at the thigh segment.

At walk onset, swing, or the initiation of swing, showed more rounded trajectories as infants began to take advantage of pendular forces to advance the segment. Near the end of swing, infants showed increased occurrence of focused, ballistic control (e.g., relatively vertical right side of figure) to bring their thigh segment down. At the beginning of stance, shape of the thigh segment phase portrait plot is flattened, suggesting active control of the thigh segment's extension. From the middle of stance to toe off, we again see relatively vertical lines (left side of figure), but with increased rounding as infants approached toe off, suggesting initial application of active forcing or ballistic control that rapidly transformed into pendular forces at the thigh segment.

Lycra garment. While wearing the garment, compared to the diaper-only condition, infants showed a general increase in consistency and convexity during swing for the trajectory shape of their thigh segments. The increase in convexity during swing indicates that infants applied more force over time to move the thigh segment during swing while wearing the LG compared to only a diaper. In general, during stance phase, infants showed decreased incidence of inflections and loops as well as a shift in position of the thigh segment across visits while wearing the LG compared to only a diaper. Most noticeably, the LG tended to constrain infant's thigh segment motion and cause a compared to when cruising in only their diaper.

Shank

Diaper. Shank segment trajectories from early visits showed vertical sides, flat bottom (stance) and round top (swing), similar to the thigh segment trajectory. At the initiation of swing, toe off tended to occur near zero velocity for the shank segment (near base of left vertical side) as infants provided focused, ballistic control to lift their foot from the floor. In mid swing, shank segment trajectories became rounded as infants took advantage of pendular forces to advance the segment. We saw the occurrence of inflections during swing more frequently for the shank segment compared to the thigh or foot during the earliest visits. During mid to end swing, infants showed an increase in forcing or ballistic control (e.g., relatively vertical right side of figure) as they extended their shank segment for touch down. Throughout stance, shape is relatively flattened, but with many inflections and loops, indicating attempts to actively control the shank segment's extension, but with limited control. From the end of stance to toe off, we see relatively vertical lines (left side of figure), suggesting increased active forcing or ballistic control at the shank segment.

By walk onset, swing, or initiation of swing, continued to show round trajectories, but stance also showed more round trajectories as infants learned to take advantage of pendular forces to advance the shank segment. At touchdown for stance, infants showed an inflection (sometimes a loop) at the beginning of stance, suggesting either an abrupt reduction with immediate resumption or increase in movement velocity or reversal of the shank's segment. Infants showed more consistency in the shape, as evidenced by less overlap, and location of event (e.g., toe off, touchdown) occurrences within the state space when cruising in their diaper at walk onset.

Lycra garment. During early visits while wearing a diaper plus LG, infants showed less consistency in the shank segment trajectory's shape with 1-2 cycles typically showing a relatively square shape while the subsequent 1-2 cycles became round.

Additionally, infants showed less consistency in the shank segment trajectory's location of event occurrences within the state space. Noticeably, at earlier visits, infants showed increased occurrence of inflections and loops while wearing a diaper plus LG compared to only a diaper, primarily during stance phase.

At walk onset, infants tended to show more consistency in shape (more overlap) between shank segment trajectories while wearing a diaper plus LG compared to only a diaper. During end stance to the initiation of swing at toe off and end of swing to touchdown for stance, infants showed increased incidence of vertical angular velocity displacement for the shank segment while wearing a diaper plus LG compared to diaper-only. Shape in mid swing and stance remained round, suggesting use of pendular forces to advance the shank segment. Overall, diaper plus LG appeared to contribute to a shift in position within the state space for the shank segment compared to cruising in only a diaper.

Foot

Diaper. During early visits, while cruising in only a diaper, overall shape for the foot segment trajectory had a square shape with a relatively flat bottom (stance) and top (swing) with vertical sides. At the beginning of swing, toe off tended to occur near the top of the left vertical side, suggesting swing was initiated via ballistic control. In swing, foot segment trajectories flattened, indicating active control of the foot segment's flexion. In end swing, infants again showed an increase in forcing or ballistic control to bring their flexed foot segment down for touchdown. During stance, shape is relatively flattened, but with inflections and loops, indicating attempts to actively control the foot segment's extension, but with limited control. Infants tended to show less consistency in shape (less overlap) and location of event occurrences in state space at earlier visits between foot segment trajectories while wearing only a diaper.

By walk onset, toe off tended to occur near the middle of a rounded trajectory as infants took advantage of pendular forces to advance the foot segment. Near the end of swing, infants showed a rapid transition from the round trajectory to a vertical segment, suggesting active control of the foot segment's flexion that abruptly terminated in an inflection (sometimes a loop) at touchdown. This sudden change in shape at touchdown indicates a sudden reduction (or reversal) and then resumption in the foot segment's movement velocity. Following touchdown, shape in stance showed relatively flattened foot segment trajectories as infants actively controlled foot extension in preparation for toe off. Overall, infants showed more consistency in foot segment trajectory shape (more overlap) and location of event (e.g., toe off, touchdown) occurrences within the state space when cruising in their diaper at walk onset.

Lycra garment. During early visits while wearing a diaper plus LG compared to diaper only, infants showed less consistency in the foot segment trajectory's shape (less overlap) and increased occurrence of inflections and loops, primarily during stance.

However, at walk onset infants showed more consistency in shape (more overlap) between foot segment trajectories while wearing a diaper plus LG compared to only a diaper. In general, the shape and trajectory of the foot segment was similar to the diaper-only condition, but the plot itself tended to shift location within state space.

Segmental Angle-Angle Plots

To examine how the coordination between adjacent lower limb segments unfolded throughout the stride cycles, we created segmental angle-angle plots between the thigh-shank, shank-foot, and thigh-foot for three stride cycles as infants cruised for each visit and each condition (Appendix G, Figure G.1). Figure 3.14 provides an exemplar from one infant for the segmental angle angle-angle plots across visits between conditions.

Diaper. The thigh-shank, shank-foot, and thigh-foot segmental angle-angle plots show a generally positive diagonal slope, albeit with less consistency (less overlap) at earlier visits. For the thigh-shank segment, shape was relatively round with a slight positive diagonal slope and rounded reversals indicating decoupling between segments. However, for the shank-foot segmental angle-angle plots, shape was more consistent (more overlap) with positive diagonal slope indicating in-phase coordination and slightly angled reversals suggesting more coupling between the segments. In the thigh-foot segment, shape was generally in a positive diagonal slope with broad round reversals near the bottom of the slope and more angled reversals near the top of the slope with a flat top. The round reversals indicate a decoupling between the thigh and foot segments succeeded by maintenance of the foot relatively constant while changing position of the thigh segment (flat top) followed finally by more coupling between the thigh and foot segments (angled reversal) before become decoupled again.

At walk onset, shape for all segmental angle-angle plots of the lower limb segments are more congruent (more overlap), indicating an increase in consistency between segments and similar changes in displacement for the segmental angles across the gait cycles. During the gait cycle, some infants showed rounded trajectories at reversals for the thigh-shank segments, indicating a phase offset with decoupled coordination between the thigh and shank segments. Presence of this decoupling between the thigh and shank segments became more apparent by the last visit at walk onset. The shank-foot segment also showed rounded reversals, indicating a decoupling between segments at earlier visits, but this shifted toward more tightly angled reversals, suggesting more coupling as infants approached walk onset. However, the thigh-foot segments maintained a similar shape to early visits, but with more consistency (more overlap). Thus, as the thigh and shank segments became more differentiated, the shank and foot segments were moved more closely together as infants gained cruising experience.

Lycra garment. When infants wore a diaper plus LG, they maintained the positive diagonal slope of the thigh-shank and shank-foot pairs seen for the diaper only condition. However, the thigh-shank segmental angle-angle plots show increased consistency of trajectory and rounded shape, suggesting a decoupling between segments. Generally, at earlier visits, the shape and slope of the plots were more consistent between strides in comparison to the diaper-only condition. In the thigh-shank and thigh-foot plots, thigh range of motion decreased, suggesting increased constraint of the thigh segment across the gait cycle. The shank-foot plots show occurrence of visible vertical segments during which the shank was held more rigidly constant while the foot segment moved in the diaper plus LG condition.

Some infants (e.g., LH, LB, KM) showed more incidence of rounded trajectories between the thigh-shank segments at various visits while wearing the LG compared to the diaper-only condition. Commonly, the prevalence of rounded trajectories at reversal for the shank-foot decreased, but increased for the thigh-foot segments as infants approached walk onset while wearing the garment compared to only their diaper. Additionally, most infants showed a shift in state space for the intersegmental coordination behavior within the shank-foot pairing while wearing a diaper plus LG when cruising compared to only a diaper.

Anchoring Indices

To investigate the coordination among the head, shoulders, and trunk in infants while learning to cruise and subsequently walk independently, we calculated the anchoring index values (Assaiante, Thomachot, Aurenty, and Amblard, 1998) in the frontal plane and sagittal planes (Figure 3.15a,b,c,d, respectively; anchoring index values for each individual per visit per condition are presented in Appendix D (Table D.6)). During cruising, none of the anchoring indices were significantly negative (<0) for the segments we measured, regardless of cruising experience, between Visits or Conditions. However, note that the shoulder segment was maintained in a relatively stable position while infants cruised due to positioning of their hands on the push cart handle for support; thus, results for the shoulders and trunk must be viewed cautiously.

Trunk Anchoring Index in the frontal plane. The trunk anchoring index (Figure 3.15c) was always positive and showed a trend for significance for Visit [F(3,20.80)=2.82, p=.06], but no effect for Condition, indicating hip stabilization in space. This finding suggests that hip stabilization in space while cruising may be learned early. The trunk anchoring index displayed a gradual increase from infant’s first visit to walk onset.

Head and Shoulder Anchoring Indices in the frontal plane. The head (Figure 3.15a) and shoulder (Figure 3.15b) anchoring indices were not significant, remaining near zero, at any visit or between conditions, indicating no preference for stabilization of the head or shoulders in space or relative to the inferior segment (e.g., shoulders and trunk, respectively) appeared while infants learned to cruise with or without the LG.

Head Anchoring Index in the sagittal plane. We did not find a significant main effect for Visit or Condition or interaction when we examined the head anchoring index (Figure 3.15d) in the sagittal plane (pitch). This finding indicates no preference for stabilization of the head in the sagittal plane was detected while infants were learning to cruise pushing a cart.

Discussion

While infants are learning to cruise, they are developing and refining the control which will enable them, ultimately, to walk independently. While cruising, infants learn to harness a multitude of complex, critical components including strength, coordination, and motor control. However, this process is far from easy and requires repeated practice. Our goal was to examine how a flexible assistive device worn around the hips and pelvis, LG, sometimes used in physical therapy to assist children who have motor control disabilities interfering with gait, affects the cruising patterns of typically developing infants.

In our first hypothesis, we proposed that wearing a LG may cause changes in overt cruising behaviors such as step width, stride length, step frequency, and stride velocity. Overall, our results showed that the LG condition was not significantly different from the diaper-only (control) condition. For the classic gait parameters used to reflect control, infants showed similar step widths, stride lengths, and spent almost identical amounts of time in swing and double support. However, the LG condition did produce a slight decrease in cruising speed. Similarities were also seen upon examination of the coefficient of variability (CV) values for stride length, step frequency, stride velocity, but not step width. The CV for step width was lower across visits while infants wore a diaper plus LG than only a diaper while cruising. Infants also showed improved symmetry for rotation of their feet at mid-stance while wearing the LG (primarily visits -2 and -1), but was not significant due to high variability. This finding may also suggest some increase in control of their lower limbs when cruising with the garment. These findings for step width variability and foot rotation symmetry are very interesting considering that the LG is applied to the pelvis and hips, proximal to the foot segments where these changes are identified. One possible reason for the changes observed at the foot level while infants cruised in the LG with strapping may have been an augmentation of hip abduction/adduction control similar to the findings of Maguire and colleagues (2009) in adults post stroke who wore a LG with strapping while walking. Reinforcement for hip abduction/adduction control by the LG with strapping may have enabled infants to take strides that were not only more evenly spaced, but with feet more symmetrically positioned during midstance. The similarity in findings for gait characteristics between this study and Maguire et al.'s (2009) may be due in large part to the use of identical strapping techniques. The actual LG provided a base to which the straps were anchored as well as a smooth, comfortable skin-garment interface that had a slightly adhesive quality, helping to prevent slippage of the garment on participant's skin. Additionally, the

LG may have afforded a minimal amount of structural positioning for participant's pelvis and thighs. However, the manner in which the straps were applied with participant's legs in a neutral/semi-neutral (dependent on participant cooperation while we donned the LG and strapping) position appeared to provide a considerable amount of segment stabilization and thus may have effectively afforded more neutral positioning of the shank and foot segments due to their association with the thigh. Thus, this interrelationship between the LG and strapping is a confounding variable for our results, and thus, they must be interpreted cautiously. Future research should test the impact of the LG when worn alone (without strapping) to determine if it has the same effect, no effect, or a partial effect on gait.

Segmental angle ensemble averages show that, overall, infant's lower limb segment displacement trajectories, timing of segmental reversals, and the amount of variability in segmental displacement across the gait cycle was similar between conditions. With cruising experience, range of motion for the shank segment significantly increased, but trunk significantly decreased, suggesting increased ability to maintain a stable trunk and allow the shank to contribute to leg movement through space. When infants began to cruise, similar to when infants begin to walk independently, they constrained the lower leg (shank) in a more extended position, seeking stability for the system as it performed this new, highly unsteady skill. With experience, infants were able to explore the degrees of freedom afforded by movement of the shank segment during cruising. The trunk showed less control at earlier visits (more wobble), similar to the results of Haehl et al. (2000). Trunk control increased, as evidenced by a decrease in range of motion, perhaps required for the onset of independent gait. At earlier visits, infants were assured that they would not lose trunk control because of their grasp on the cart handle; thus, they were able to explore and refine trunk control during cruising. With cruising experience, infants learned to decrease movement of their trunk while increasing their efficient forward movement with fewer falls. Thus, by walk onset, infants had become efficient and skillful cruisers by increasing their use of the shank's range of motion and decreasing the amount of movement excursion exhibited by the trunk, allowing infants to cruise faster with longer, more frequent strides, spending more time in swing and less in double limb support.

While wearing a diaper plus LG, infants tended to show, overall, more consistency for the thigh and foot segments, but a shift in position within the phase portrait state space for the shank and foot trajectories relative to the diaper-only condition, indicating that the system was perturbed, causing adaptation of the underlying coordination pattern for cruising. While the classic overt gait parameters did not show any significant changes between conditions, this may suggest that the changes caused by short-term LG use in typically developing infants were too subtle to be detected by gross measures. However, most noticeable from examination of the dynamic behavior of leg segments, five of nine infants showed improvements in control with fewer "corrections" in their position and/or movement velocity of the thigh and foot segments while cruising in a diaper plus LG. Thus, the dynamic representation of cruising behavior was more reflective of overall impact than isolated parameters (e.g., phase portrait plot shift in location in state space).

Infant's cruising patterns while wearing the LG were more consistent in shape, trajectory, and slope for the segmental angle-angle plots as infants neared walk onset.

While infants tended to show constraint of the thigh segment in the thigh-shank and thigh-foot plots while wearing the LG, we also found that the shank segment was more constrained, with greater coupling relative to the foot segment in the shank-foot plots, primarily as infants neared walk onset. The finding of increased thigh constraint makes sense considering where the garment is positioned, but the change in relative relationships seen between the shank and foot segments may be considered somewhat surprising considering the LG's position surrounding the pelvis and hips. However, previous research studies with younger infants have shown adaptations in trunk movement when weights are attached at the wrists (Rochat et al., 1999) as well as step quality and quantity when wearing different amounts of clothing on the lower body (Groenen, Kruijesen, Mulvey, & Ulrich, 2010). Additionally, application of a load to infants (torso, ankles, wrists) has been shown to cause changes in posture and center of mass position, resulting in subsequent changes in walking stability and pattern (Adolph & Avolio, 2000; Vereijken et al., 1999). Thus, our results are aligned with others showing infants have the capacity and flexibility to adapt their motor control strategies, segment coordination, and movement patterns to external manipulations of body and/or limb masses.

We also examined anchoring indices for further insight into intersegmental coordination between the head, shoulder, and trunk as infants cruised with and without the garment. Overall, anchoring indices values across visits were similar between conditions. Infants showed a gradual increase in frontal plane trunk stabilization (roll) in space across visits, possibly suggesting that infants learned to stabilize their trunks relatively early when learning to cruise similar to the findings of Assaiante and colleagues (1993, 1998) for toddlers first learning to walk independently. However, unlike Assaiante et al.’s results, the shoulder AI for our infants in the frontal plane (roll) while cruising remained close to zero across visits and did not increase, possibly because of the positioning of their hands on the push cart handle. Comparable results were also shown for the head anchoring indices in the frontal plane (roll) and trunk AI in the sagittal plane (pitch) across visits, indicating that infants did not show a preference for stabilization of the shoulders or head in frontal plane space or trunk in sagittal plane space relative to another segment. Importantly, we must be cautious in our interpretation of these results for the anchoring indices for the shoulders and trunk because the shoulders (and by relationship, the trunk) was maintained in a relatively unchanging position while infants cruised due to positioning of their hands on the push cart handle for necessary support.

Because we were able to longitudinally test infants while learning to cruise, and subsequently walk independently, we hypothesized that infants would show increased incidence of changes in their cruising patterns at earlier, rather than later, visits. However, because infants showed a wide range of strategies for performance of cruising, we were unable to identify a specific visit before walk onset during which infants cruising behaviors were more overtly influenced by wearing the LG. Despite these limitations, we think there may be some trends within the data that provide hints. Infants tended to show the most consistency for shape, trajectory, and timing of gait events in the phase portrait plots at the -1 visit (visit preceding walk onset) while infants wore the garment. Additionally, infants showed a significant difference by Condition for the CV of step width and improvements in symmetry of foot rotation that were most apparent at visits -2 and -1. Thus, wearing the LG when infants exhibit cruising patterns relatively consistent with what was shown at the -2 and -1 visits may facilitate shifting of the system.

Lastly, we had hypothesized that the LG would afford infants with the most cruising experience who were preparing to begin walking independently enough pelvic stability to take some independent steps. However, only one infant, LH, was able to do this during the visit prior to walk onset. And, per parental report, she also began to take alternating, independent steps two days following testing. Thus, we cannot conclude, based on one infant, that the LG facilitated the performance of independent walking.

Overall, TD infants showed fewer changes than expected in their cruising patterns while wearing a LG. We may have found fewer changes than expected due to a few factors. First, our participants were typically developing infants who were developing adequate lower limb segmental control. Therefore, the LG condition was not robust enough to cause adequate perturbation to the system to influence the overt gait parameters for these infants. Also, fit of the LG may have needed to exert more tension around the pelvis and/or thighs as infants neared walk onset due to increased strength in their legs and trunk. However, it is unclear if the gait parameters that did show significant change will be beneficial or detrimental for facilitating segmental control and coordination in individuals with disabilities. We contend that further testing in populations with motor control disabilities, for whom the LG is designed and marketed, is essential to determining if these flexible external devices are of benefit or hindrance to the development of functional movement patterns.

Limitations

The primary limitation to our study is the small sample size. However, we tested infants longitudinally, providing strength to the results obtained by value of 18 test sessions with cruising infants (plus 9 test sessions at walk onset). The small sample size was exacerbated by variability in the number of months of cruising prior to walk onset (e.g., 2 infants only cruised for one month while others cruised for 5 months). The small sample size and its' associated variability may have contributed to the lack of significance found between Conditions. While we expected to find a difference in range of motion for the lower limb segments while infants cruised in the LG compared to only their diaper, our results did not reach significance at the $p \leq .05$ level. However, sample size calculations with power set at 0.60 for thigh and shank displacement range of motion indicate that 32 TD infants learning to cruise and tested longitudinally will be necessary in future research to detect significance at the $p \leq .05$ level between Conditions.

Additionally, we were not able to truly control or measure the tension exerted by placement of the garment with strapping on our infant participants. We attempted to control the tension exerted through circumferential measurements of the thigh segment, hips, and abdominal region as well as through use of shims to spot-check 'closeness of fit'. However, these methods did not provide an accurate, objective, quantitative measure of tension. Future studies may consider using a digital tension indicator attached at one end of the strap during application and monitored throughout strap application to ensure consistency of tension throughout the application process.

Another limitation was lack of practice wearing the LG while cruising. How prolonged wear may have impacted infants' cruising patterns is unknown. Currently, manufacturers recommend that users wear LGs with strapping 6-8 hours per day, 7 days per week. We contend that this duration of wear may not be necessary or beneficial for all users based on our results showing changes in infants' dynamic movement when cruising with only short-term wear. Benefits of wear may be optimized by specific, targeted wear of these garments during repeated practice of functional skills. However, future research is necessary to determine if this recommendation will be appropriate for all or only specific populations.

In order to be able to assess cruising and walking in the same planes, we elicited forward cruising via a push cart. The aid of the cart may have masked some gait parameter changes that might have been observed with sideways cruising and cruising without the forward motion aid (cart). But, cruising, by its' definition, requires infants to hold onto a supporting surface (in this case the push cart) while walking (Haehl et al., 2000). Thus, we contend that, given the constraints of our population and research questions, our design was appropriate. Additionally, we did not measure the amount of force exerted by toddlers on the cart while cruising. Lastly, the results of this study with typically developing infants cannot be expected to hold true for infants, children, or adults with motor control disabilities. Future research using similar analyses to examine wear of the LG within these populations, for which they are designed and marketed, is warranted.

Table 3.1

Summary of Participant Characteristics

					Chronological Age (days)		
Infant ID	**Gender**	**Number of Visits**	**Ponderal Index (kg/m^3) Begin / End**	**Full Leg Length (cm) Begin / End**	**Lateral Cruise Onset**	**Forward Cruise Onset (begin testing)**	**Walk Onset**
AB	M	2	28.26 27.21	30.60 31.40	334	379	414
CW	F	4	22.48 20.51	27.90 30.80	253	303	391
CY	F	3	26.29 25.76	31.30 32.60	240	279	335
KK	F	4	25.40 25.50	30.10 31.30	269	288	367
KM	F	4	28.72 22.39	27.50 31.50	339	342	415
LB	F	3	25.85 23.77	27.60 29.10	236	250	287
LH	F	5	30.14 26.15	24.30 29.20	256	275	400
MV	M	2	30.73 31.38	28.60 29.80	282	302	334
OM	F	5	29.72 27.96	29.72 30.50	248	266	378
	OVERALL *M* (SD)	4	27.51 (2.70) 25.63 (3.19)	28.62 (2.12) 30.69 (1.16)	273 (38.64)	298 (40.09)	369 (42.98)

Lateral Cruise Onset: age when infant able to cruise laterally with consistent, alternating steps for 6 feet (lenth of a couc

Forward Cruise Onset: age when infant able to cruise forward with consistent, alternating steps for 6 feet. Testing initiat

Walk Onset: age when infant able to take 3-5 consistent, alternating steps; observed in lab.

Table 3.2

Ratio between Diaper-only condition and Diaper + Lycra Garment condition at three sites.

Visit		**Ratio between Diaper : Diaper+Garment** **Umbilical circumference (cm)** ***M* (SD)**	 **Greater Trochanter circumference (cm)** ***M* (SD)**	 **Thigh circumference (cm)** ***M* (SD)**
-4	(n=2)	0.97 (0.01)	0.92 (0.02)	0.97 (0.06)
-3	(n=5)	0.93 (0.08)	0.98 (0.03)	0.93 (0.12)
-2	(n=7)	0.97 (0.04)	0.93 (0.03)	0.97 (0.07)
-1	(n=9)	0.97 (0.04)	0.94 (0.02)	0.97 (0.05)
0	(n=9)	0.94 (0.05)	0.92 (0.04)	0.94 (0.05)
Overall				
***M* (SD)**		0.95 (0.05)	0.94 (0.03)	0.95 (0.07)

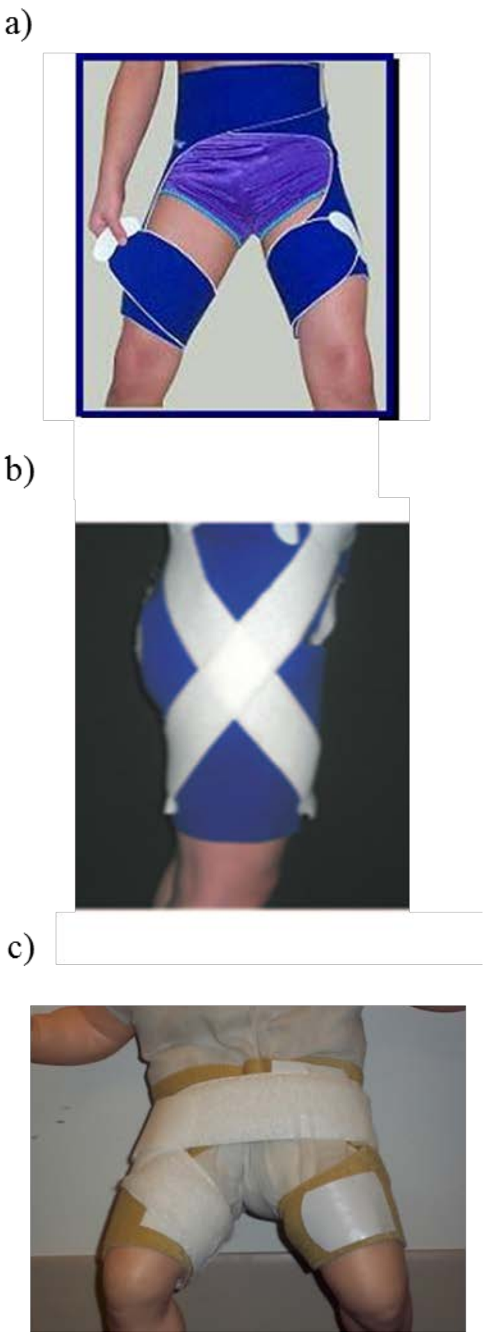


Figure 3.1a) TheraTog^{TM} hipster garment, b) TogRite Strapping, lateral view, c) TogRite Strapping on right leg, frontal view.

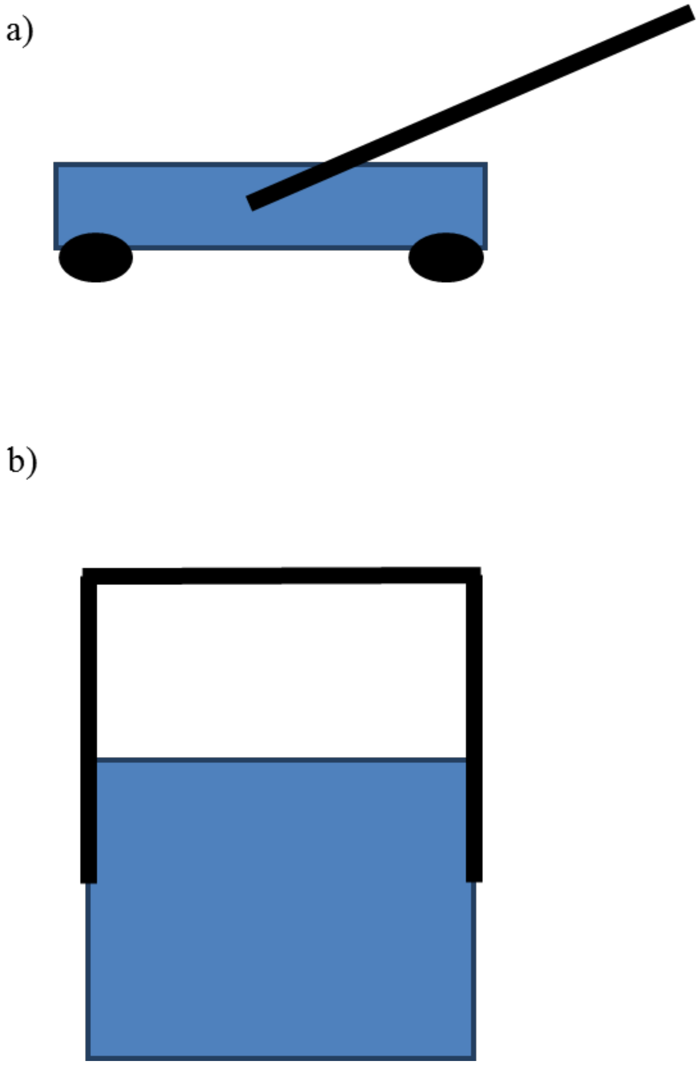


Figure 3.2a) Schematic of push cart (side view), b) Schematic of push cart (top view)

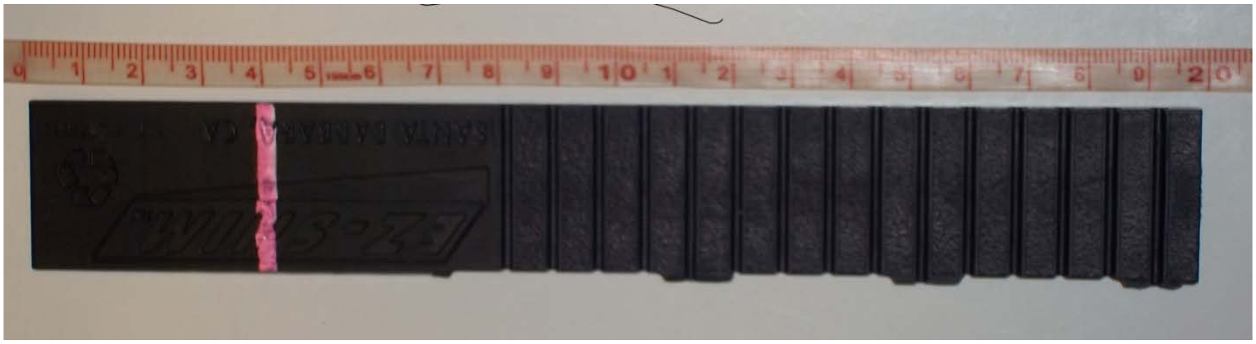
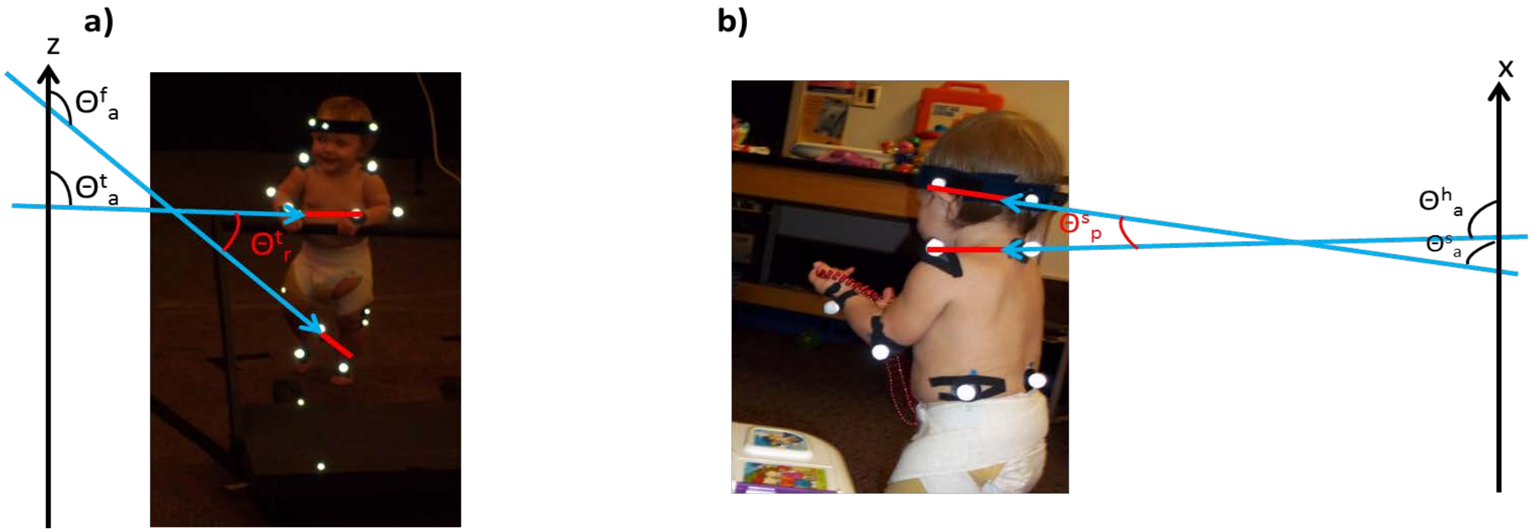


Figure 3.3. Shim used to monitor consistency of LG fit within and between infants across visits



$$\Theta^t_r = \Theta^t_a - \Theta^f_a$$

$$\Theta^s_p = \Theta^h_a - \Theta^s_a$$

$$\text{Normalized Trunk AI (roll)} = \frac{\sigma(\Theta^t_a) - \sigma(\Theta^f_a)}{\sigma(\Theta^t_a) + \sigma(\Theta^f_a)}$$

$$\text{Normalized Head AI (pitch)} = \frac{\sigma(\Theta^h_a) - \sigma(\Theta^s_a)}{\sigma(\Theta^h_a) + \sigma(\Theta^s_a)}$$

Figure 3.4a,b. Examples for calculation of Normalized Anchoring Index (AI). a) Normalized AI for roll of the trunk segment: Θ^t_a represents the trunk roll angle with respect to the external axis; Θ^t_r is the trunk roll angle with respect to the right foot axis; Θ^f_a is the right foot roll angle with respect to the external axis. b) Normalized AI for pitch of the head segment: Θ^h_a represents the head pitch angle with respect to the external axis; Θ^s_p is the head pitch angle with respect to the shoulder axis; Θ^s_a is the shoulder pitch angle with respect to the external axis. σ represents the respective standard deviations for the calculated angles.

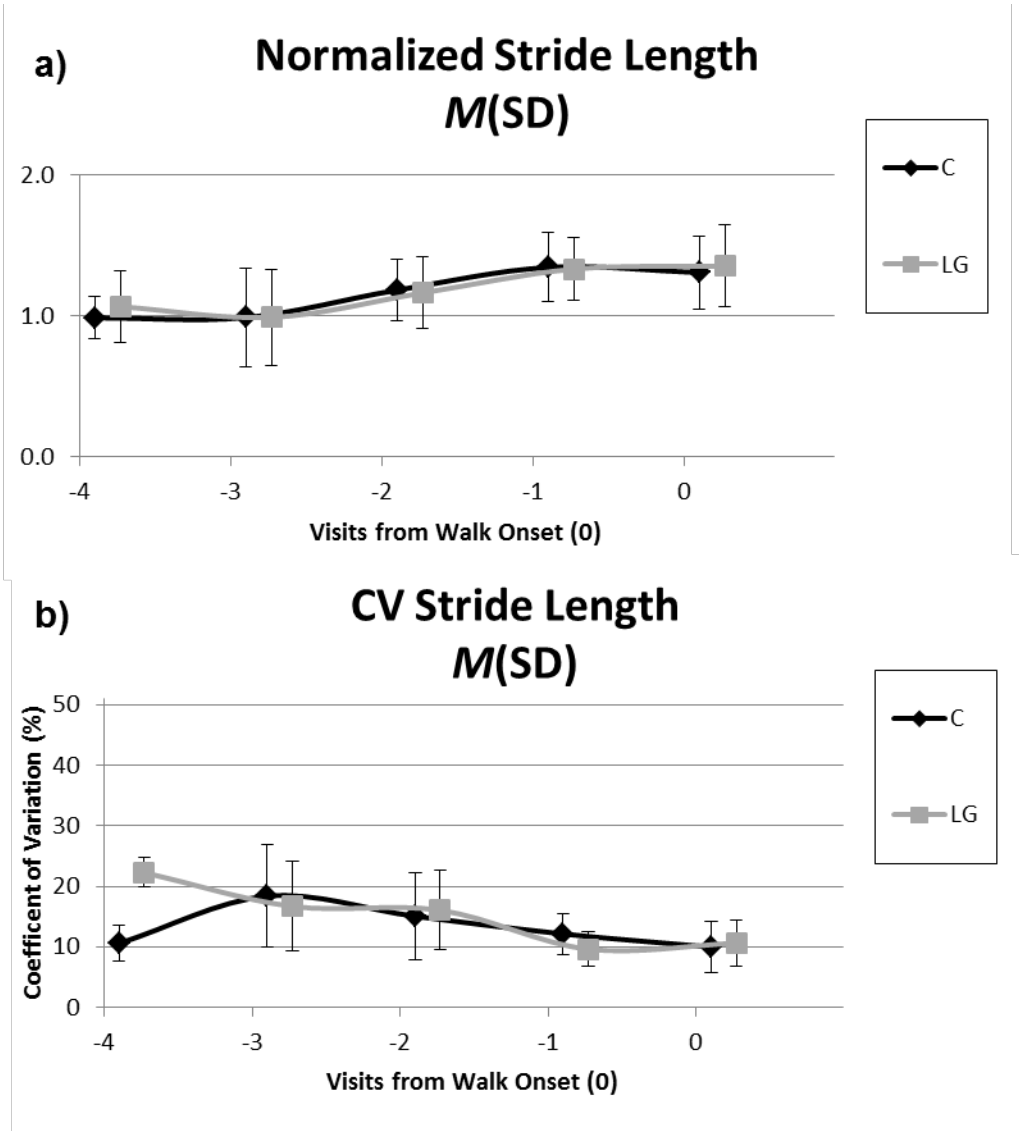


Figure 3.5a,b. a) Normalized stride length and b) coefficient of variation (CV) values for all infants across visits with ±1 standard deviation. C=Control (diaper-only condition), LG=Lycra Garment condition.

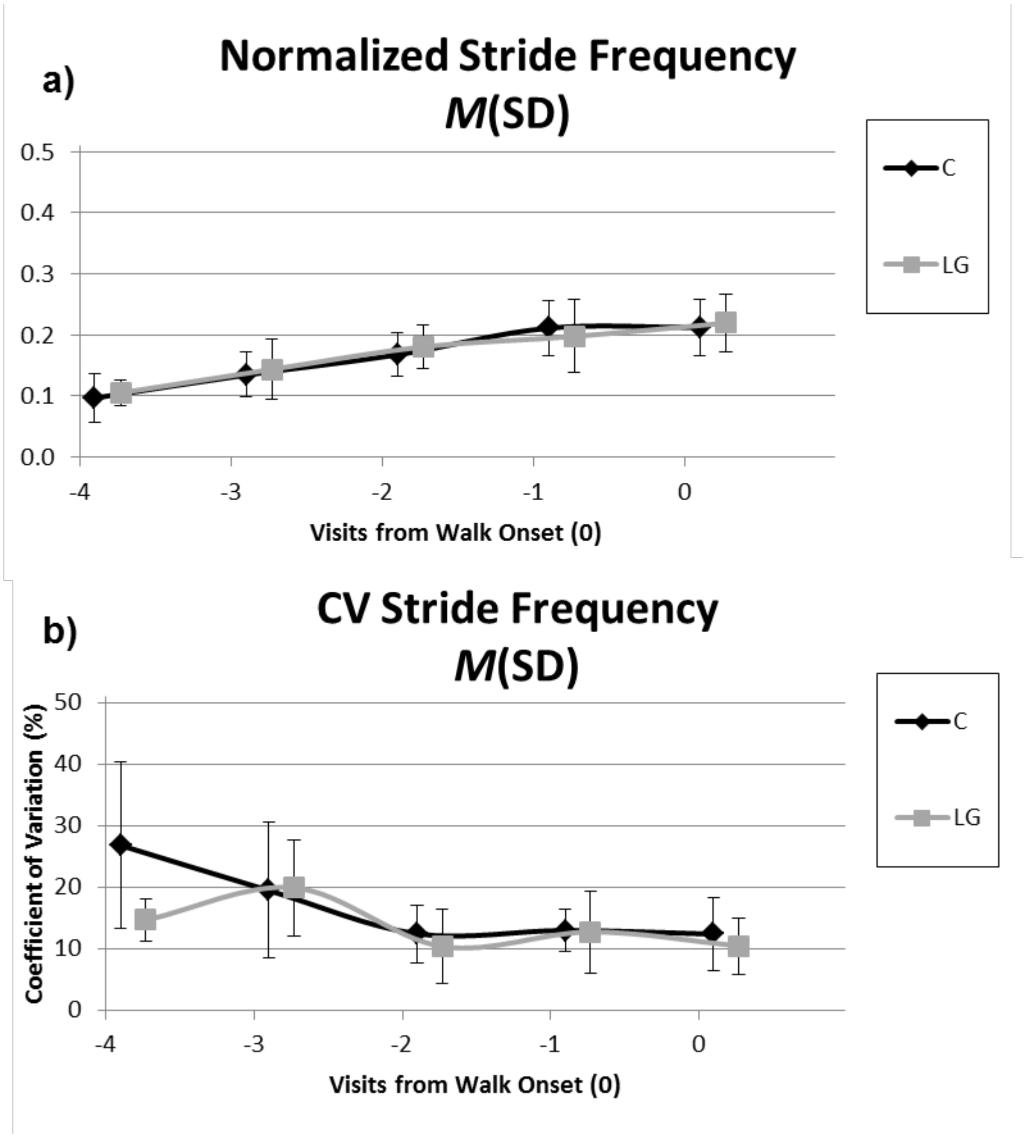


Figure 3.6a,b. a) Normalized stride frequency and b) coefficient of variation (CV) values for all infants across visits with ±1 standard deviation. C=Control (diaper-only condition), LG=Lycra Garment condition.

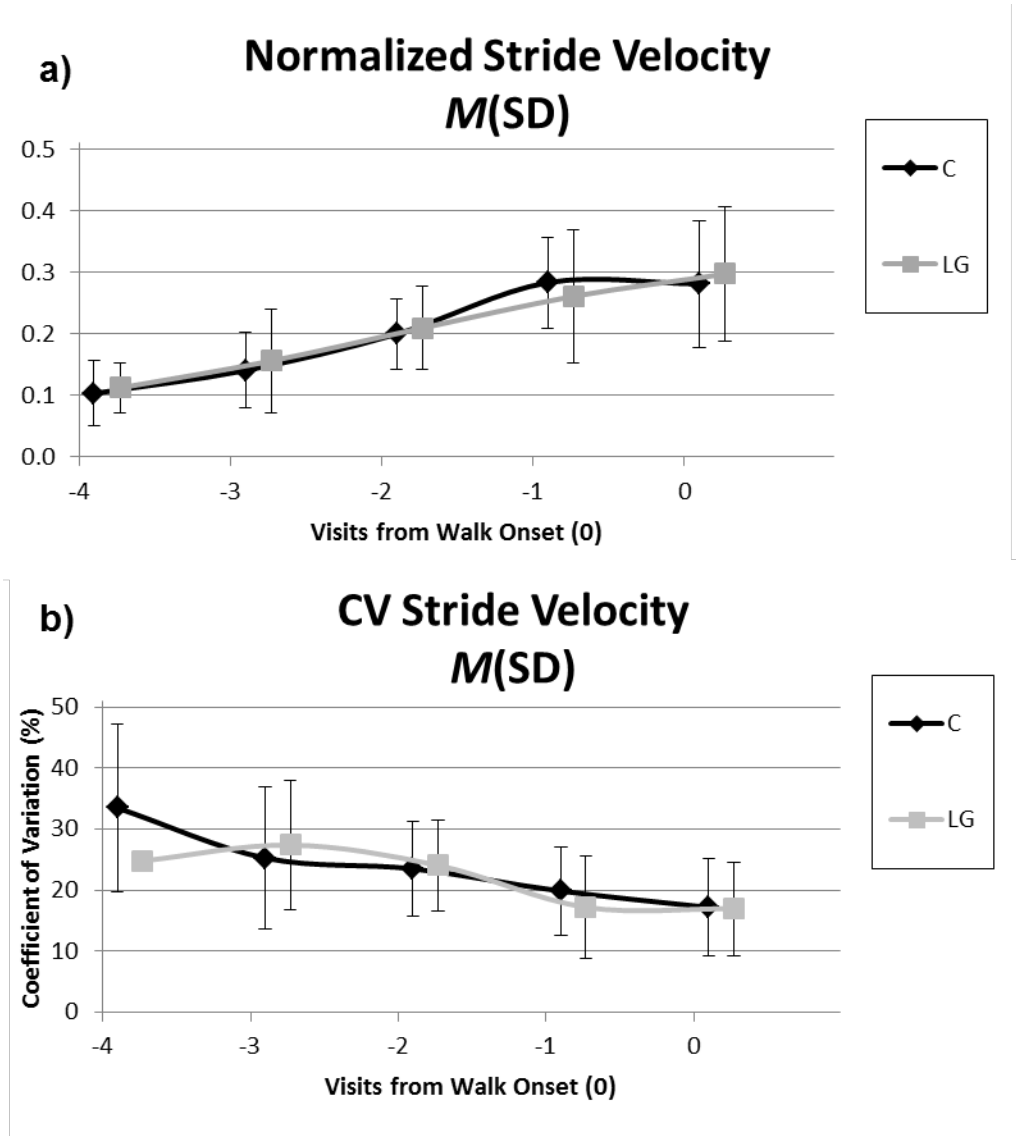


Figure 3.7a,b. a) Normalized stride velocity and b) coefficient of variation (CV) values for all infants across visits with ±1 standard deviation. C=Control (diaper-only condition), LG=Lycra Garment condition

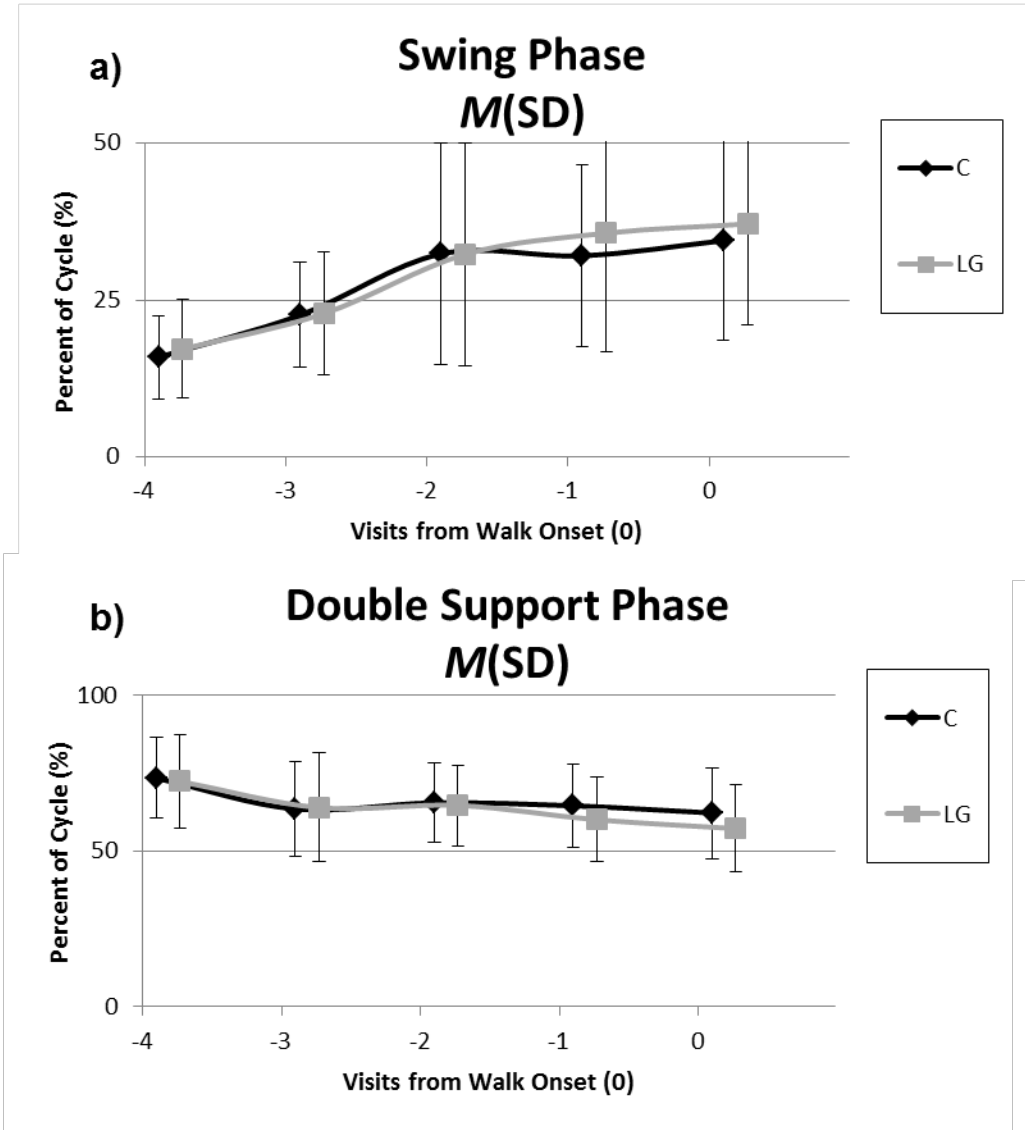


Figure 3.8a,b. a) Swing Phase and b) Double Support Phase of the gait cycle for all infants across visits with ±1 standard deviation. C=Control (diaper-only condition), LG=Lycra Garment condition

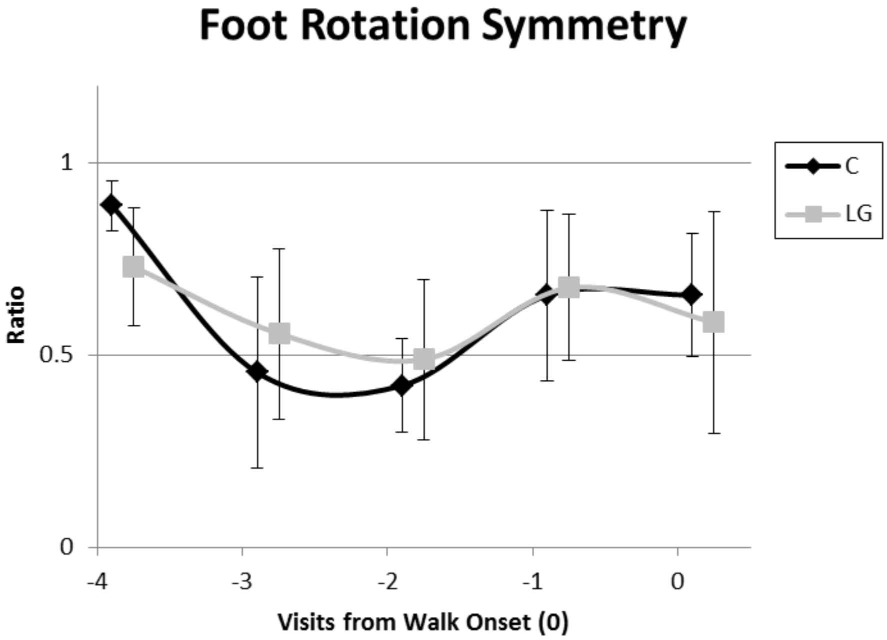


Figure 3.9. Symmetry between infant's feet while cruising across visits. A value of 1 indicates that both feet had the same angle of rotation at midstance; a 0 value indicates a 90 degree misalignment at midstance. C=Control (diaper-only) condition; LG=Lycra Garment condition.

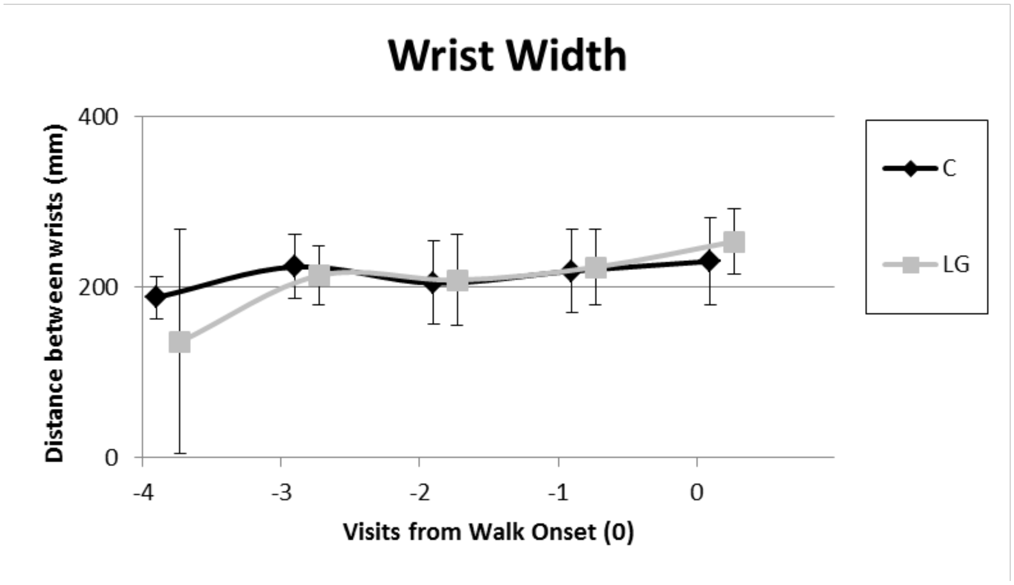


Figure 3.10. Distance between infant's wrists on the push cart handle while cruising across visits. C=Control (diaper-only) condition; LG=Lycra Garment condition.

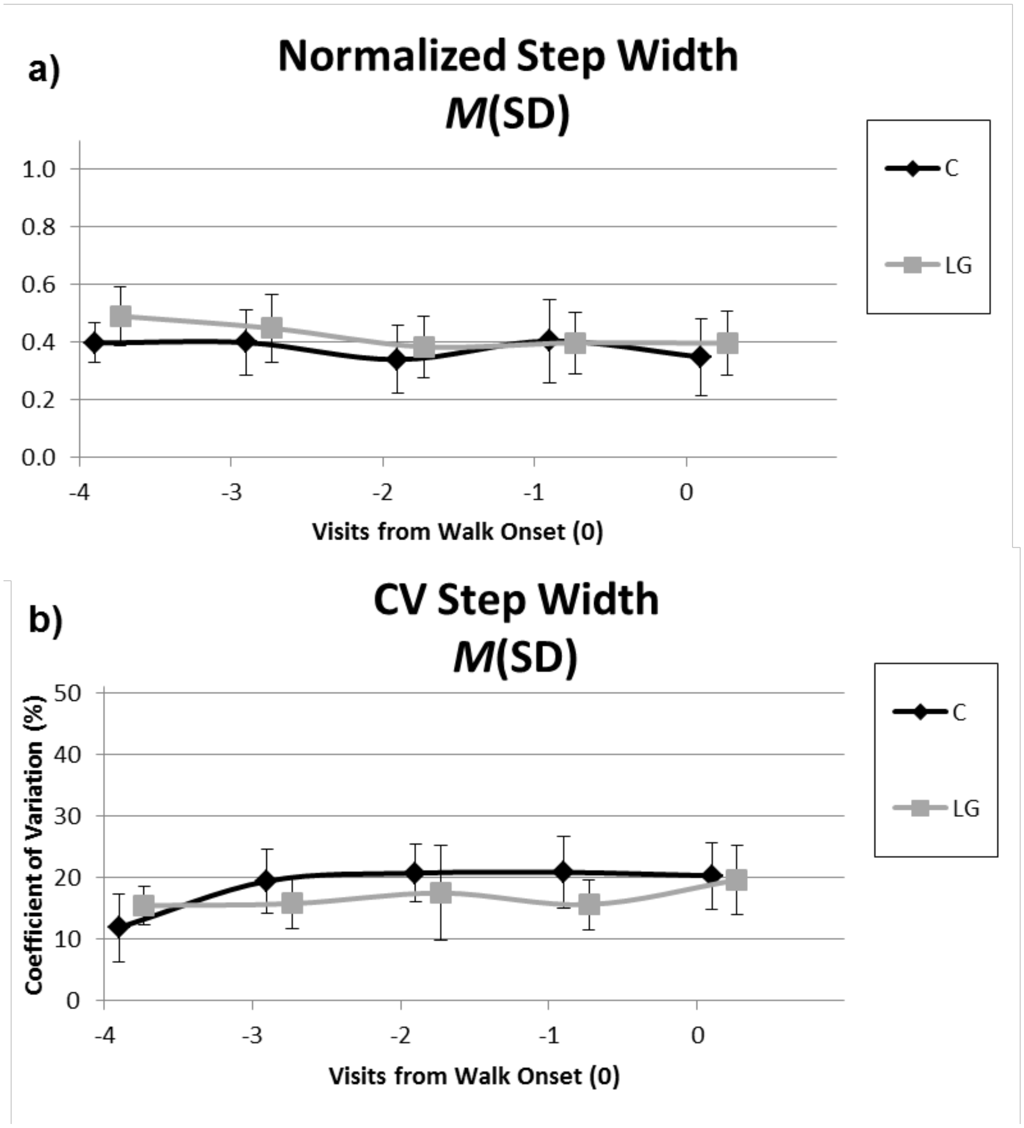


Figure 3.11a,b. a) Normalized step width and b) coefficient of variation (CV) values for all infants across visits with ±1 standard deviation. C=Control (diaper-only condition), LG=Lycra Garment condition

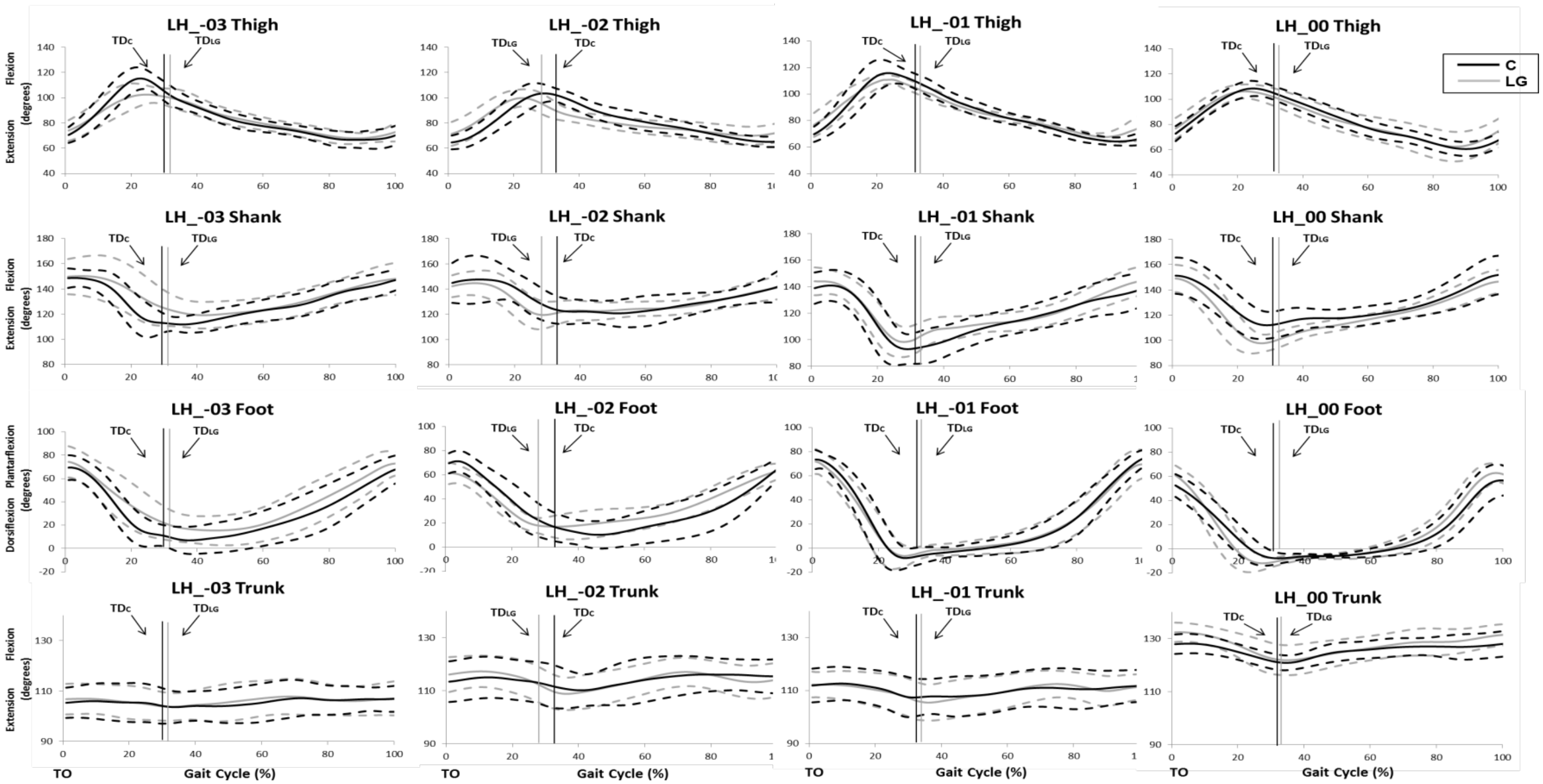


Figure 3.12a,b,c. Exemplar segmental angles for the ensemble average ±1 standard deviation envelope of the a) thigh, b) shank, c) foot, and d) trunk segments across visits for one infant. TO=Toe off; TD_C=Touchdown during the Control (diaper-only) condition; TD_{LG}=Touchdown during the Lycra Garment + diaper condition. C=Control (diaper-only), LG=Lycra Garment (plus diaper) conditions.

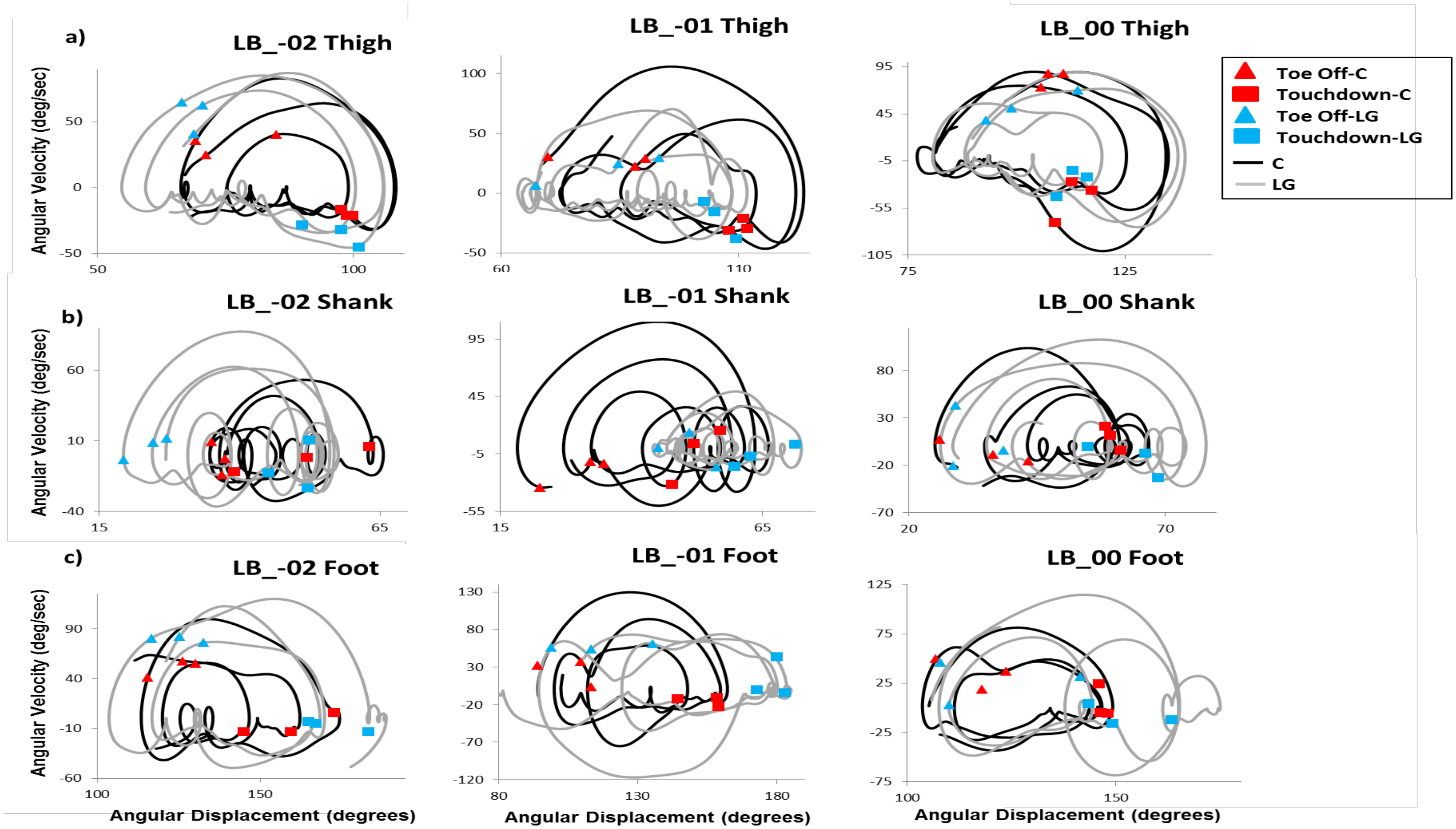


Figure 3.13a,b,c. Exemplar phase portrait plots for 3 consecutive cruising strides per condition for one infant across visits; a) thigh, b) shank, c) foot segments. Trajectories unfold in clockwise direction over the cycle duration. C=Control (diaper-only) condition; LG=Lycra Garment + diaper condition.

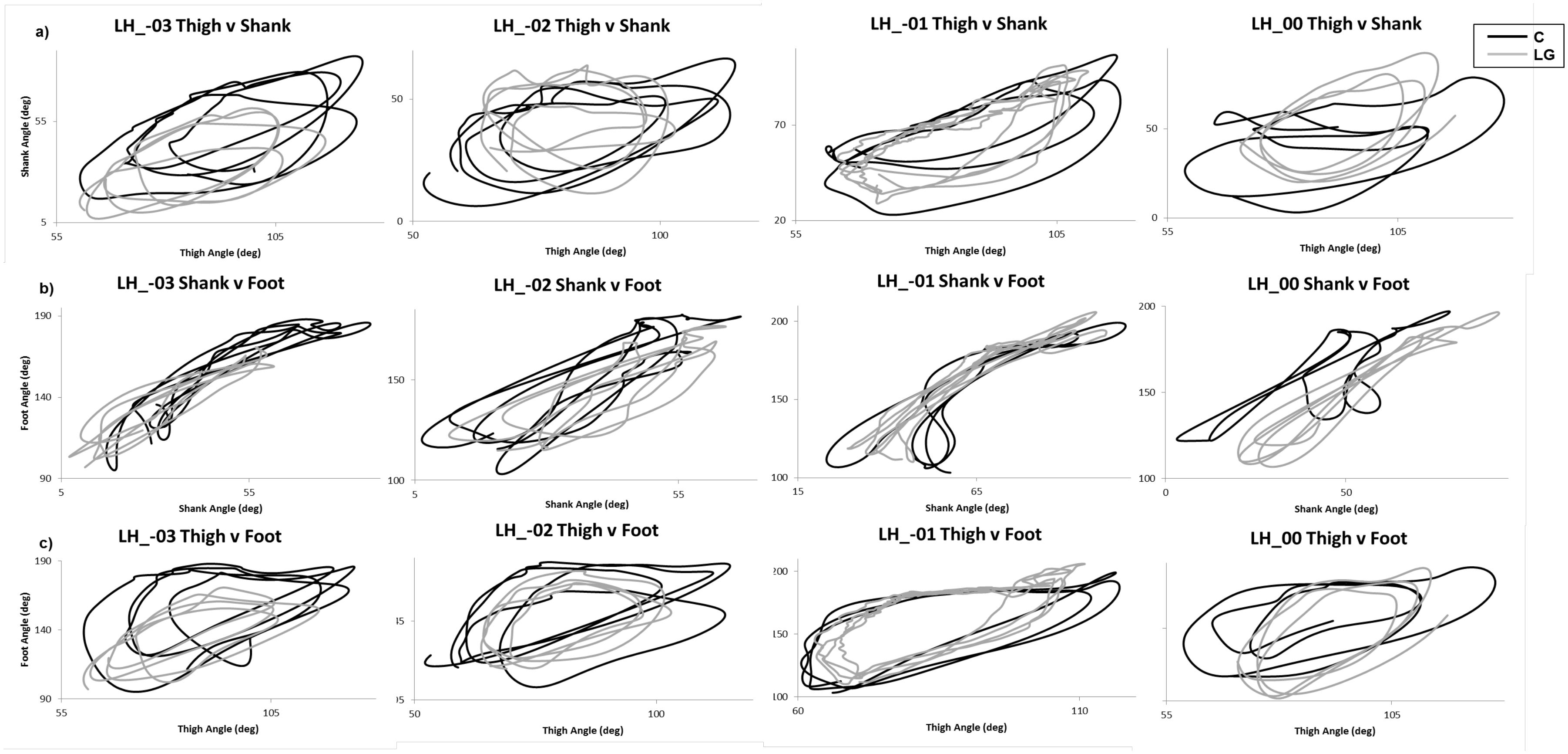


Figure 3.14a,b,c. Exemplar segmental angle-angle plots for 3 consecutive strides for one infant across visits; a) thigh, b) shank, c) foot segments. Trajectories unfold in clockwise direction over the cycle duration. C=Control (diaper-only) condition; LG=Lycra Garment + diaper condition. (This infant had -4 visits, but -4 visit data not included here because infant did not take 3 consecutive strides in the Control condition; refer to Appendix G for -4 visit plots).

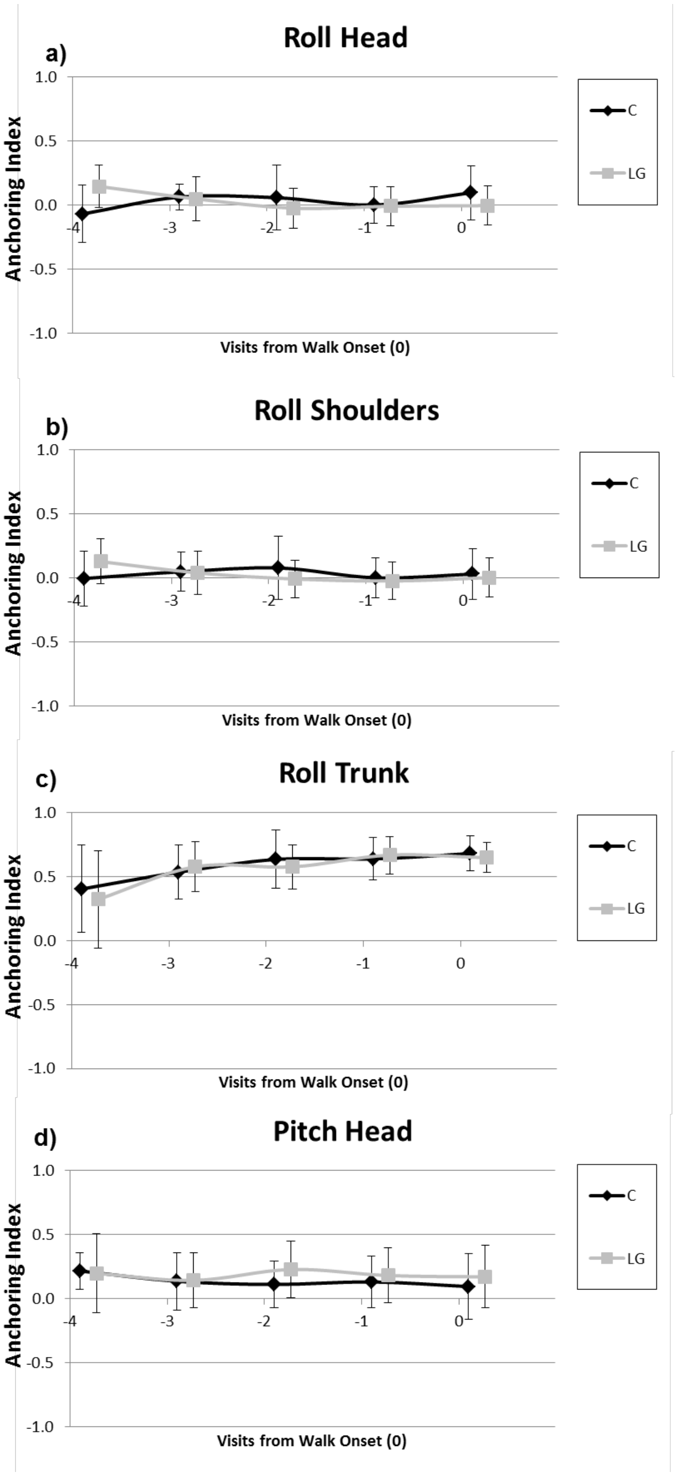


Figure 3.15a,b,c,d. Normalized Anchoring Index values for the a) roll of the head segment, b) roll of the shoulder segment, c) roll of the trunk segment, and d) pitch of the head segment for all infants across visits with ±1 standard deviation. C=Control (diaper-only condition), LG=Lycra Garment condition.

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Chapter IV CONCLUSION

General Discussion

Independent walking is a challenging and sometimes unachievable skill for many infants, children, and adults with motor control disabilities. One of the ways rehabilitation professionals have tried to provide a mechanism to aid those who are unable to acquire this skill independently is through the use of assistive devices (ADs). When using an AD, individuals show an observable change in their locomotor pattern. However, little research exists objectively documenting the acute, real-time changes in gait that occur. Understanding the acute changes in movement that occur with use of an AD is a very important component in the decision-making process for rehabilitation professionals who recommend and provide ADs to patients in the hospital or clinic to foster patient compliance and thus, safety. Our goal in this series of research studies was to quantitatively and qualitatively report both the overt and underlying changes that occurred while individuals locomoted, in real-time, while being influenced by use/application of external assistive devices.

Our results show that use of ADs caused changes at multiple levels while children with MMC and TD infants learning to cruise moved through space. Children with MMC showed changes in not only their overt gait characteristics while walking with various rigid devices, but they also showed concomitant adaptations in muscle activation patterns and energy consumption requirements. For TD infants learning to cruise, wear of a flexible garment at the pelvis and hips also resulted in some adaptations in infants gait patterns such as a slight widening in their base of support, but also in the underlying mechanical dynamics and intersegmental coordination as evidenced in phase portraits and angle-angle plots, respectively. Thus, use of these devices in very different participant populations resulted in adaptations in their gait patterns. However, what we don't know is whether these adaptations, in the long-term, would result in further improvements in independence or increased dependency on the AD during functional behaviors.

Limitations

Our primary limitations for Study 1 and 2 involving children with MMC walking with various rigid ADs were sample size and age distribution. Our criteria required children with MMC who were independent ambulators, making identification and recruitment more difficult than initially anticipated because many children with MMC are unable to walk unaided. Ideally, we would have liked to recruit more children in the C- group because these are the children who are most likely to benefit from use of a device and may transition to use of a device and/or wheelchair during adolescence or young adulthood. Additionally, the ages for our sample of participants was skewed, with all of the children in the C+ group being 5 years old, a significant confounding variable. Additionally, for Study 1 and 2, we only tested children once for each study, limiting the reliability of our results. However, for Study 1, children performed 3 trials for each condition, strengthening our results. Due to time limitations and children's tolerance levels, we were limited to only 1 trial per condition in Study 2.

Another limitation for Study 1 and 2 resulted from our participants only having a limited amount of practice walking with the various ADs prior to testing. However, all children were provided instruction by the same person (the principle investigator) and up to 15 minutes or practice walking with each AD, comparable to the amount of practice time that would occur in a clinical environment, was allotted prior to the testing sessions. Additionally, we wanted to know how acute use of these devices impacted gait, similar to when individuals are provided these devices in the clinic or hospital; we feel our results provide a first step toward greater understanding of these acute effects during device use. For Study 3 involving TD infants learning to cruise, our primary limitation was also a small sample size. However, we tested infants longitudinally, providing strength to our results. Because we required infants to be able to cruise forward before beginning to test, we may not have captured the true onset of cruising –lateral sidestepping with support. We also were unable to quantitatively control the amount of tension exerted by the straps that were applied over the LG. Lastly, we tested infants who started to cruise between the ages of 8-11 months, since this was the average age of cruising onset cited in the literature. However, not all infants begin cruising within this narrow age range. By expanding our age range, <8 and >11 months old, it may provide a greater understanding of what the ‘normal’ continuum for cruising behavior looks like before we begin to test special populations who show even greater variability in their movement patterns.

Future Studies

In all studies presented here, we conducted a lot of measurements and considered a multitude of variables within our analyses. This was necessary due both to the novel nature of our investigations involving AD use in real-time and the unique populations we tested. While our results provide a wealth of important information for research scientists and clinicians who work with these populations, we recommend that future studies should focus on measurements of trunk inclination and foot position for greater efficiency and translatability of results to the community-environment. For research scientists, use of motion capture provides the most effective means of obtaining these measures. However, it may be beneficial to consider reducing the number of retro-reflective markers placed on participants in both groups to improve the efficiency of testing sessions and subsequent data reduction.

While our results were obtained in a controlled, laboratory environment, translation to the clinic is always a difficult challenge. However, for clinicians working in time and resource-constrained clinic environments, we contend that clinicians also need to be more attentive to how individuals position their feet (e.g., step width, foot rotation, variability in foot placement during gait) relative to their trunk (e.g., trunk inclination and range of motion during gait) while using an AD. Measurements for trunk inclination and foot positioning can be readily measured in the clinic through review of video recordings of the functional skill in the sagittal and frontal planes; trunk inclination can even be quickly estimated using an application available to users of smart phones (e.g., GetMyROM by Interactive Medical Productions) and foot positioning with a pressure sensitive walkway (e.g., GAITRite mat), if available. However, these gait characteristics need to be carefully considered relative to the patient's safety requirements and modified accordingly to, first and foremost, maintain the individual’s safety during performance of functional movement.

Future studies should also examine the manner in which users of ADs employ propulsive and stabilizing forces during walking acutely and after practice. This will allow determination of how individuals who use ADs not only apply force through specific devices during functional movement, but also the manner in which they adapt their gait pattern to these changes in force distribution. With this information about propulsive and stabilizing force production, researchers and designers will then be better equipped to re-design and/or modify ADs (e.g., rocker bottoms on walking poles) to facilitate force distribution and forward momentum for users. Additionally, by following users from the time they first begin use of an AD with follow-up testing after 6, then 12 weeks of use, we will gain a greater understanding of the adaptations that occur with well-practiced use and facilitate clinician determination for advancement to more challenging devices as appropriate to specific needs and functional goals for individual users.

Future studies investigating infants learning to cruise while wearing an LG with strapping should focus on recruitment of TD infants who begin cruising early (<8 months old) and those who begin late (>11 months old) because we are interested in eventually testing the impact of these flexible support garments in infants with motor control disabilities. By testing TD infants at each end of the average age range for cruising, we will gain a greater understanding of what ‘normal’ looks like for the skill of cruising. Thus, we will be better prepared to test infants with motor control disorders who have more variability inherent to their performance of motor behaviors.

We also contend that future research testing the effects of wearing an LG while infants cruise should measure both muscle activity and dynamic force production. This valuable information will provide insight into the underlying motor control strategies to perform the functional movements involved in cruising and how the LG with strapping impacts both muscle recruitment and force production.

Because LG's with strapping are designed and marketed for special populations, another important direction for future research will be to test the effects of both acute and long-term use in infants with motor control disabilities (e.g., Down Syndrome) while learning to cruise. Determination of effects for long-term use of these garments is crucial due to current manufacturer recommendations of garment wear for 6-8 hours per day, regardless of age or diagnosis. We contend that recommendations for duration need to be based on objective data and may show that these flexible devices are most beneficial during specific times and activities, like when an infant is practicing a functional movement such as cruising. Thus, individuals may only need to wear these garments during practice of functional movements with possible modification of duration and frequency specific to their needs and goals.

Concluding Remarks

The series of research studies presented here examined the impact of assistive device use on well-practiced and developing gait patterns. In these three studies, our results show the adaptive mechanisms and flexibility employed by children with MMC and typically developing infants to produce gait while being influenced by use of an AD. For the two studies in which children with MMC walked with rigid ADs, our results showed more overt changes in children's gait patterns as evidenced by adjustments in segmental angle trajectories and gait characteristics. However, for TD infants cruising while wearing a flexible garment at the hips and pelvis, the changes exerted by infants to control their movements were more subtle as evidenced by little change in segmental angle trajectories, but more apparent adaptations were seen in the plots that elucidated underlying dynamic control processes: phase portrait and angle-angle plots. However, further research is warranted to investigate the impact of long-term use of these devices on both overt movement behaviors and the underlying control mechanisms before we can determine if use of ADs help or hinder functional mobility.

The studies presented here show that children with MMC and TD infants have the capacity and flexibility to acutely adapt their motor control strategies, segment coordination, and movement patterns to the application of external manipulations. What we don't know is if, over the long-term, these adaptations will result in decreased or increased dependency on ADs. Therefore, further research is warranted to investigate the impact of these devices on both overt movement behaviors and underlying control mechanisms before we can determine if use of ADs help or hinder functional mobility.

Appendix A.

Table A.1

Participant physical characteristics.

Participant	**Age**	**Lesion Level**	**More Involved leg**	**More Involved foot length (cm)**	**Less Involved foot length (cm)**	**Mean foot length (cm)**	**More Involved shank length (cm)**	**Less Involved shank length (cm)**	**Mean shank length (cm)**	**More Involved shank circumference (cm)**	**Less Involved shank circumference (cm)**
LS	5	L4/L5	L	14.80	14.40	14.60	21.80	20.20	21.00	20.90	20.50
EL	5	L4/L5	N/A	14.50	14.40	14.45	21.70	21.50	21.60	21.50	21.40
EW	5	L5	N/A	15.60	15.70	15.65	23.40	23.00	23.20	20.20	19.10
TR	5	S1	N/A	14.70	14.60	14.65	24.00	23.20	23.60	20.50	19.50
MD	6	L1	R	16.60	16.80	16.70	27.70	25.30	26.50	20.00	20.50
AD	7	L2	R	16.00	16.90	16.45	26.20	27.20	26.70	20.30	22.60
JB	7	L4	L	16.50	17.20	16.85	25.50	28.40	26.95	18.70	19.90
CV	11	L3/L4	R	17.60	17.90	17.75	27.50	26.50	27.00	21.30	21.20
SL	12	L3/L4	R	19.90	18.80	19.35	27.30	29.90	28.60	24.30	23.30
Overall											
M (SD)	7.00(2.69)			16.24(1.71)	16.30(1.61)	16.27(1.63)	25.01(2.38)	25.02(3.26)	25.02(2.71)	20.86(1.53)	20.89(1.39)
C+											
M(SD)	5.00(0.00)			14.90(0.48)	14.78(0.62)	14.84(0.55)	22.73(1.15)	21.98(1.41)	22.35(1.25)	20.78(0.56)	20.13(1.03)
C-											
M(SD)	8.60(2.70)			17.32(1.55)	17.52(0.83)	17.42(1.19)	26.84(0.95)	27.46(1.77)	27.15(0.84)	20.92(2.11)	21.50(1.42)

(L=Left leg; R=Right leg)

Table A.1 (continued)

Participant physical characteristics.

Participant	**More Involved thigh length (cm)**	**Less Involved thigh length (cm)**	**Mean thigh length (cm)**	**More Involved thigh circumference (cm)**	**Less Involved thigh circumference (cm)**	**More Involved full leg length (m)**	**Less Involved full leg length (m)**	**Mean full leg length (m)**	**Standing Height (cm)**	**Weight (kg)**
LS	24.30	24.00	24.15	36.20	36.40	0.50	0.50	0.50	108.50	21.82
EL	21.10	22.20	21.65	36.10	34.70	0.45	0.45	0.45	95.30	19.20
EW	26.40	24.40	25.40	31.00	30.40	0.52	0.52	0.52	112.20	18.40
TR	22.00	22.00	22.00	30.90	30.10	0.46	0.47	0.47	99.30	18.18
MD	26.50	26.80	26.65	35.00	35.60	0.56	0.57	0.57	112.90	23.18
AD	24.50	24.40	24.45	35.60	37.80	0.62	0.65	0.63	122.70	25.50
JB	28.20	29.00	28.60	38.60	31.50	0.55	0.57	0.56	122.80	27.05
CV	30.50	31.60	31.05	37.30	36.70	0.68	0.70	0.69	133.00	30.20
SL	31.00	30.20	30.60	43.80	43.90	0.66	0.63	0.64	137.80	42.00
Overall										
M (SD)	26.06(3.46)	26.07(3.50)	26.06(3.44)	36.06(3.90)	35.23(4.32)	0.56(0.08)	0.56(0.08)	0.56(0.08)	116.06(14.33)	25.06(7.56)
C+										
M(SD)	23.45(2.38)	23.15(1.23)	23.30(1.78)	33.55(3.00)	32.90(3.14)	0.48(0.03)	0.48(0.03)	0.48(0.03)	103.83(7.86)	19.40(1.67)
C-										
M(SD)	28.14(2.72)	28.40(2.85)	28.27(2.76)	38.06(3.51)	37.10(4.49)	0.61(0.06)	0.62(0.05)	0.62(0.05)	125.84(9.76)	29.59(7.39)

Table A.2

Seven-site skinfold measurements (in millimeters).

Participant	Group	Triceps (mm)	Biceps (mm)	Subscapularis (mm)	Supralliacus (mm)	Abdomen (mm)	Thigh (mm)	Calf (mm)
LS	C+	15.00	9.00	6.00	8.00	*U	12.00	18.00
EL	C+	17.00	21.00	16.00	18.00	20.00	25.00	29.00
EW	C+	8.00	7.00	6.00	5.00	5.00	12.00	14.00
TR	C+	11.50	11.00	11.00	6.00	10.00	20.00	13.00
MD	C-	21.75	10.50	10.00	19.00	18.00	24.17	22.50
AD	C-	*U	*U	*U	*U	*U	*U	*U
JB	C-	14.33	12.00	6.33	8.67	11.17	25.00	21.33
CV	C-	12.00	9.00	U	8.00	16.00	19.00	23.00
SL	C-	18.00	15.00	13.00	19.00	28.00	29.00	30.00
Overall	*M* (SD)	14.70(4.28)	11.81(4.41)	9.76(3.90)	11.46(6.09)	15.45(7.55)	20.77(6.24)	21.35(6.24)
C+	*M* (SD)	12.88(3.97)	12.00(6.22)	9.75(4.79)	9.25(5.97)	11.67(7.64)	17.25(6.40)	18.50(7.33)
C-	*M* (SD)	16.52(4.27)	11.63(2.56)	9.78(3.34)	13.67(6.16)	18.29(7.08)	24.29(4.11)	24.21(3.92)

*U=Unavailable.

Table A.3

Normalized Coefficient of Variation values for Center of Mass displacements, normalized to participant height.

		Independent (%)	***Crutches (%)***	***Walker (%)***	***Poles (%)***
Overall	*COM A/P*	36.44	33.61	28.94	31.92
	COM M/L	25.27	35.35	37.05	38.90
	COM Vertical	32.97	38.63	33.49	40.37
C+	*COM A/P*	24.67	46.30	32.02	31.04
	COM M/L	20.70	17.20	29.79	5.35
	COM Vertical	15.48	47.23	19.12	38.56
C-	*COM A/P*	44.12	20.76	29.79	33.53
	COM M/L	28.93	48.86	44.31	56.10
	COM Vertical	40.91	35.46	45.33	43.42

Table A.4

Gait characteristics overall and by subgroup.

	Condition	**Cadence (step/min)**	**Step Velocity (cm/sec)**	**Cycle Time (sec)**	**Stance phase (%)**	**Double support Phase (%)**	**Normalized Cadence**	**Normalized Velocity**	**Normalized BOS Width**	**Normalized Step Length**	**Normalized Stride Length**	**Normalized Cycle Time**	**Normalized Step Width**
		M (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)
Overall	Independent	129.84(13.98)	81.73(14.06)	0.97(0.12)	64.46(4.74)	22.25(4.65)	0.51(0.05)	0.35(0.05)	0.27(0.06)	0.68(0.10)	1.37(0.19)	4.09(0.40)	0.67(0.23)
	Crutches	112.80(22.60)	62.76(13.35)	1.09(0.26)	65.69(5.94)	22.42(4.42)	0.44(0.08)	0.27(0.05)	0.28(0.08)	0.63(0.14)	1.24(0.29)	4.78(0.93)	0.69(0.13)
	Walker	115.79(22.15)	62.05(15.93)	1.12(0.24)	65.12(5.56)	25.18(6.12)	0.46(0.08)	0.27(0.06)	0.23(0.08)	0.59(0.10)	1.18(0.23)	4.62(0.85)	0.64(0.09)
	Poles	119.43(24.98)	67.61(21.40)	1.09(0.24)	64.42(4.99)	21.42(4.63)	0.47(0.10)	0.29(0.08)	0.25(0.07)	0.64(0.16)	1.27(0.33)	4.57(0.98)	0.70(0.12)
C+	Independent	135.49(8.69)	79.65(6.03)	0.92(0.09)	68.38(3.39)	20.85(4.03)	0.50(0.02)	0.37(0.03)	0.24(0.04)	0.73(0.05)	1.46(0.10)	4.15(0.29)	0.59(0.34)
	Crutches	117.67(30.43)	59.60(9.34)	1.11(0.32)	70.70(5.31)	20.72(5.13)	0.43(0.10)	0.27(0.05)	0.29(0.06)	0.67(0.19)	1.35(0.38)	4.95(1.29)	0.73(0.18)
	Walker	124.97(20.29)	63.02(9.18)	1.03(0.18)	69.62(3.27)	23.18(7.04)	0.46(0.06)	0.29(0.04)	0.24(0.04)	0.64(0.12)	1.27(0.27)	4.51(0.77)	0.69(0.08)
	Poles	121.49(26.19)	59.95(9.90)	1.08(0.28)	68.42(4.03)	20.28(4.88)	0.45(0.09)	0.28(0.04)	0.27(0.02)	0.65(0.17)	1.29(0.38)	4.81(1.11)	0.71(0.15)
C-	Independent	125.32(16.64)	83.38(18.99)	1.01(0.14)	61.32(2.95)	23.36(5.24)	0.51(0.06)	0.33(0.07)	0.28(0.07)	0.64(0.11)	1.30(0.23)	4.04(0.50)	0.72(0.08)
	Crutches	108.90(16.85)	65.29(16.53)	1.08(0.23)	61.69(2.07)	23.78(3.76)	0.45(0.07)	0.27(0.07)	0.27(0.10)	0.59(0.09)	1.16(0.20)	4.65(0.64)	0.65(0.08)
	Walker	108.44(22.81)	61.29(21.05)	1.20(0.26)	61.53(4.18)	26.77(5.53)	0.45(0.09)	0.26(0.08)	0.21(0.11)	0.55(0.08)	1.10(0.18)	4.70(0.99)	0.60(0.08)
	Poles	117.79(26.95)	73.73(27.15)	1.10(0.24)	61.22(2.95)	22.33(4.77)	0.49(0.11)	0.30(0.11)	0.23(0.09)	0.63(0.17)	1.25(0.33)	4.38(0.94)	0.69(0.10)

Table A.5

Segmental angle range of motion across the stride cycle, overall, and by subgroup.

Maximum:

		Independent	***Crutches***	***Walker***	***Poles***
	Segmental Angle	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)
Overall	*LI thigh F/E*	96.25 (8.14)	93.61 (7.49)	94.66 (5.83)	94.10 (7.01)
	MI thigh F/E	102.08 (20.04)	91.91 (6.73)	95.14 (6.44)	93.43 (4.46)
	LI shank F/E	140.29 (13.51)	138.98 (11.15)	138.18 (11.65)	139.83 (13.66)
	MI shank F/E	136.79 (13.57)	135.69 (11.76)	137.11 (12.81)	136.90 (11.79)
	LI ankle DF/PF	98.47 (49.24)	93.90 (52.84)	97.20 (46.80)	96.68 (50.24)
	MI ankle DF/PF	97.02 (42.83)	92.42 (45.05)	94.92 (43.58)	95.97 (45.13)
C+	*LI thigh F/E*	98.37 (6.97)	94.30 (9.24)	96.89 (8.22)	94.56 (6.76)
	MI thigh F/E	97.51 (3.02)	92.62 (7.67)	97.12 (8.51)	93.56 (6.18)
	LI shank F/E	141.29 (12.17)	140.13 (5.31)	138.45 (8.74)	138.92 (9.01)
	MI shank F/E	144.60 (8.75)	143.07 (4.35)	143.32 (11.06)	143.35 (2.19)
	LI ankle DF/PF	113.09 (55.35)	103.13 (65.36)	102.38 (61.73)	102.34 (65.90)
	MI ankle DF/PF	108.99 (55.47)	100.20 (60.17)	102.81 (61.53)	101.78 (65.54)
C-	*LI thigh F/E*	94.56 (9.38)	93.06 (6.88)	92.88 (2.89)	93.73 (7.97)
	MI thigh F/E	105.73 (27.55)	91.19 (6.74)	93.15 (3.74)	93.32 (3.35)
	LI shank F/E	139.49 (15.87)	138.05 (15.00)	137.97 (14.63)	140.56 (17.63)
	MI shank F/E	130.55 (14.19)	129.78 (12.82)	132.15 (12.92)	131.73 (14.12)
	LI ankle DF/PF	86.77 (46.54)	86.51 (47.19)	93.05 (38.40)	92.16 (41.63)
	MI ankle DF/PF	87.45 (33.22)	86.19 (35.13)	88.61 (29.11)	91.32 (28.11)

Table A.5 (continued)

Minimum:

	Segmental Angle	***Independent*** *M* (SD)	***Crutches*** *M* (SD)	***Walker*** *M* (SD)	***Poles*** *M* (SD)
Overall	*LI thigh F/E*	57.46 (6.35)	55.55 (5.34)	57.90 (6.91)	56.17 (5.85)
	MI thigh F/E	62.04 (8.41)	56.94 (8.24)	59.82 (6.96)	56.34 (8.50)
	LI shank F/E	81.55 (4.06)	86.30 (4.72)	85.02 (5.02)	82.53 (4.13)
	MI shank F/E	83.08 (4.38)	85.24 (5.37)	86.35 (5.11)	83.39 (5.63)
	LI ankle DF/PF	40.14 (38.39)	41.01 (42.61)	41.18 (41.18)	42.30 (39.44)
	MI ankle DF/PF	44.54 (32.65)	46.81 (35.46)	50.08 (34.04)	46.13 (34.71)
C+	*LI thigh F/E*	58.97 (4.97)	56.50 (6.38)	61.43 (7.96)	57.68 (4.60)
	MI thigh F/E	60.82 (6.90)	54.88 (8.82)	58.20 (7.88)	55.90 (7.37)
	LI shank F/E	81.39 (2.54)	83.57 (2.58)	82.19 (4.56)	82.39 (2.93)
	MI shank F/E	85.62 (2.85)	87.94 (6.62)	84.59 (5.36)	86.41 (4.38)
	LI ankle DF/PF	56.30 (41.46)	52.54 (52.56)	50.25 (50.78)	50.55 (49.15)
	MI ankle DF/PF	54.50 (44.36)	53.44 (51.47)	52.44 (48.47)	53.01 (51.45)
C-	*LI thigh F/E*	56.24 (7.61)	54.80 (4.98)	55.08 (5.05)	54.97 (6.97)
	MI thigh F/E	63.02 (10.15)	59.00 (8.32)	61.43 (6.63)	56.70 (10.18)
	LI shank F/E	81.68 (5.31)	88.48 (5.12)	87.28 (4.53)	82.65 (5.25)
	MI shank F/E	81.05 (4.55)	83.09 (3.41)	87.75 (5.00)	80.98 (5.70)
	LI ankle DF/PF	27.21 (34.46)	31.78 (36.33)	33.92 (36.18)	35.69 (34.30)
	MI ankle DF/PF	36.57 (21.85)	41.50 (21.20)	48.19 (23.37)	40.63 (18.43)

Table A.5 (continued)

Displacement:

	Segmental Angle	***Independent*** *M* (SD)	***Crutches*** *M* (SD)	***Walker*** *M* (SD)	***Poles*** *M* (SD)
Overall	*LI thigh F/E*	37.80 (8.81)	38.61 (9.61)	36.04 (7.29)	39.19 (9.48)
	MI thigh F/E	37.25 (9.33)	34.84 (10.31)	34.08 (5.86)	38.55 (10.60)
	LI shank F/E	59.23 (9.92)	52.97 (10.32)	52.73 (8.46)	55.88 (12.45)
	MI shank F/E	53.78 (13.13)	50.56 (12.53)	49.26 (11.47)	53.18 (13.04)
	LI ankle DF/PF	60.24 (15.53)	53.21 (14.84)	53.46 (12.11)	51.80 (16.49)
	MI ankle DF/PF	52.24 (14.48)	46.35 (12.86)	44.96 (13.30)	48.03 (14.49)
C+	*LI thigh F/E*	37.85 (9.46)	39.04 (11.65)	33.83 (6.95)	38.69 (10.77)
	MI thigh F/E	36.86 (7.74)	37.50 (11.79)	36.45 (5.22)	40.28 (9.19)
	LI shank F/E	59.03 (9.63)	57.22 (5.94)	55.36 (5.71)	56.59 (7.82)
	MI shank F/E	58.60 (10.46)	55.40 (10.12)	55.47 (8.30)	59.27 (6.55)
	LI ankle DF/PF	54.47 (20.23)	51.31 (15.24)	50.23 (15.95)	53.35 (18.06)
	MI ankle DF/PF	51.37 (18.28)	48.42 (11.42)	50.03 (16.73)	49.23 (15.00)
C-	*LI thigh F/E*	37.77 (9.38)	38.26 (9.09)	37.80 (7.83)	39.59 (9.61)
	MI thigh F/E	37.56 (11.35)	32.19 (9.50)	31.72 (6.17)	37.16 (12.49)
	LI shank F/E	59.39 (11.28)	49.58 (12.41)	50.63 (10.32)	55.31 (16.23)
	MI shank F/E	49.92 (14.85)	46.69 (13.97)	44.30 (11.93)	48.31 (15.54)
	LI ankle DF/PF	64.86 (10.76)	54.74 (16.12)	56.05 (9.16)	50.57 (17.17)
	MI ankle DF/PF	52.95 (12.94)	44.69 (15.00)	40.90 (9.89)	47.08 (15.77)

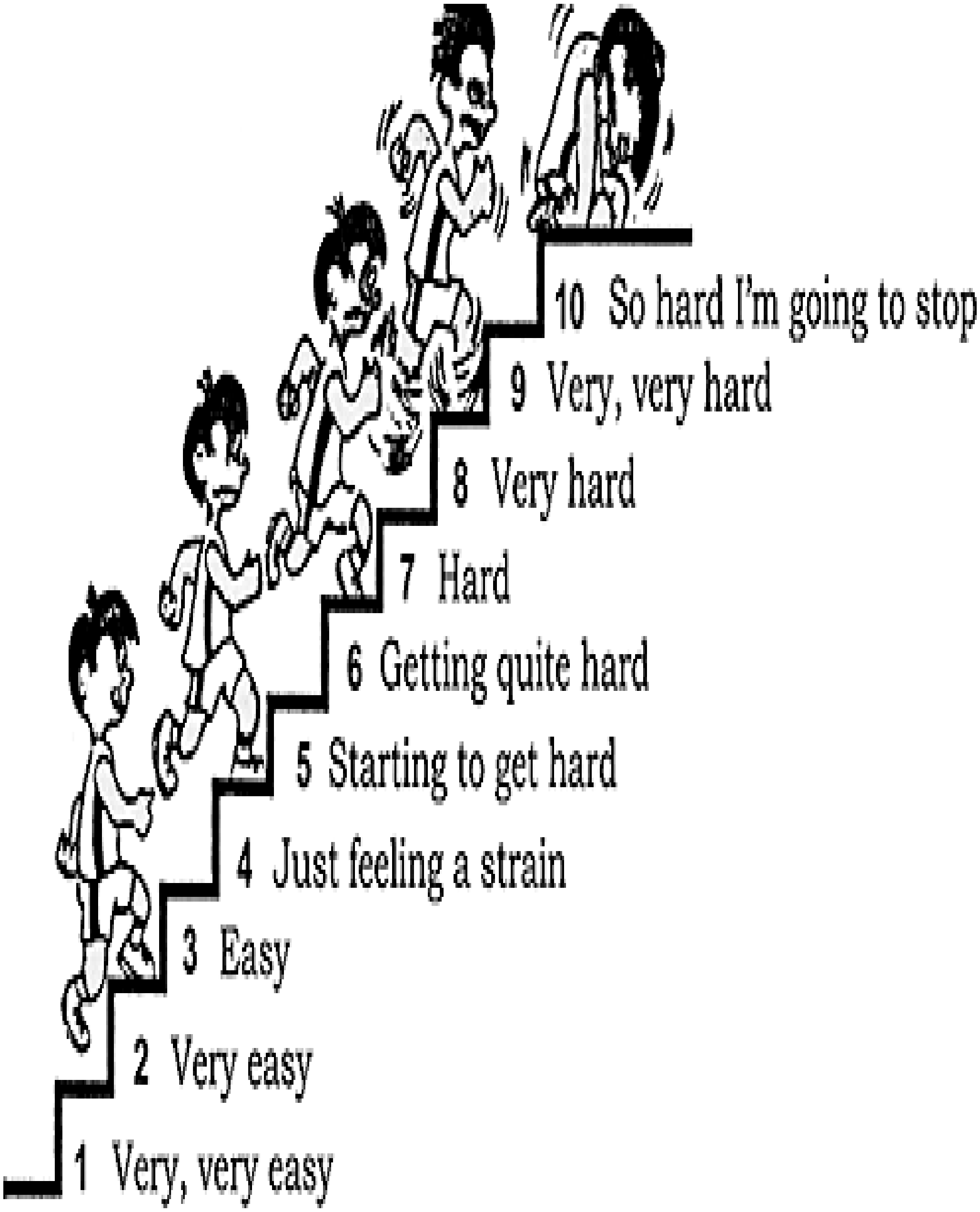


Figure A.1. Pictorial Children's Effort Rating Table (PCERT).

Appendix B.

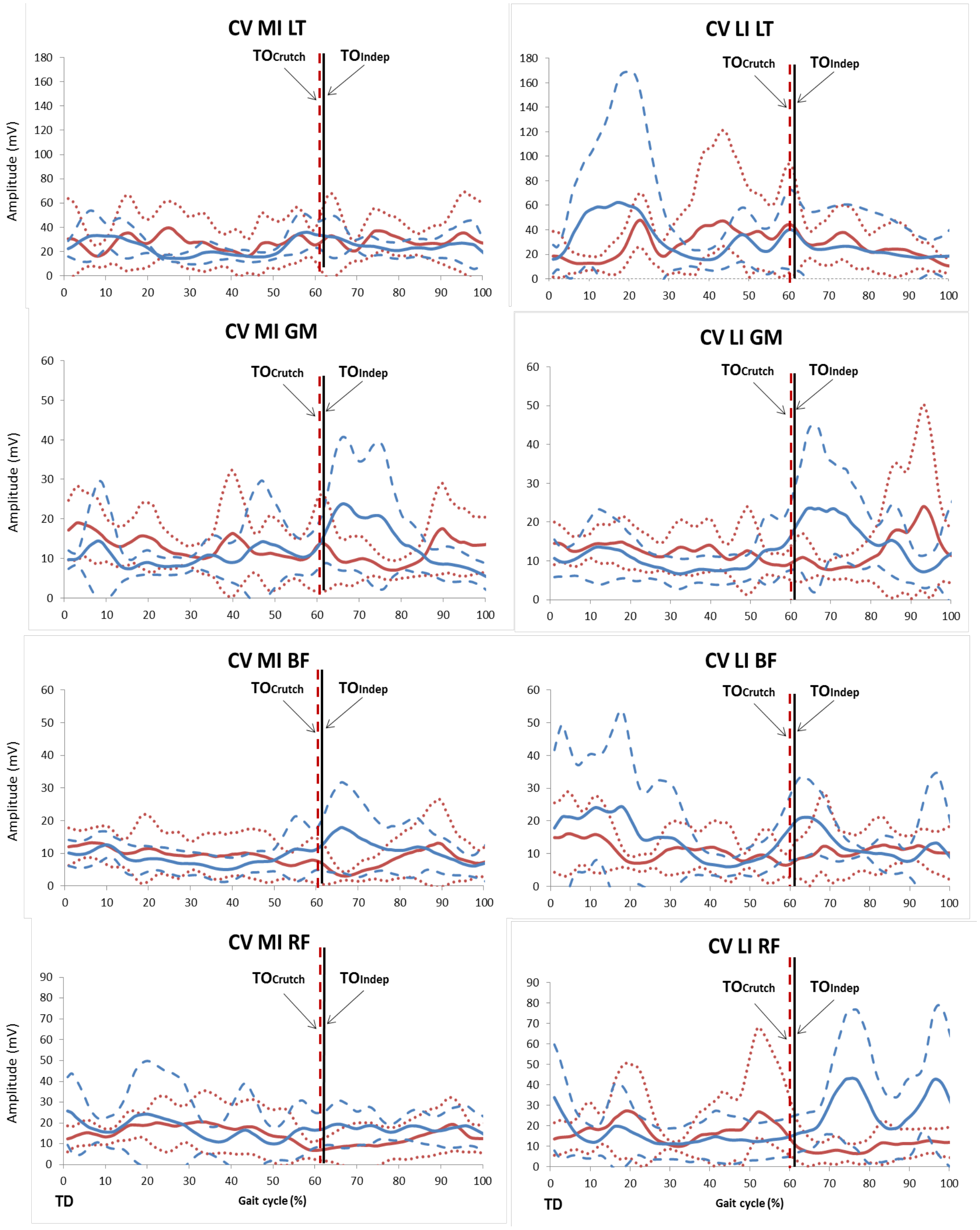


Figure B.1.1. Crutch electromyographic ensemble average traces with ±1 standard deviation envelopes for individual participants compared to independent walking. (In chart title: first 2 initials=participant's initials; MI=More involved leg; LI=Less involved leg; LT=Lower trapezius; GM=Gluteus medius; BF=Biceps femoris; RF=Rectus femoris; TO=Toe off; TD=Touch down; — Independent average; – – Independent standard deviation; — Crutch average;– – Crutch standard deviation.)

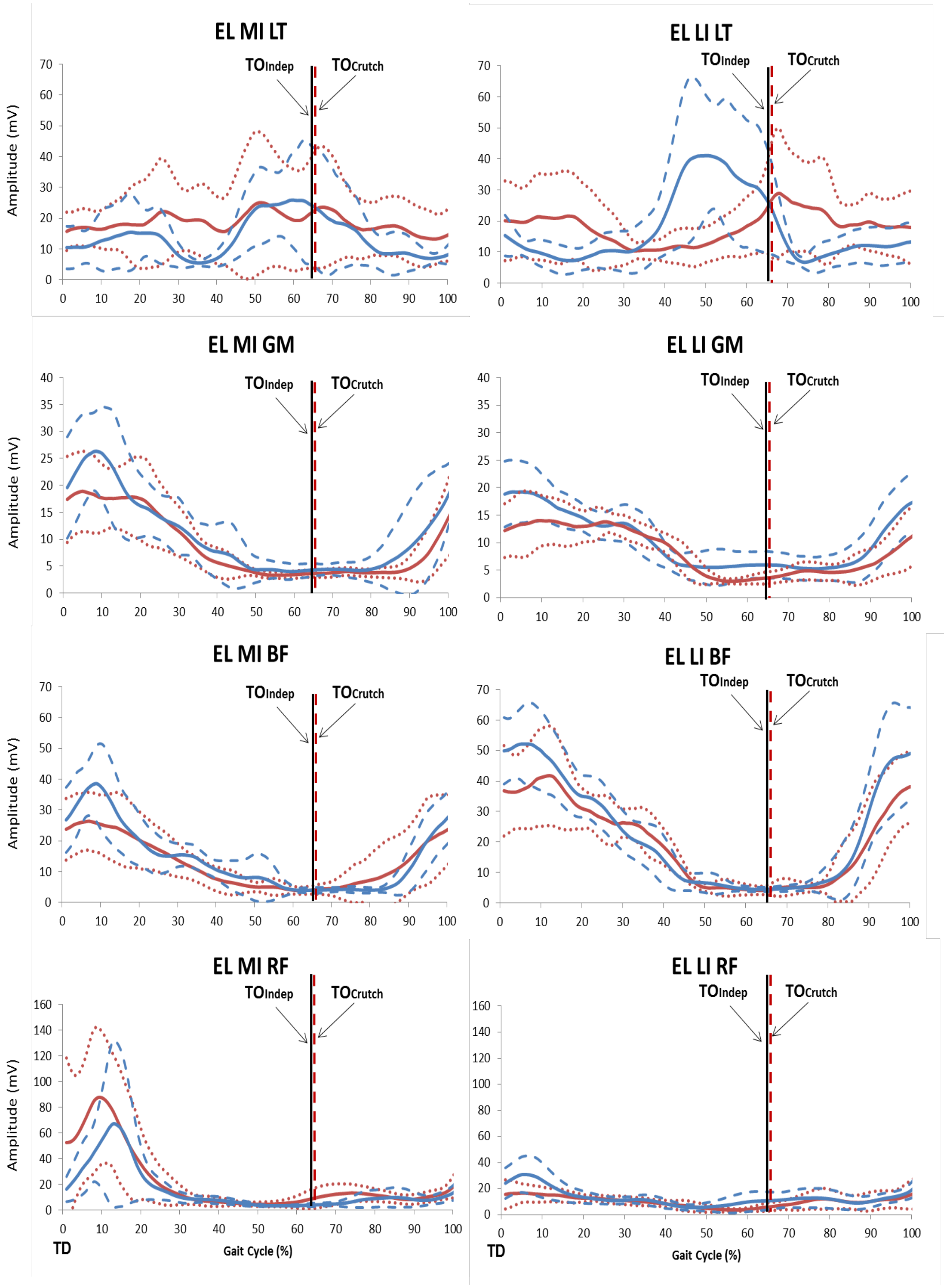


Figure B.1.1. (continued).

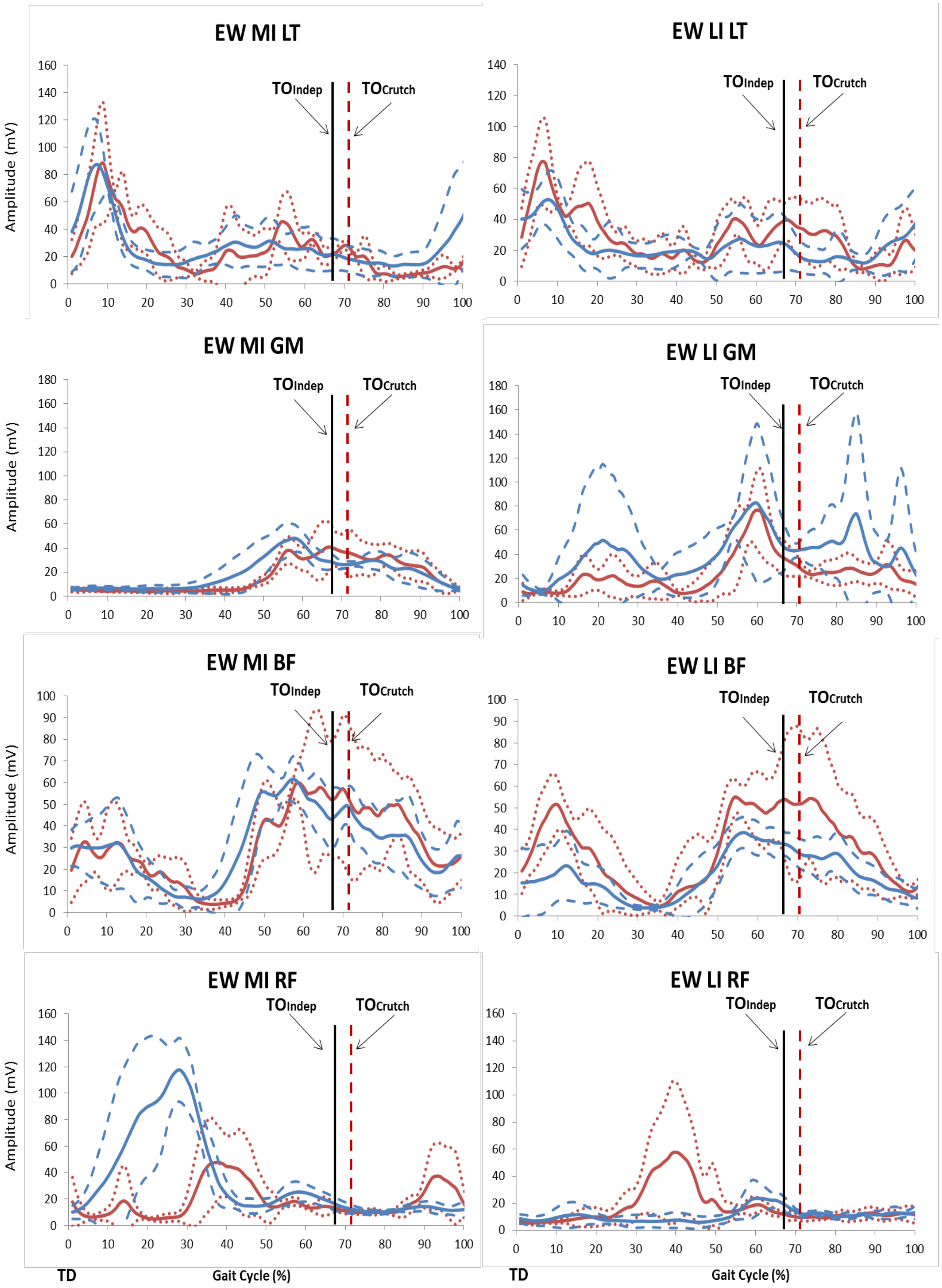


Figure B.1.1. (continued).

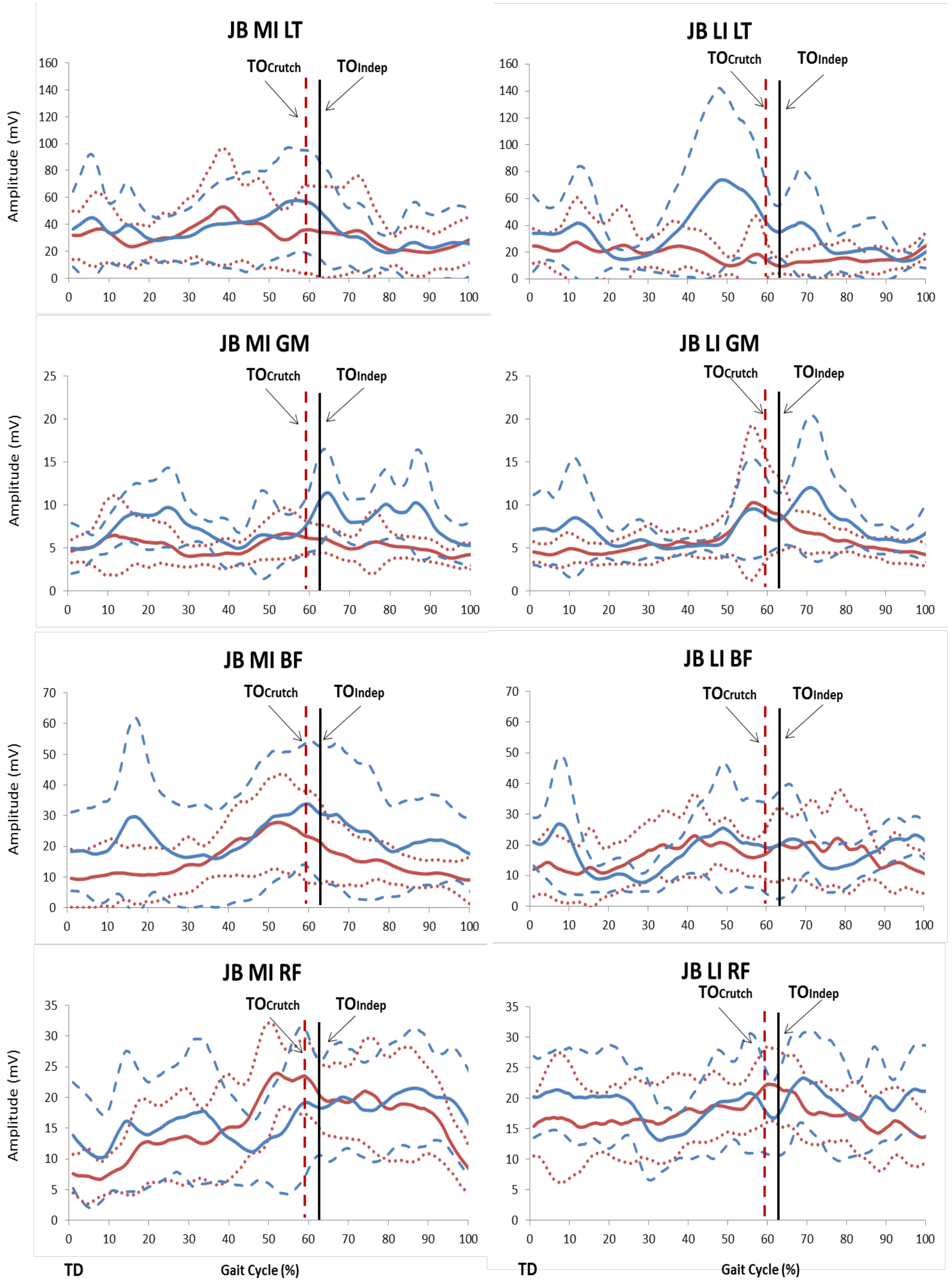


Figure B.1.1. (continued).

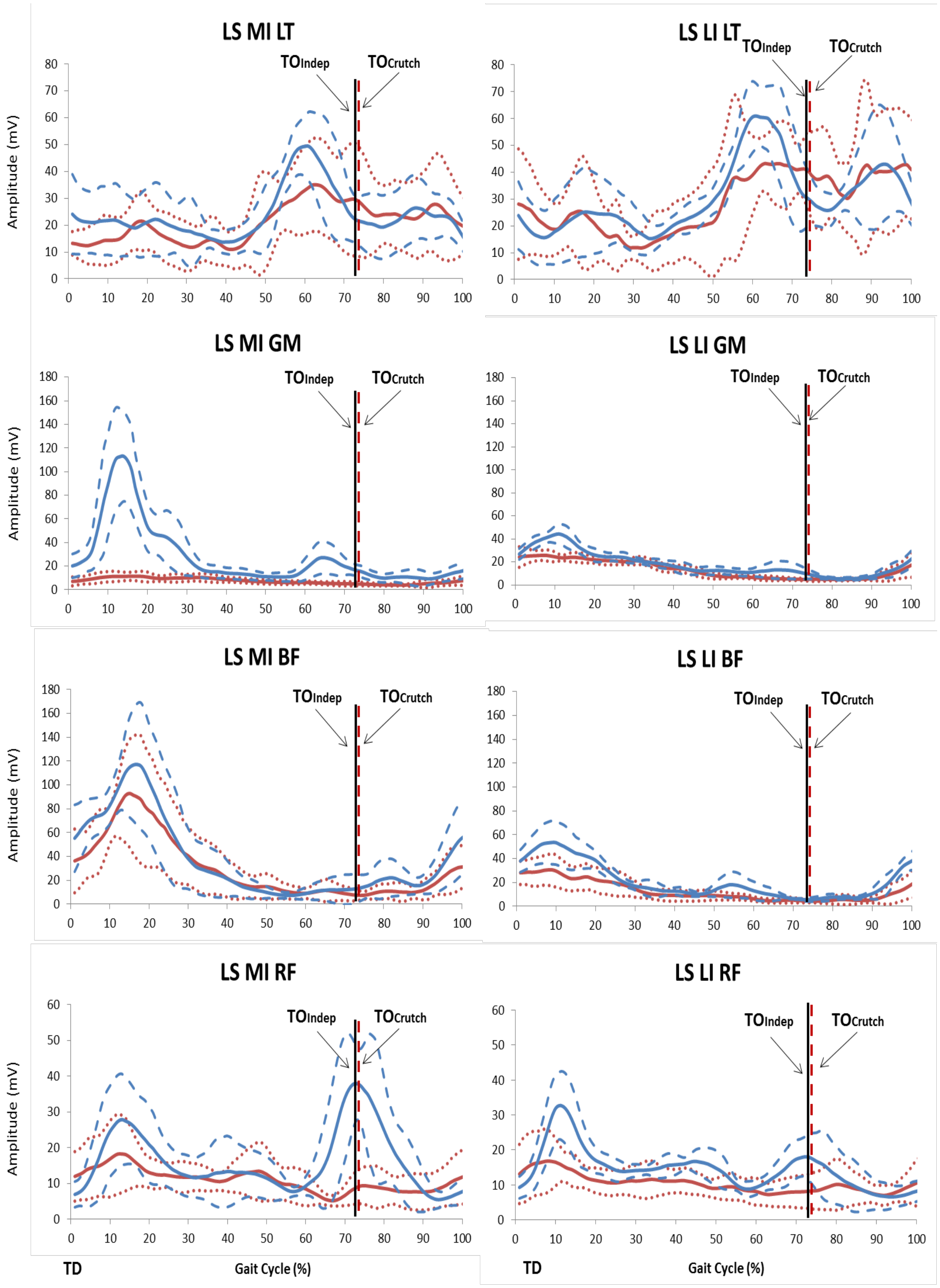
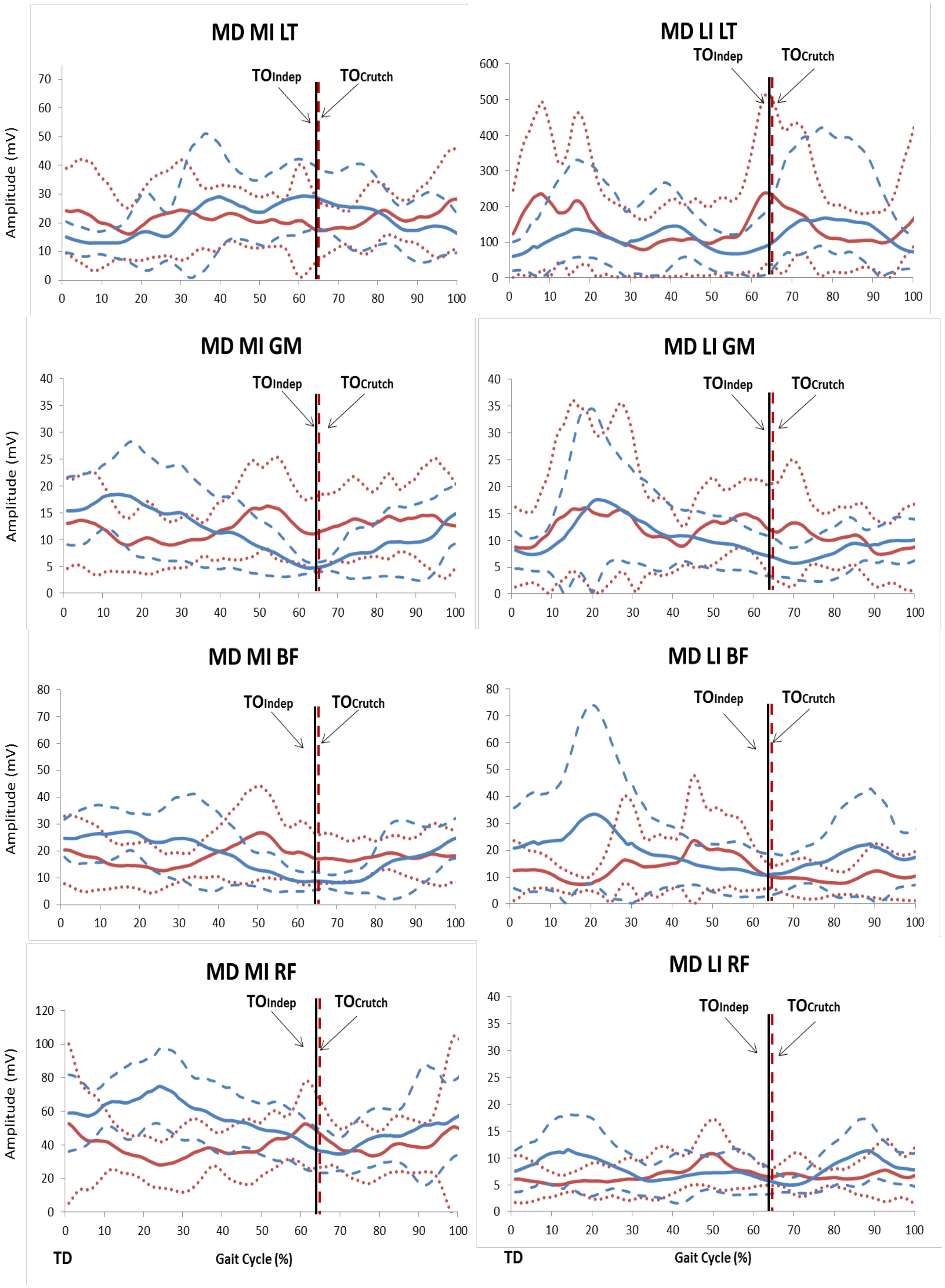


Figure B.1.1. (continued).



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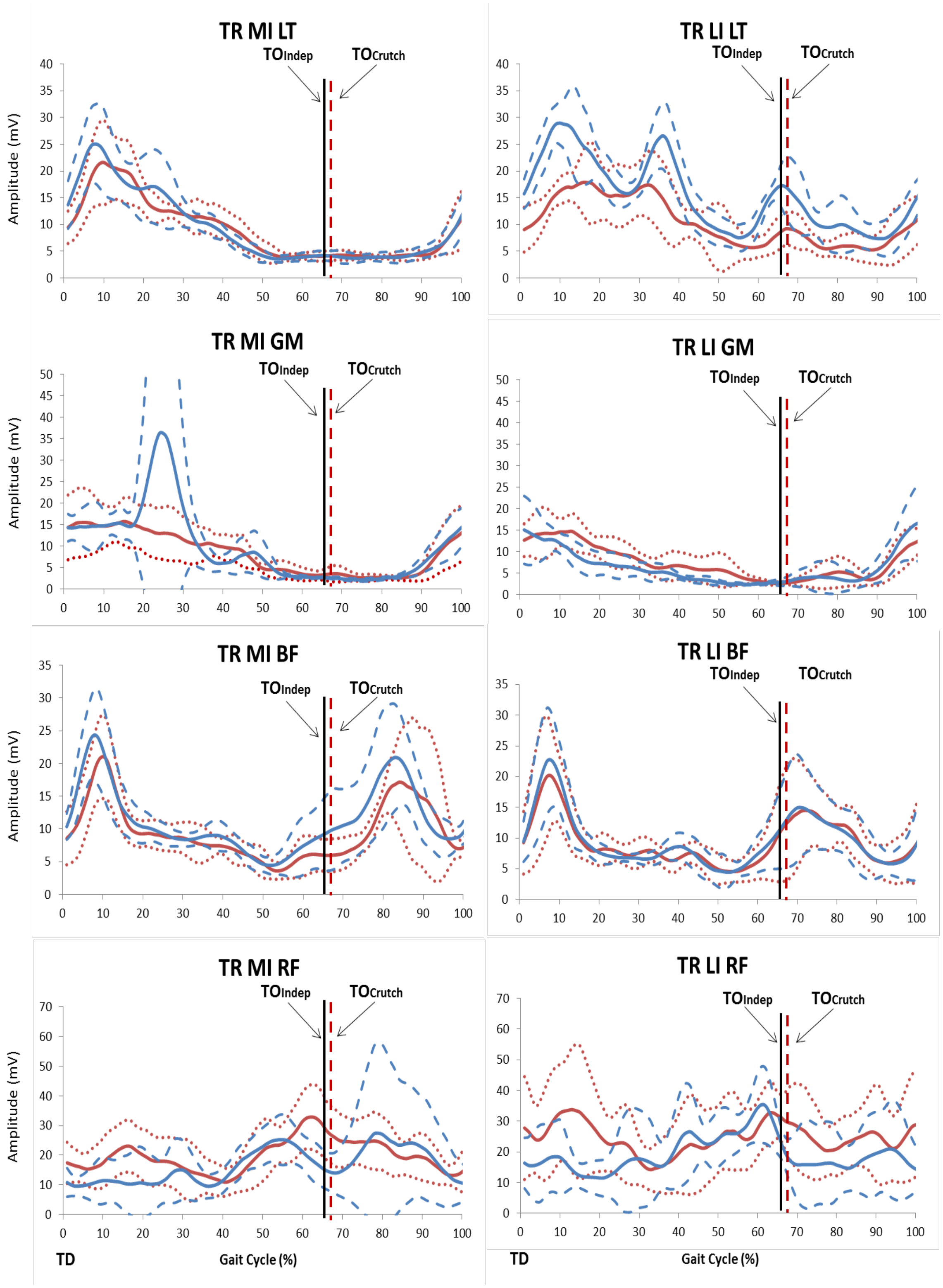

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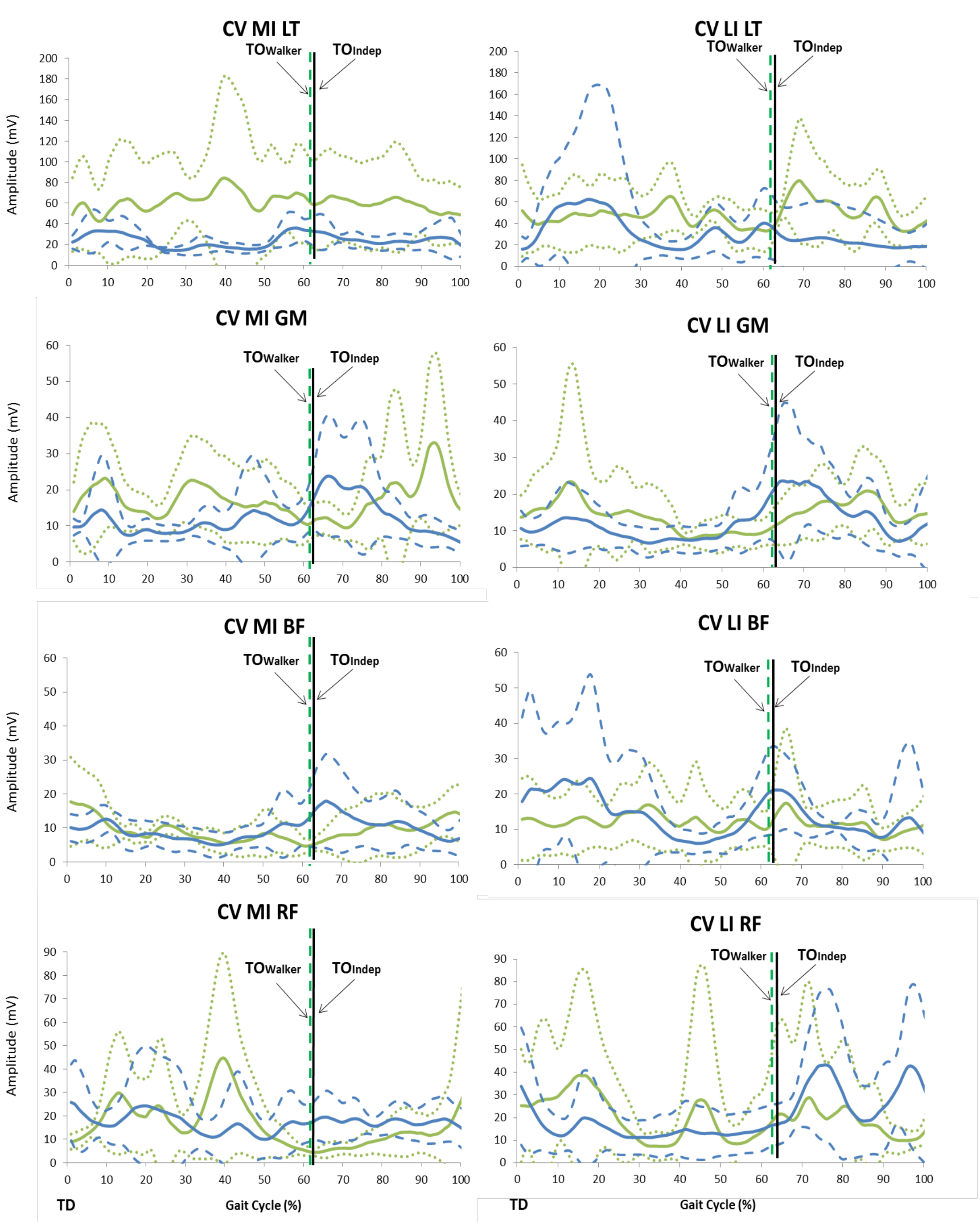


Figure B.1.2. Walker electromyographic ensemble average traces with ±1 standard deviation envelopes for individual participants compared to independent walking. (In chart title: first 2 initials=participant’s initials; MI=More involved leg; LI=Less involved leg; LT=Lower trapezius; GM=Gluteus medius; BF=Biceps femoris; RF=Rectus femoris; TO=Toe off; TD=Touch down; — Independent average; - - Independent standard deviation; —Walker average; - -Walker standard deviation.)

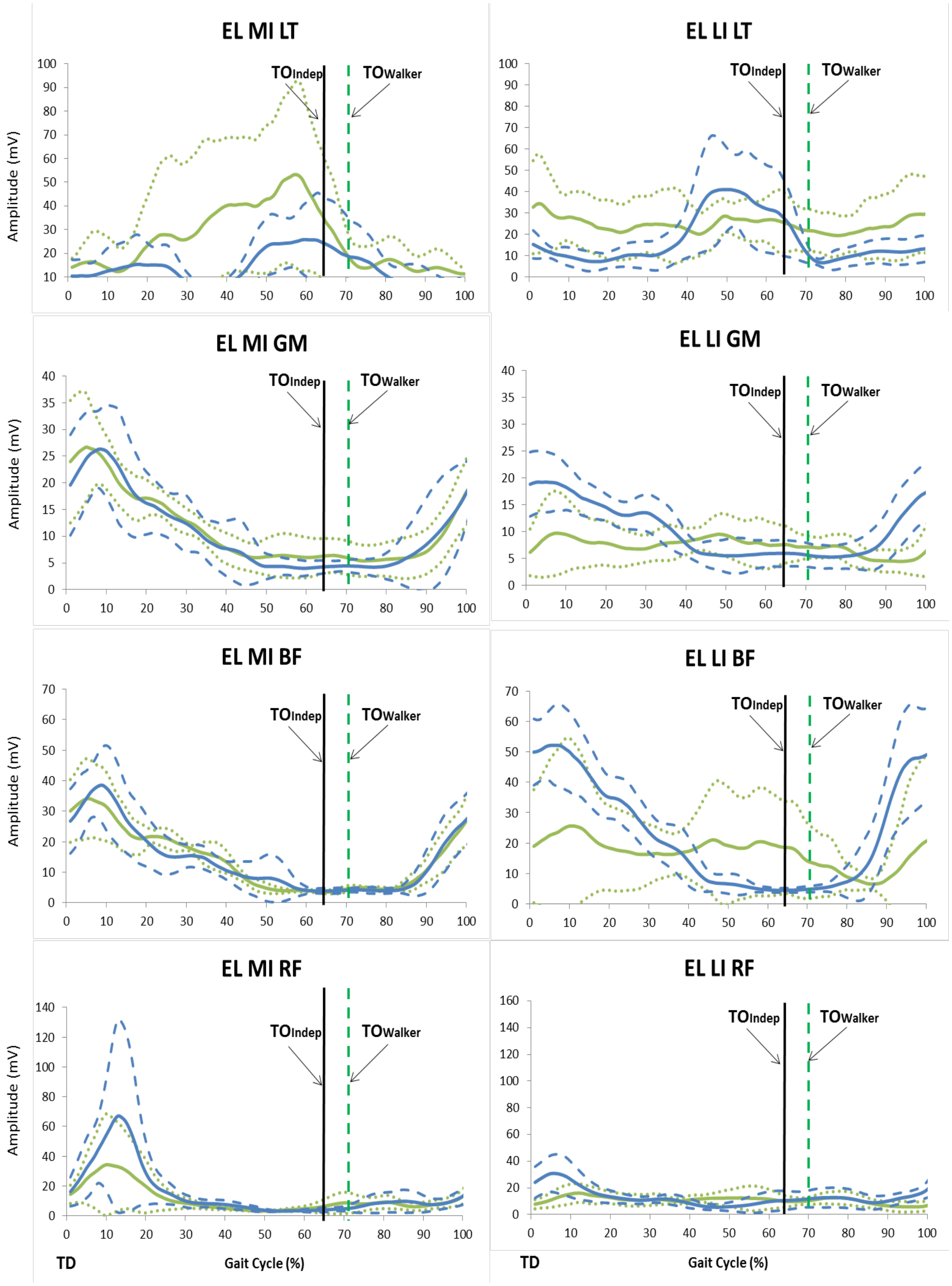
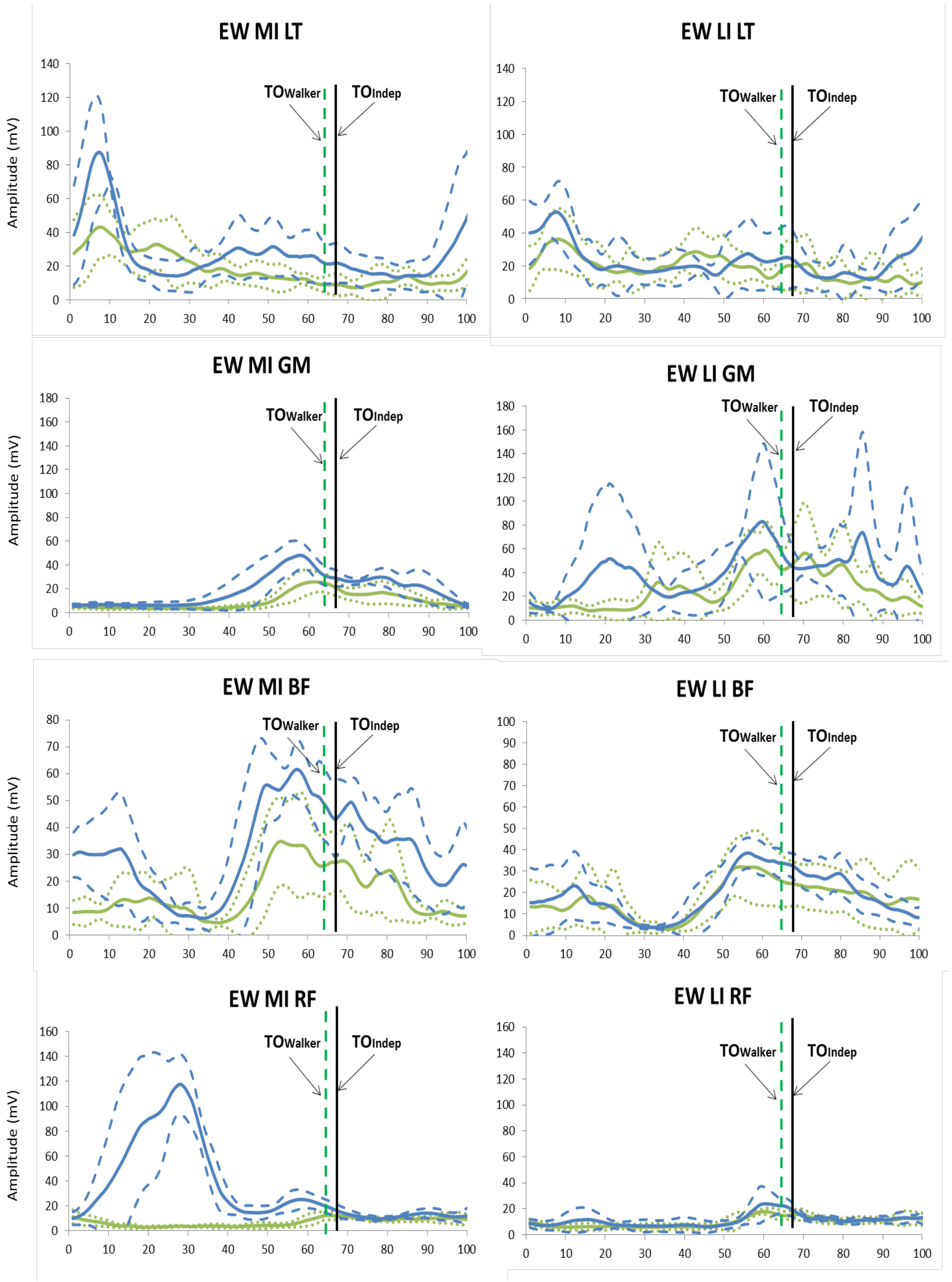


Figure B.1.2. (continued).



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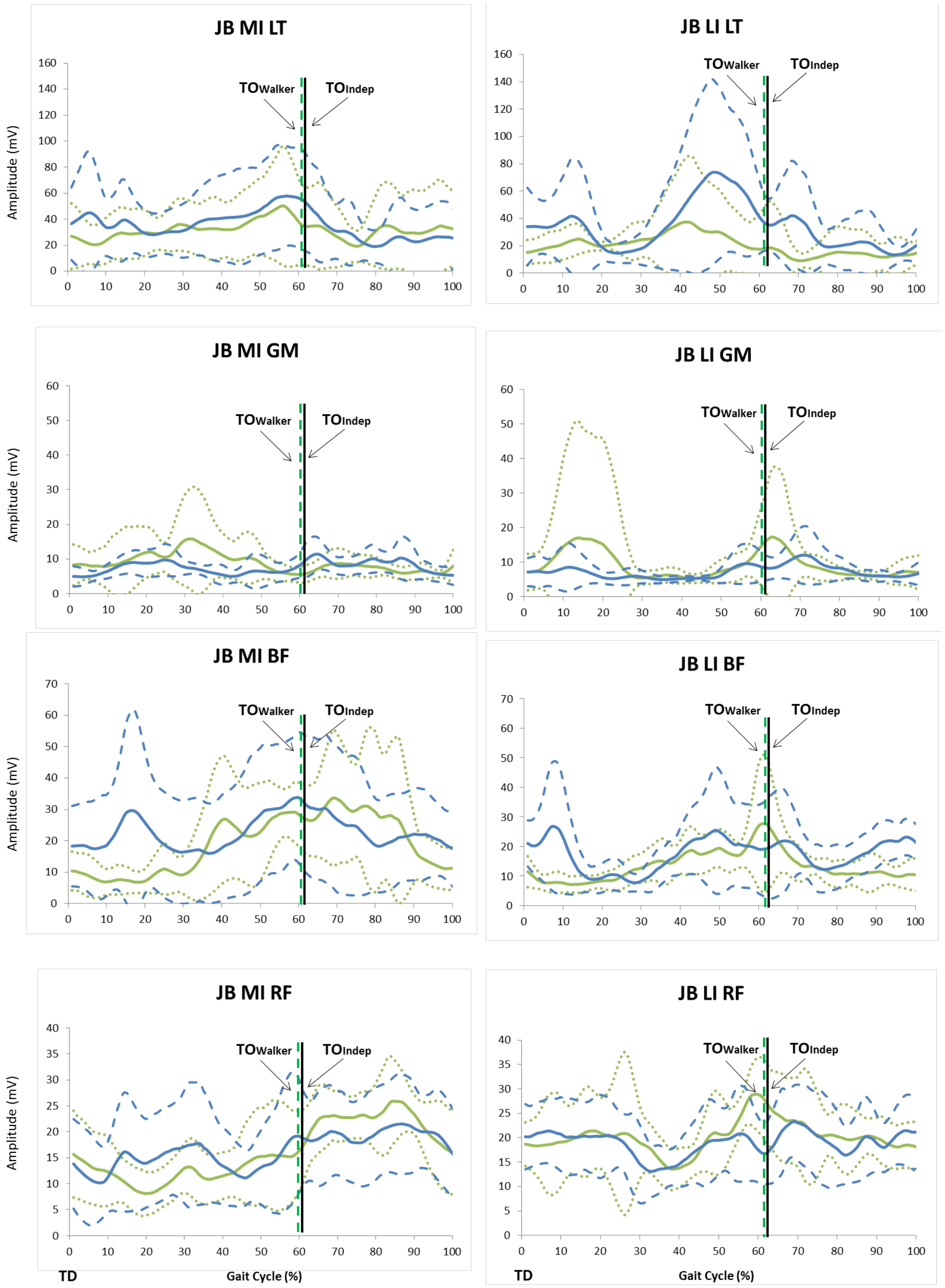

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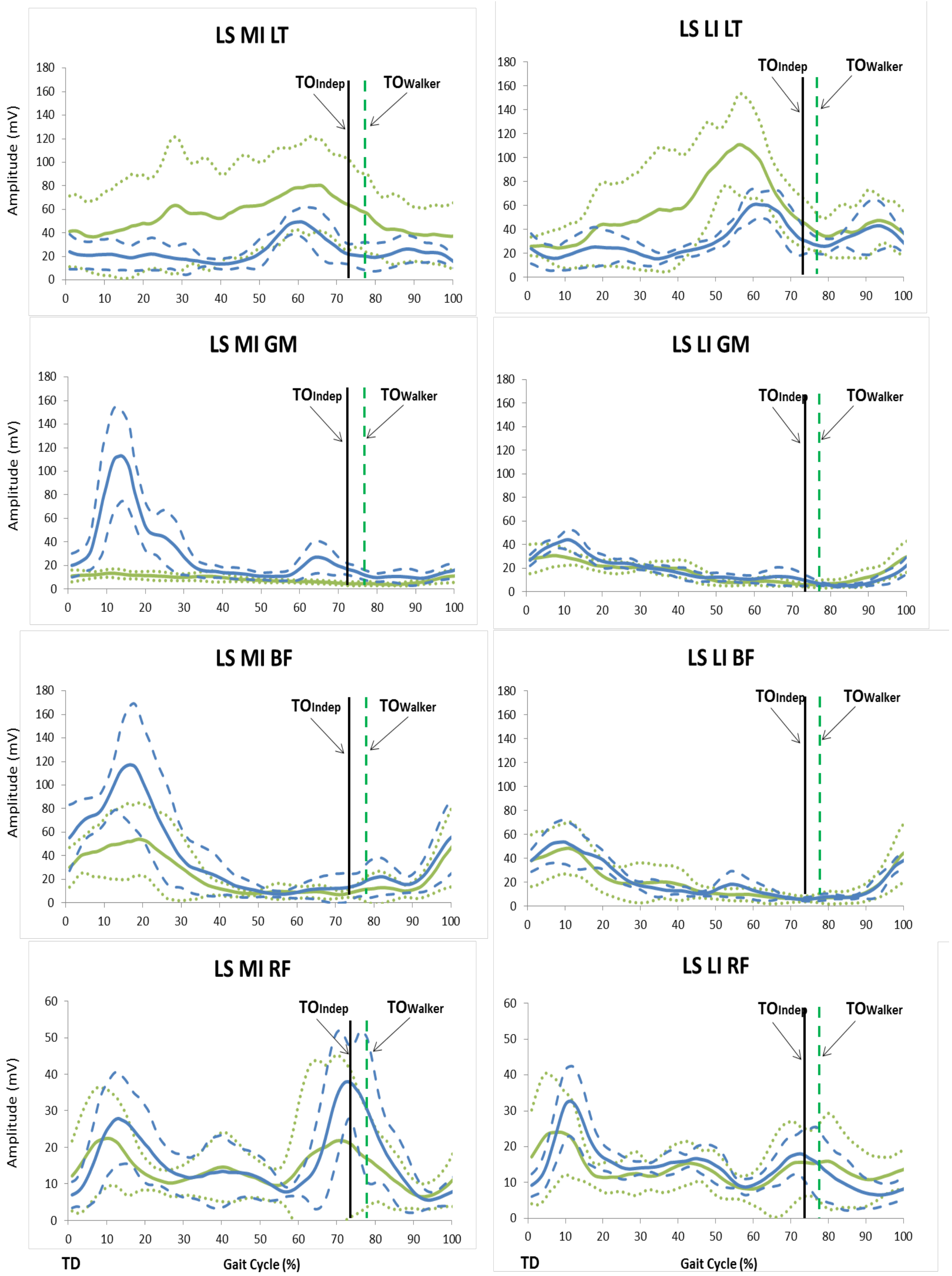

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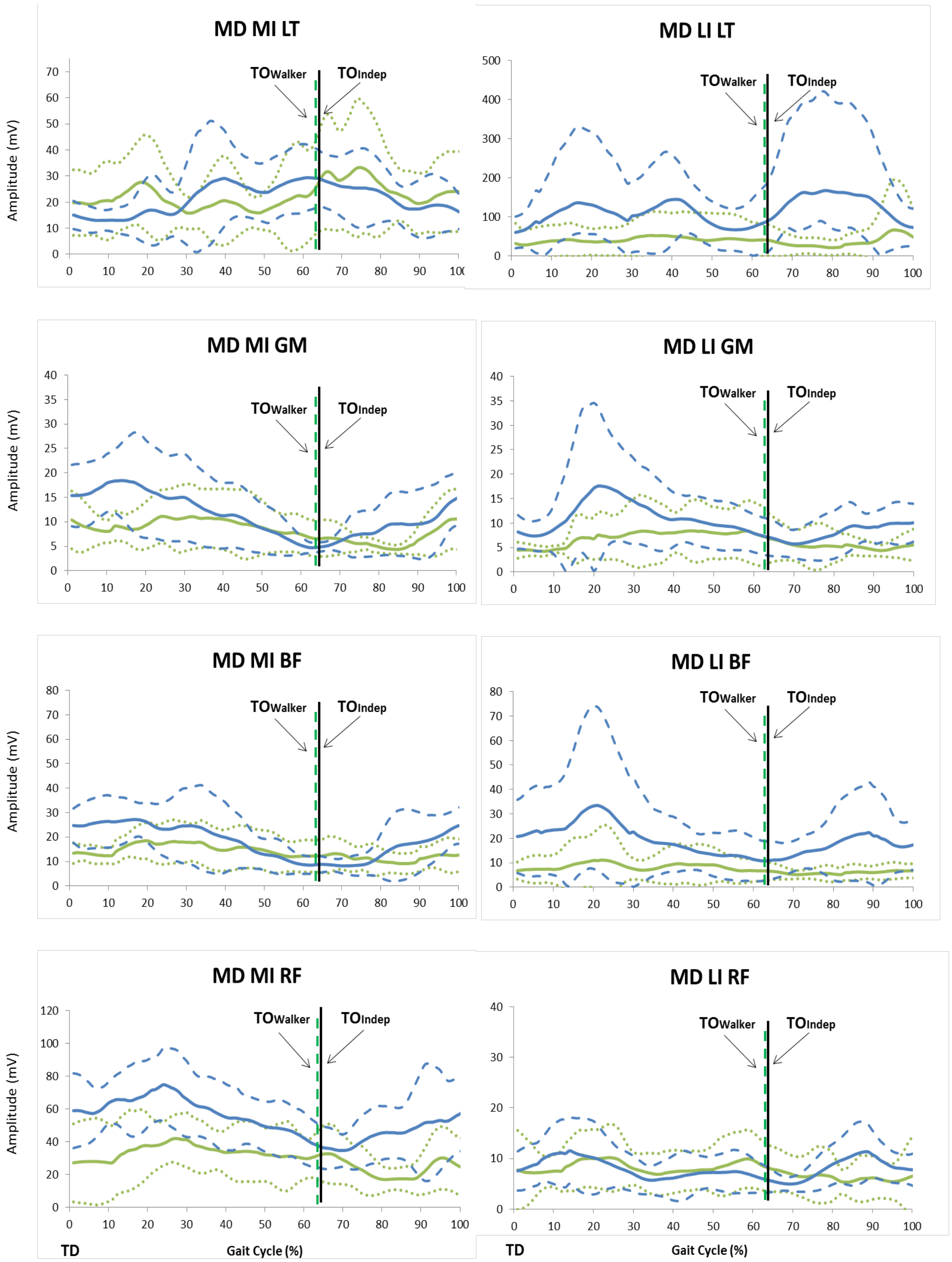

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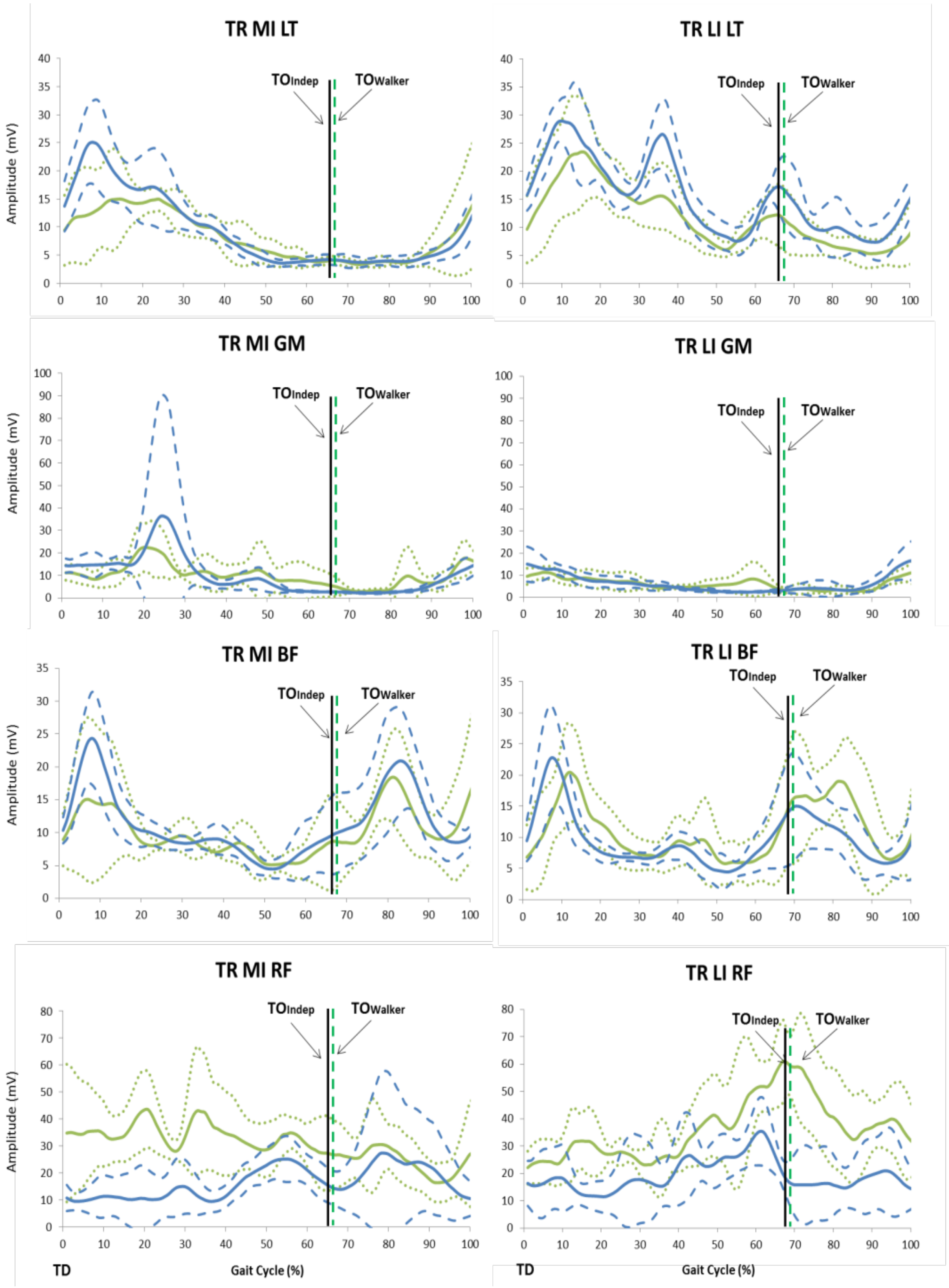


Figure B.1.2. (continued).

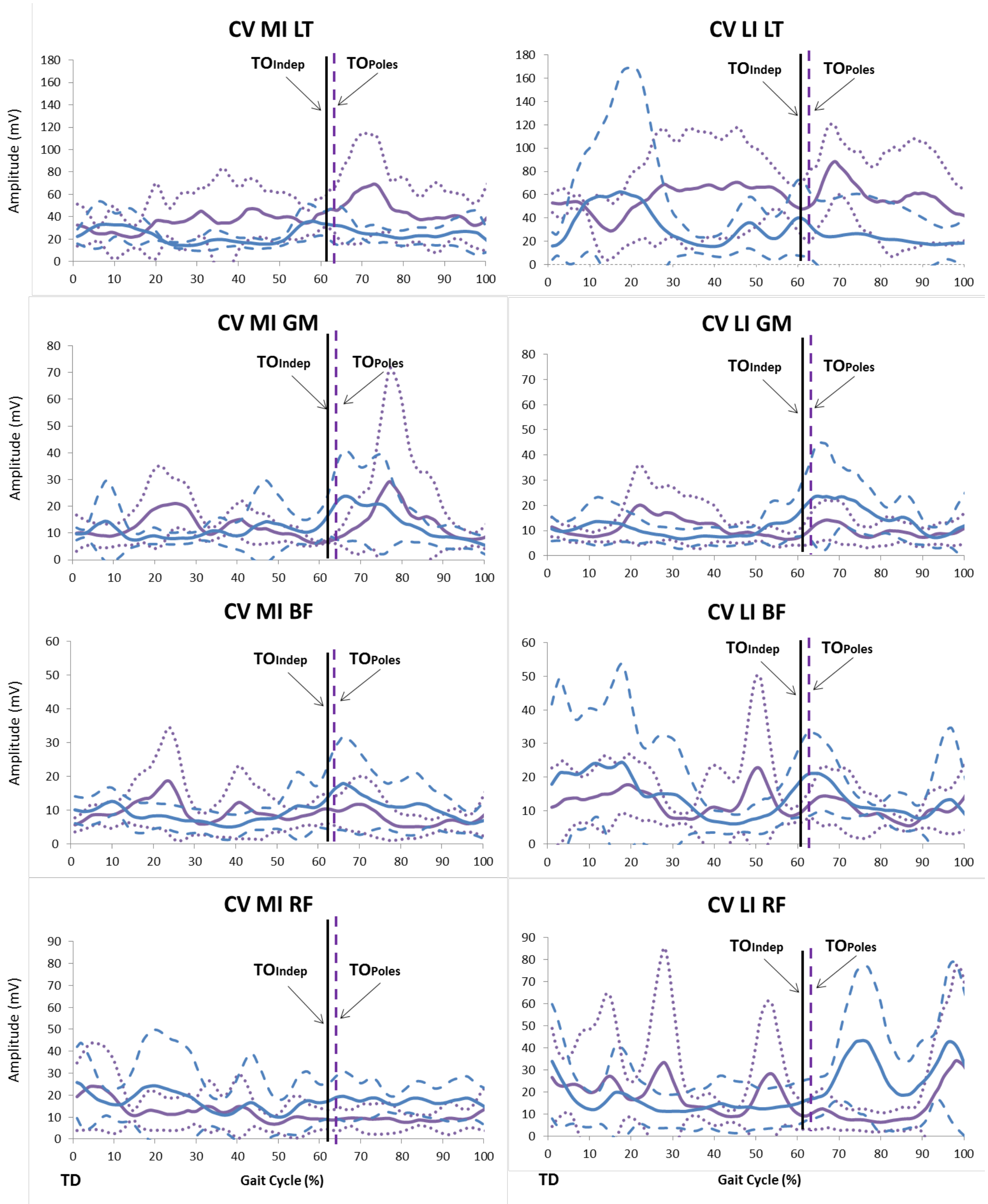


Figure B.1.3. Pole electromyographic ensemble average traces with ±1 standard deviation envelopes for individual participants compared to independent walking.
(In chart title: first 2 initials=participant's initials; MI=More involved leg; LI=Less involved leg; LT=Lower trapezius; GM=Gluteus medius; BF=Biceps femoris; RF=Rectus femoris; TO=Toe off; TD=Touch down; — Independent average; - - Independent standard deviation; — Pole average; - - Pole standard deviation.)

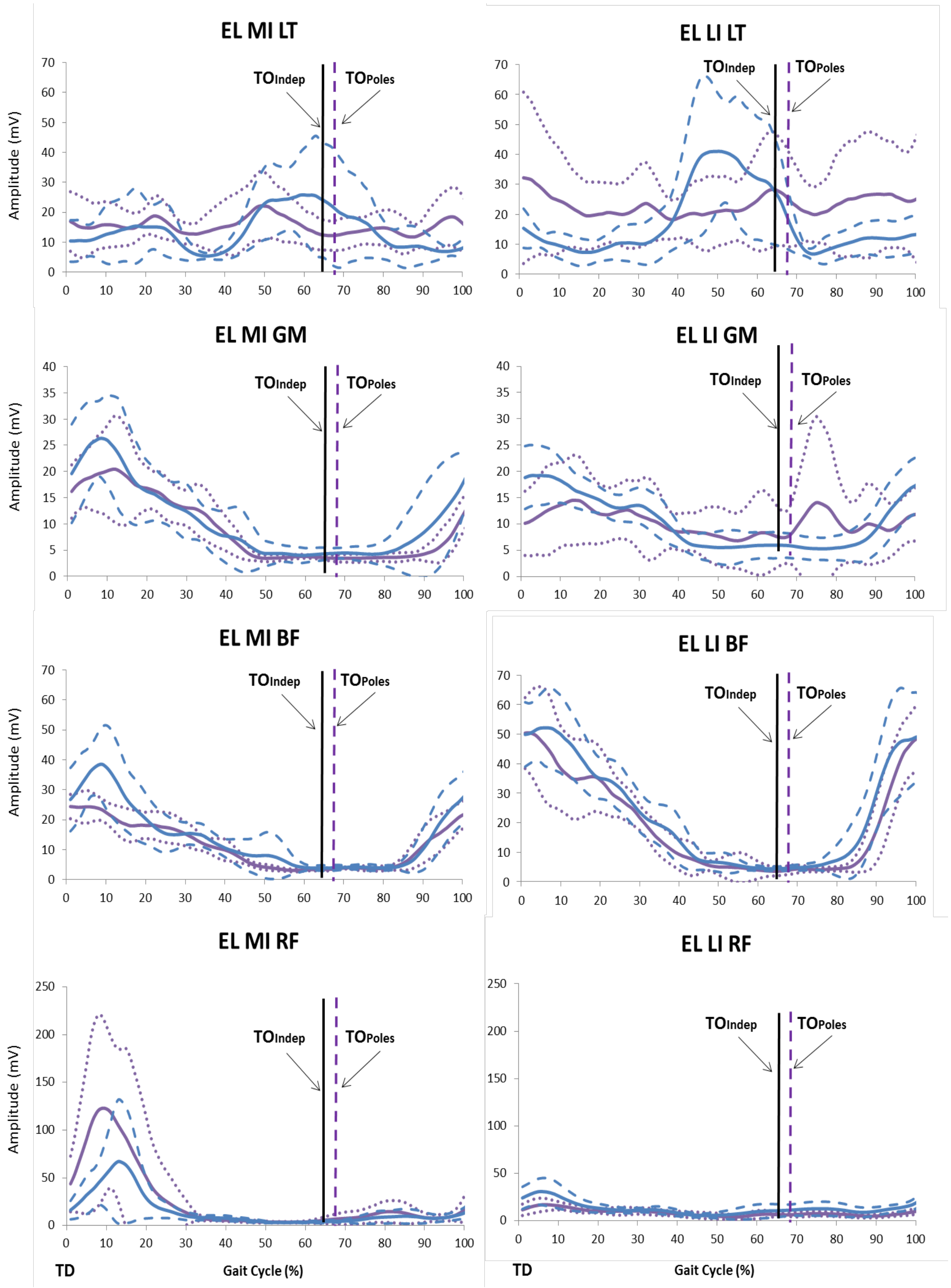


Figure B.1.3. (continued).

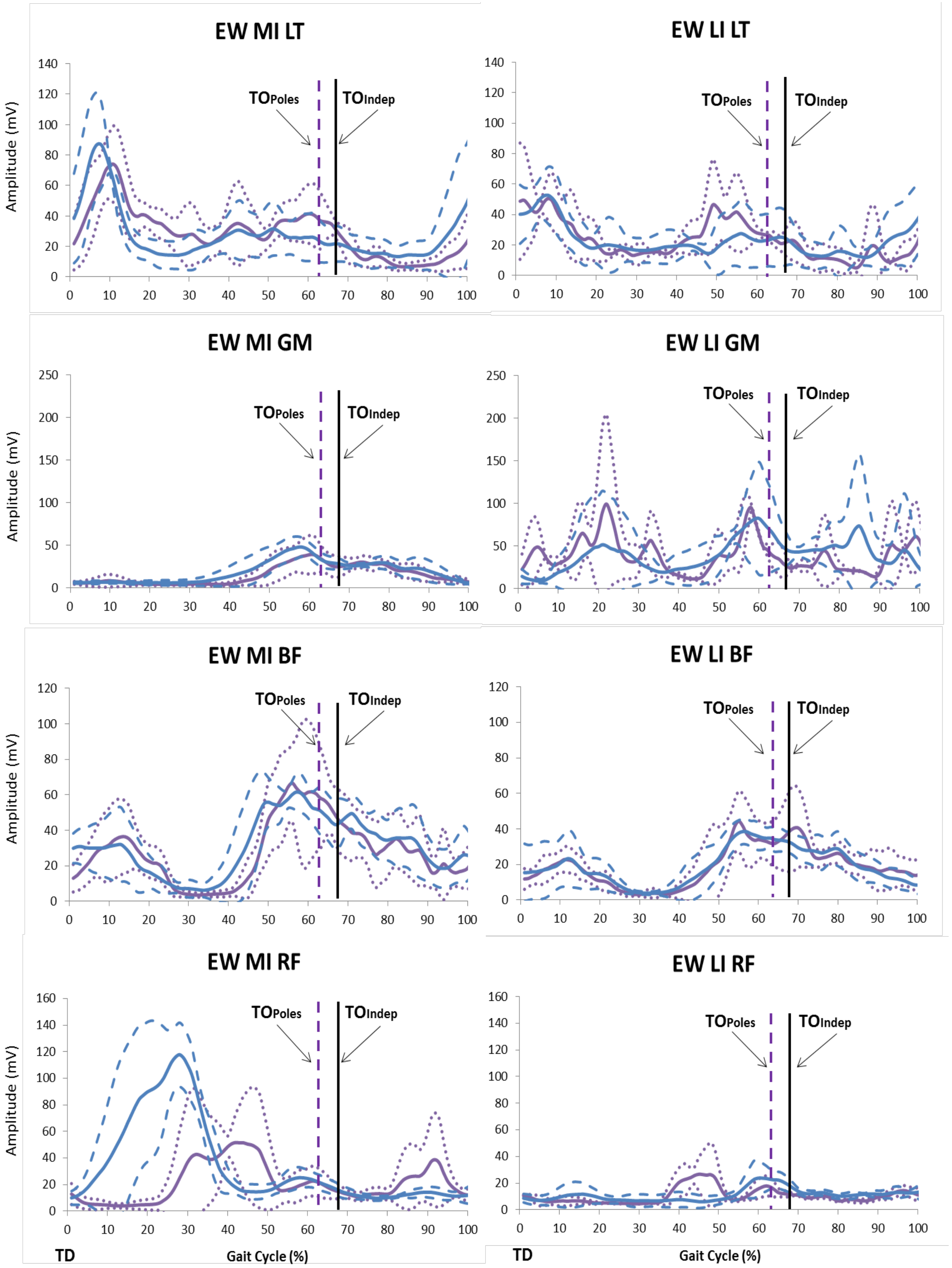


Figure B.1.3. (continued).

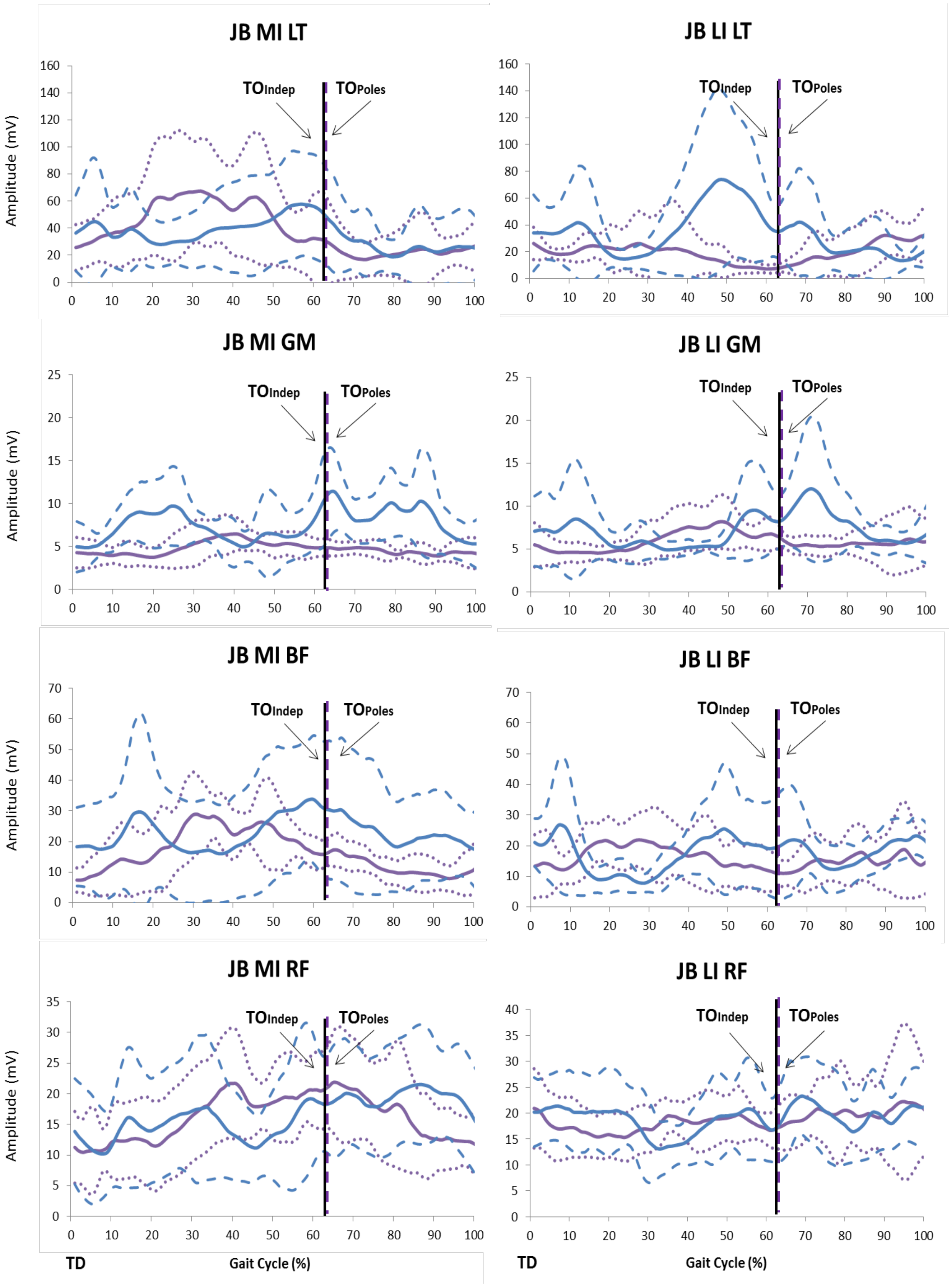


Figure B.1.3. (continued).

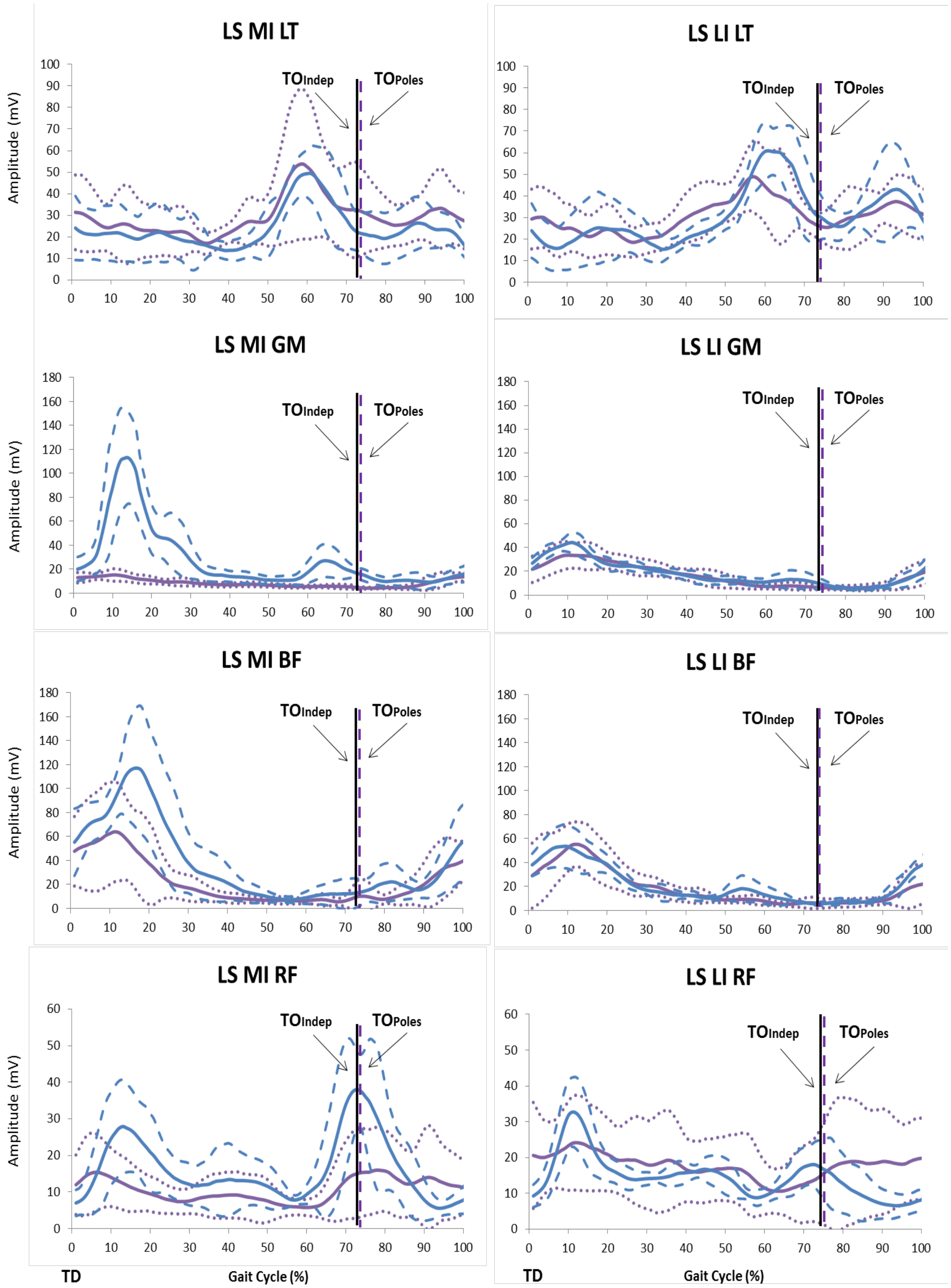


Figure B.1.3. (continued).

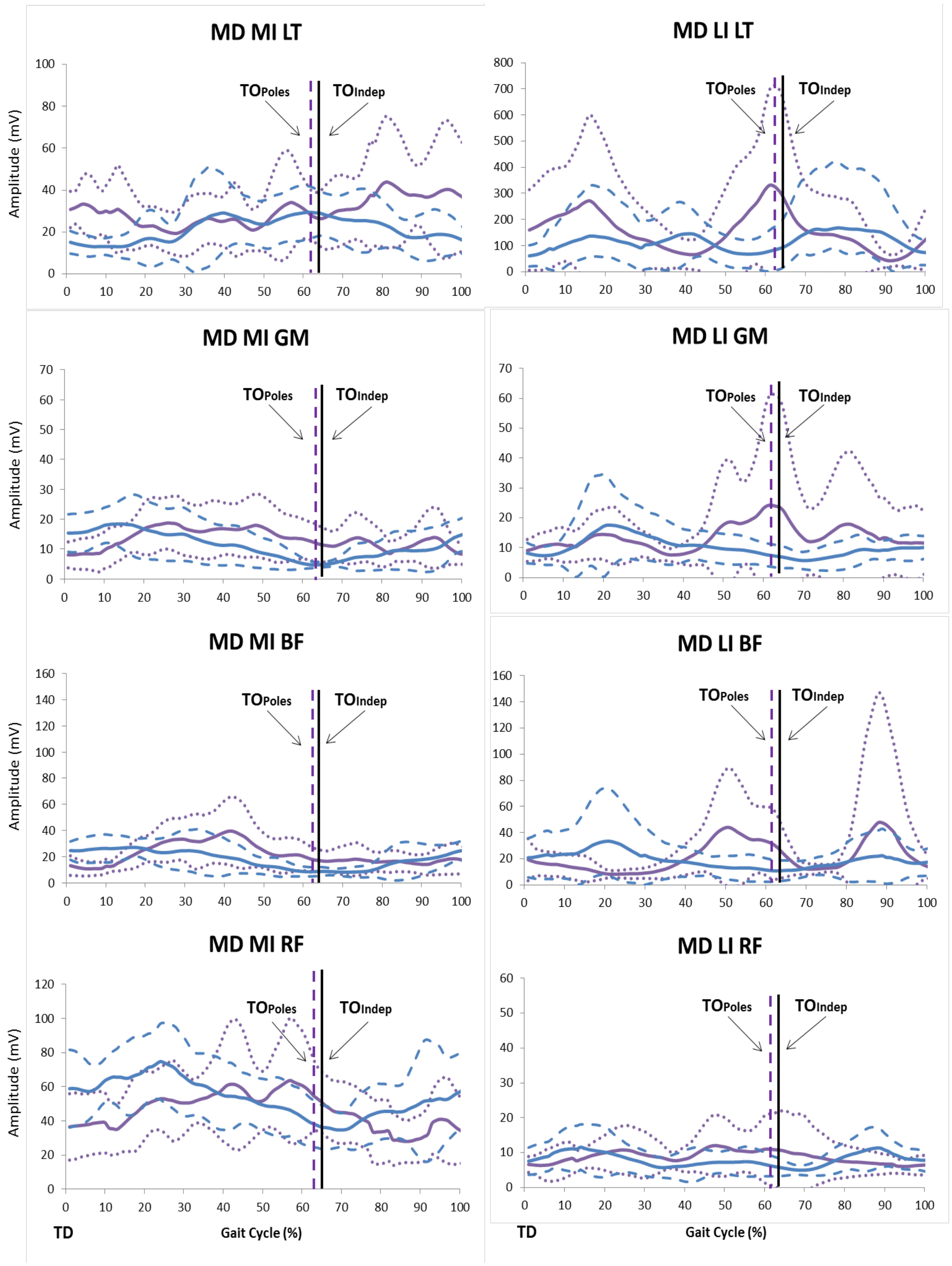


Figure B.1.3. (continued).

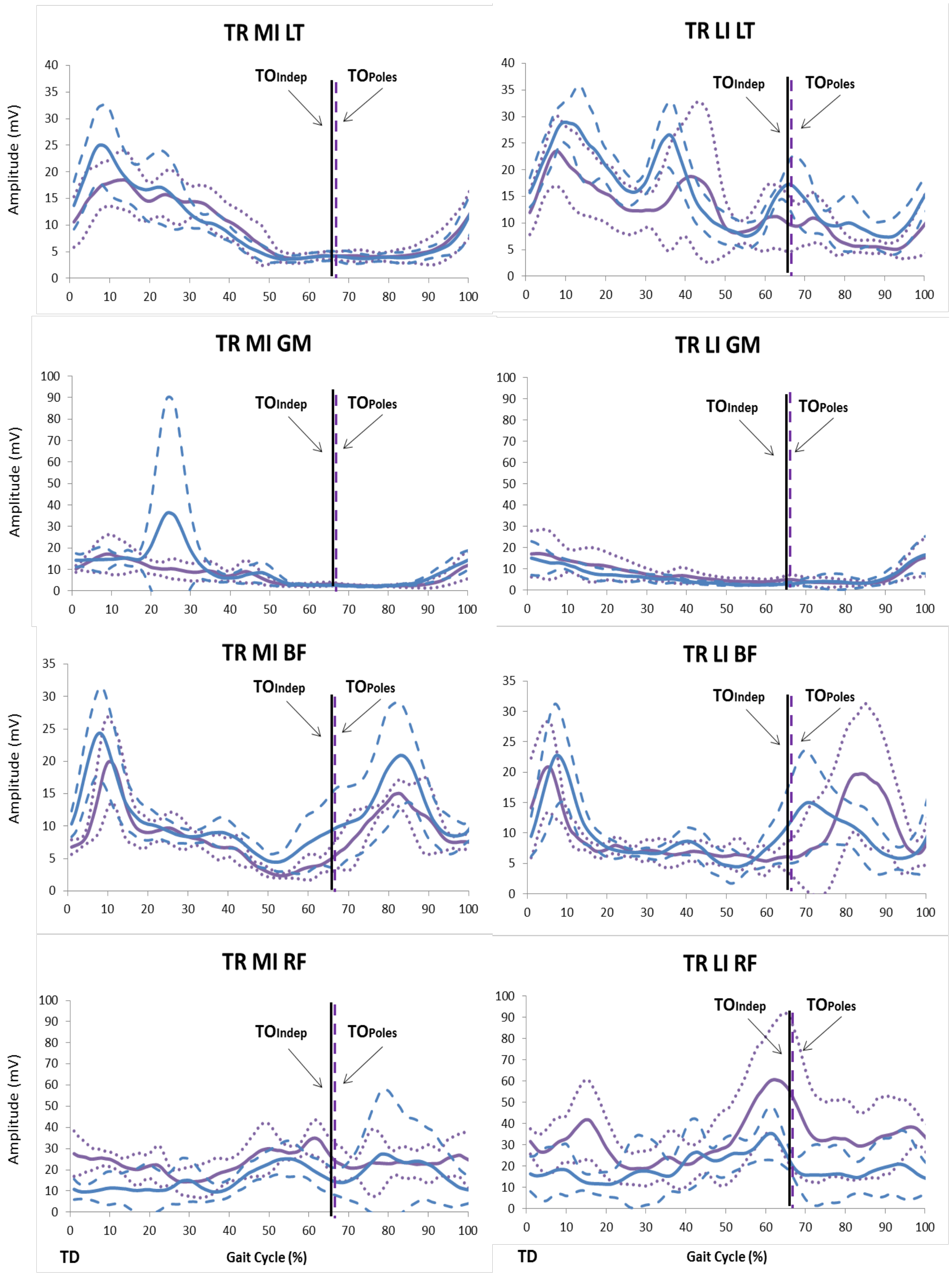


Figure B.1.3. (continued).

Appendix C.

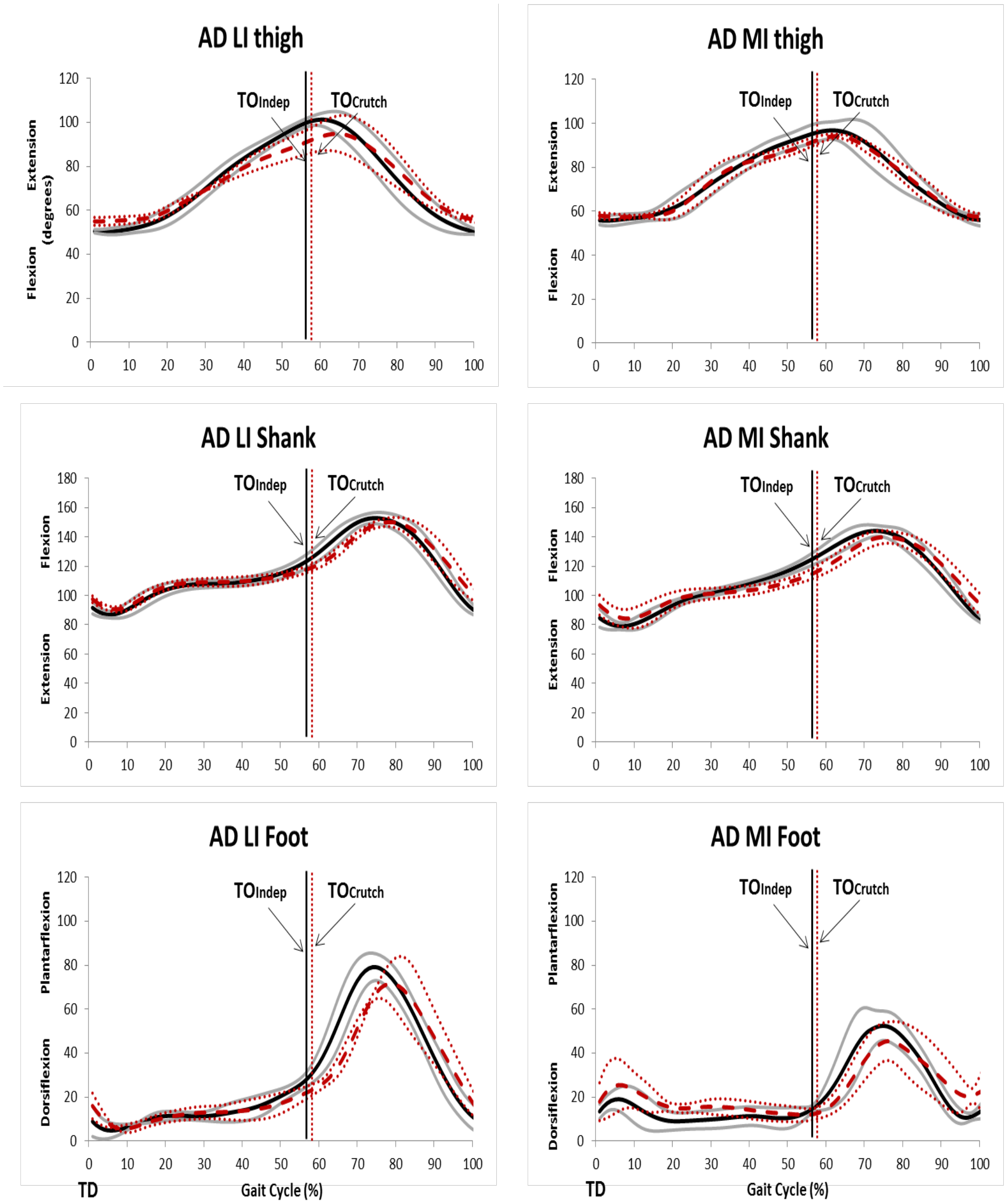


Figure C.1.1. Crutch segmental angle ensemble average traces with ±1 standard deviation envelopes for individual participants compared to independent walking.

(In chart title: first 2 initials=participant's initials; MI=More involved leg; LI=Less involved leg; TO=Toe off; TD=Touch down; ━ Independent average; — Independent standard deviation; – – Crutch average; ····· Crutch standard deviation.)

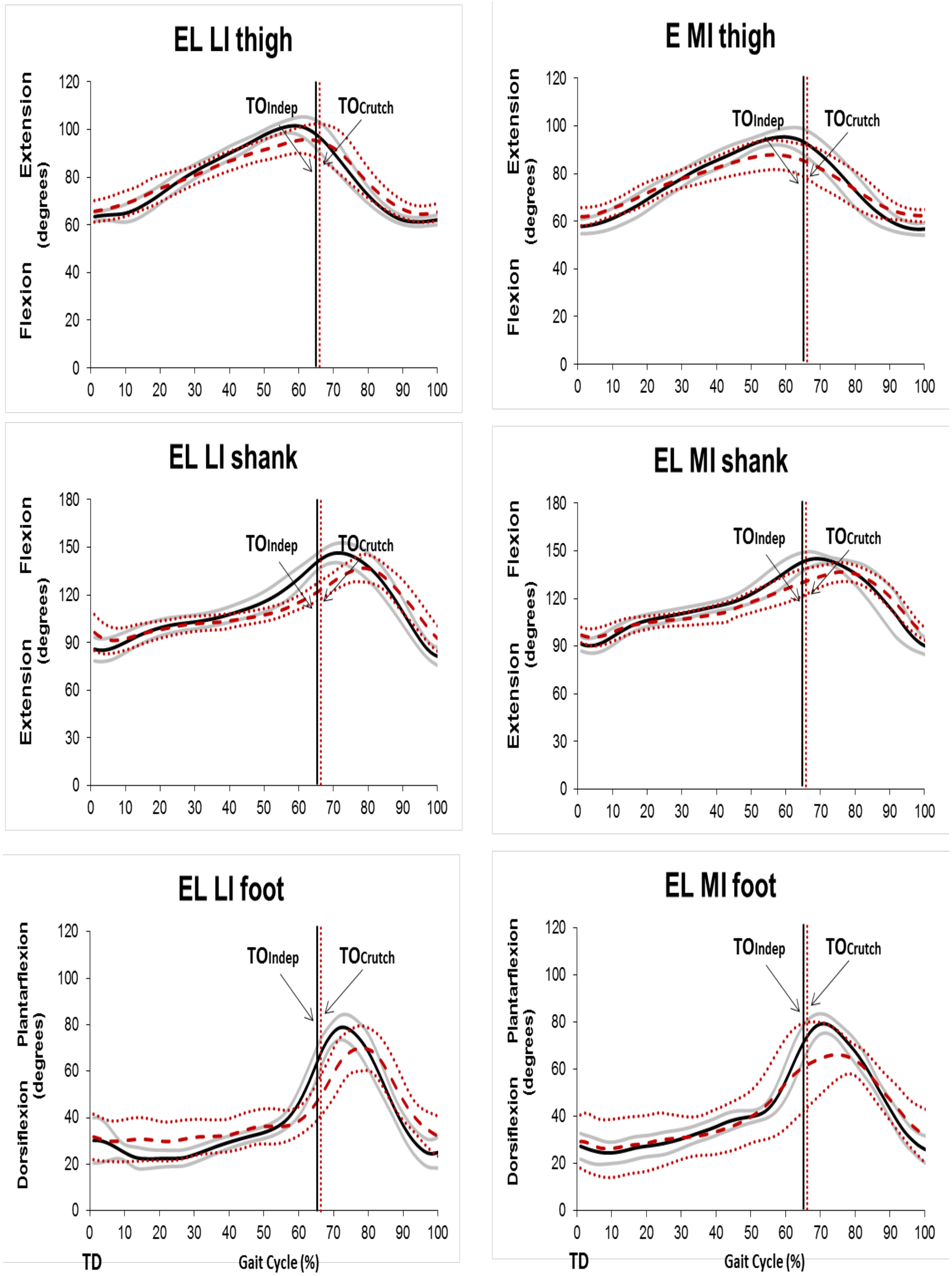


Figure C.1.1. (continued).

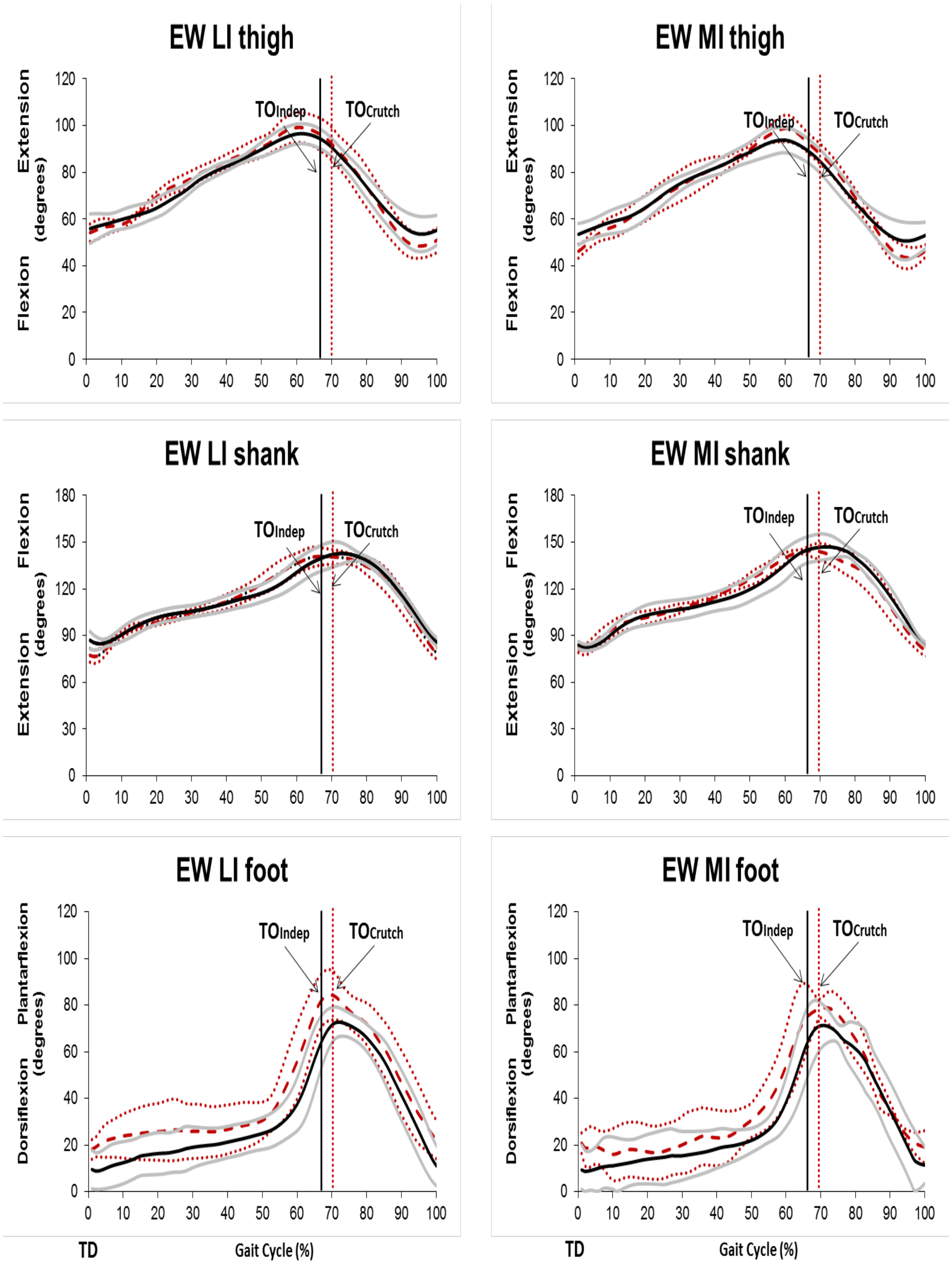


Figure C.1.1. (continued).

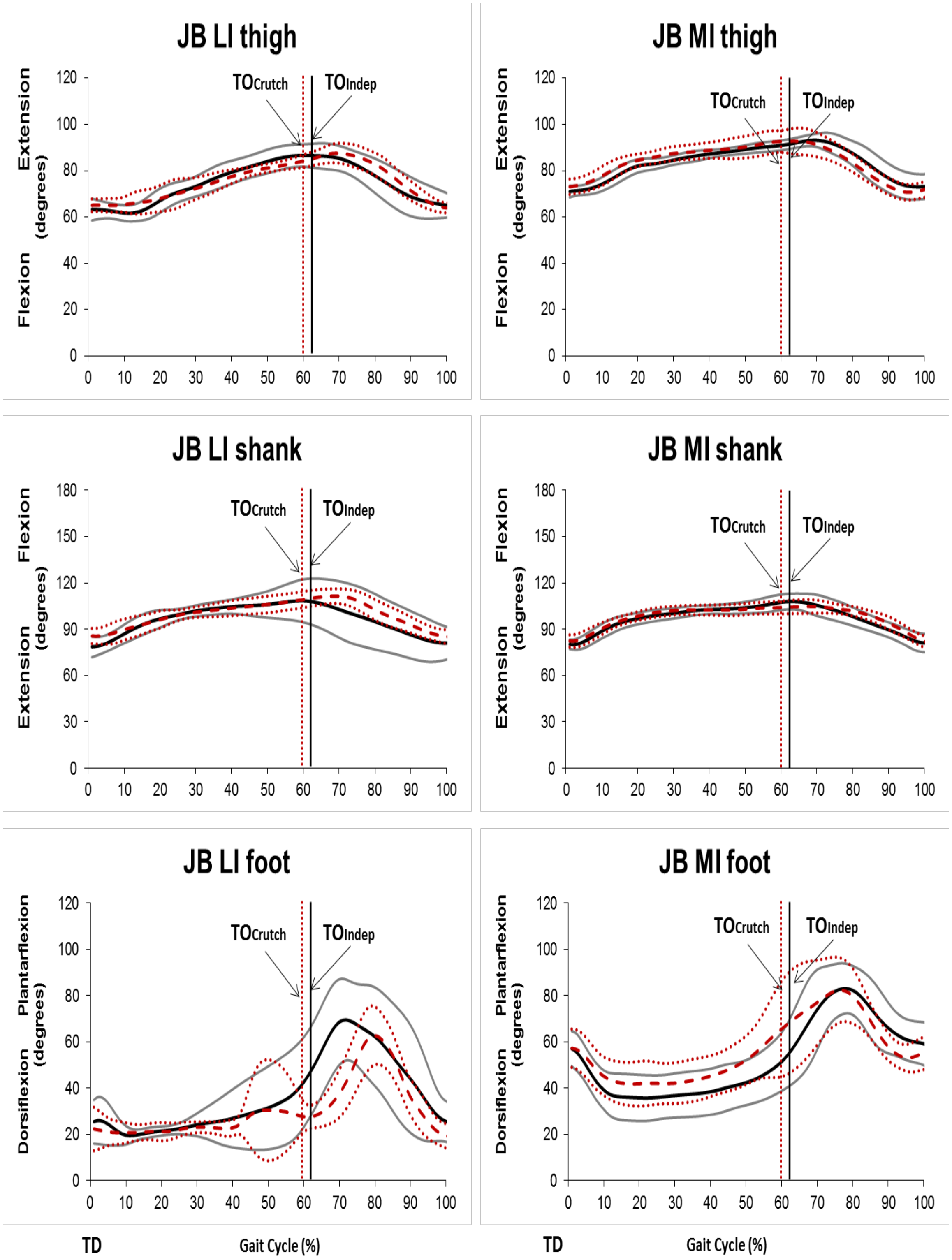


Figure C.1.1. (continued).

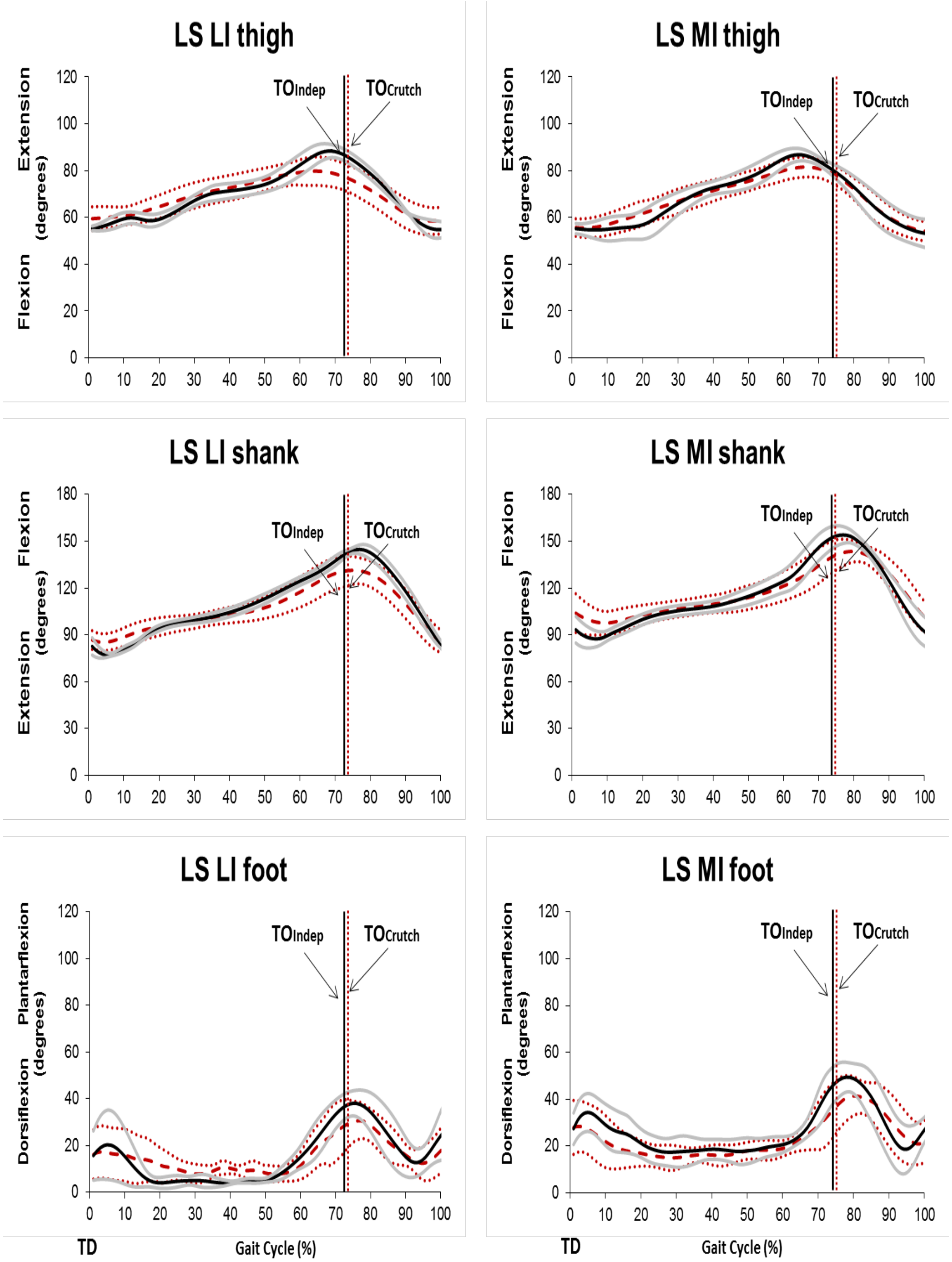


Figure C.1.1. (continued).

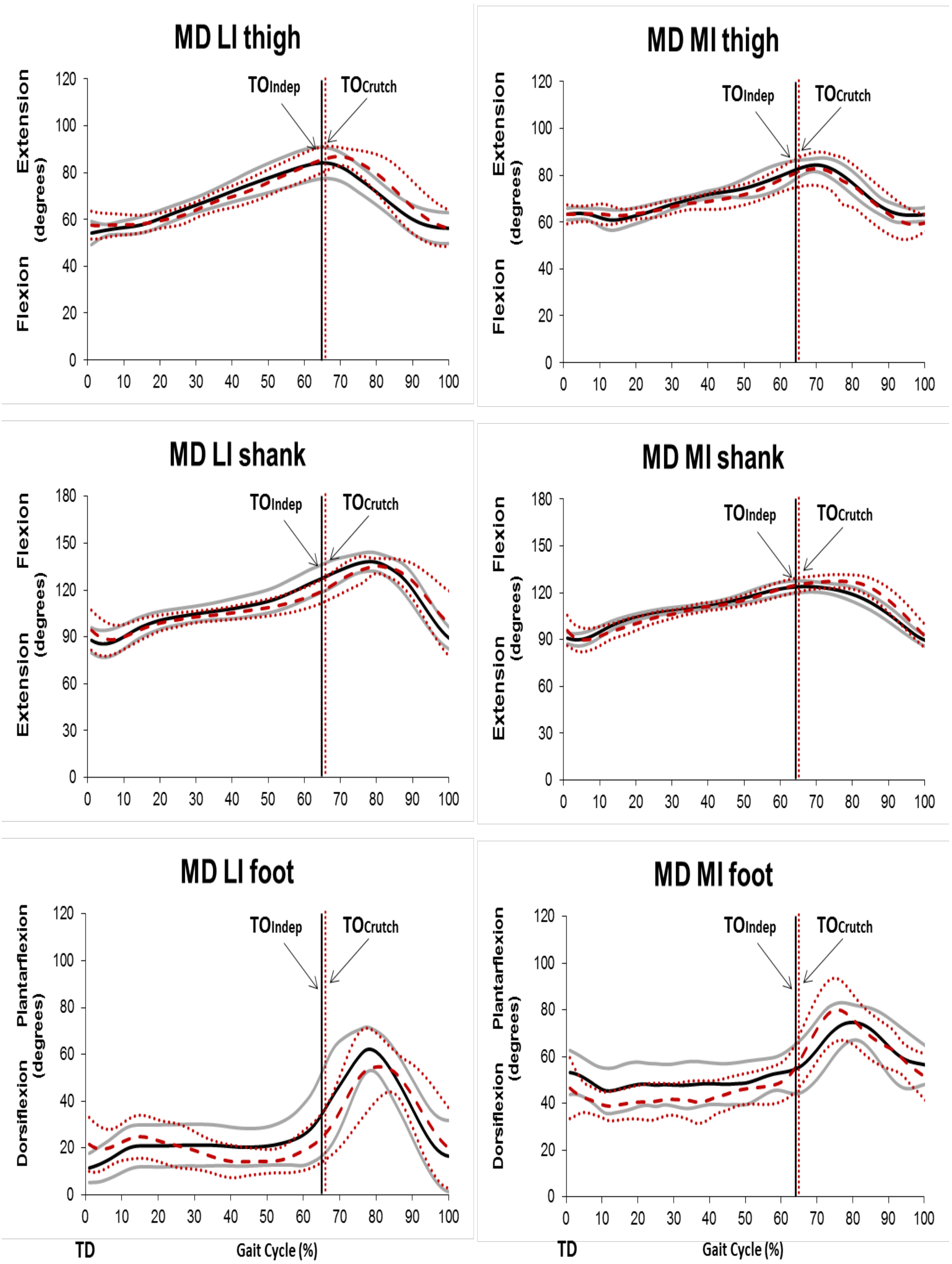


Figure C.1.1. (continued).

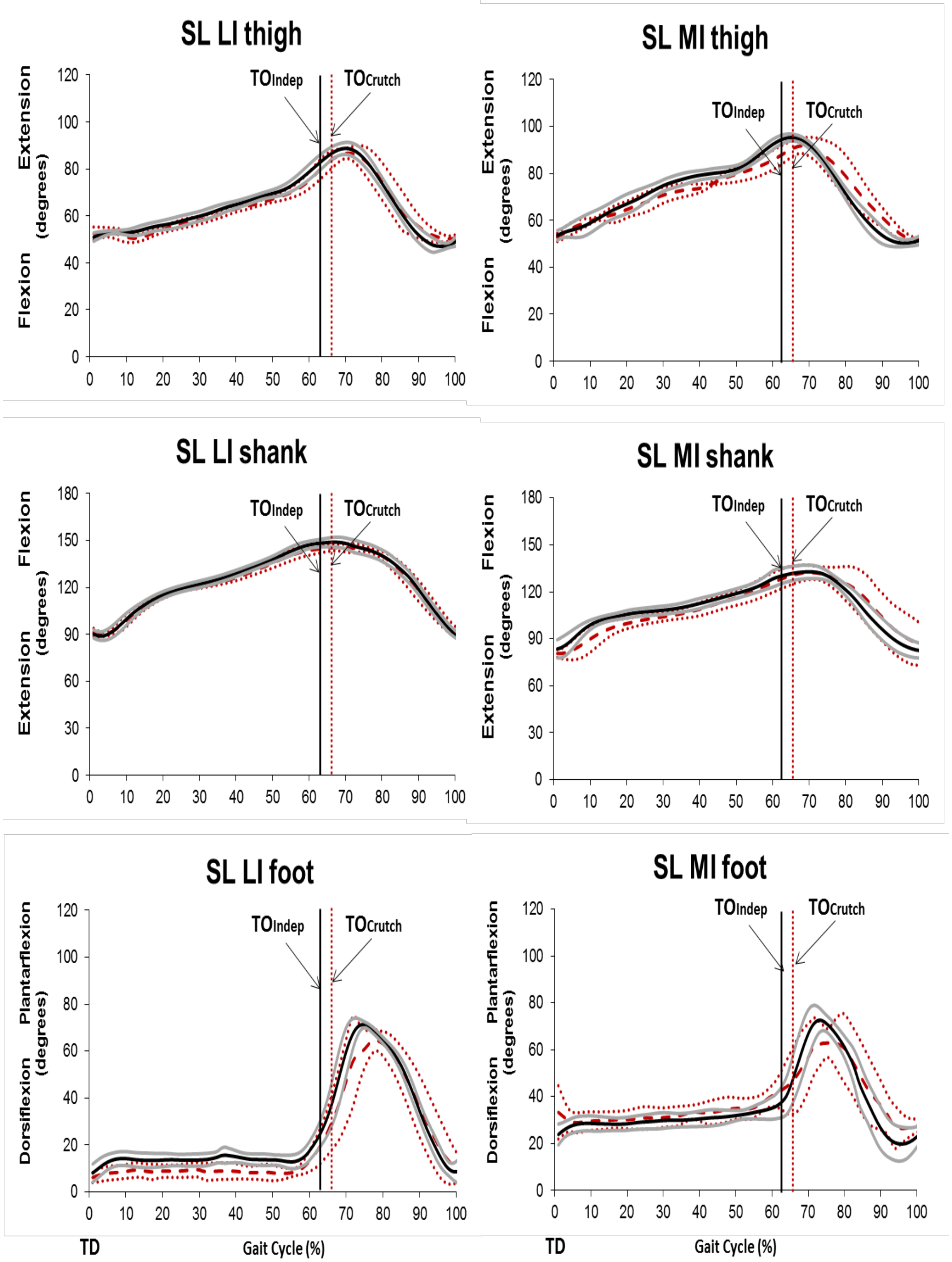


Figure C.1.1. (continued).

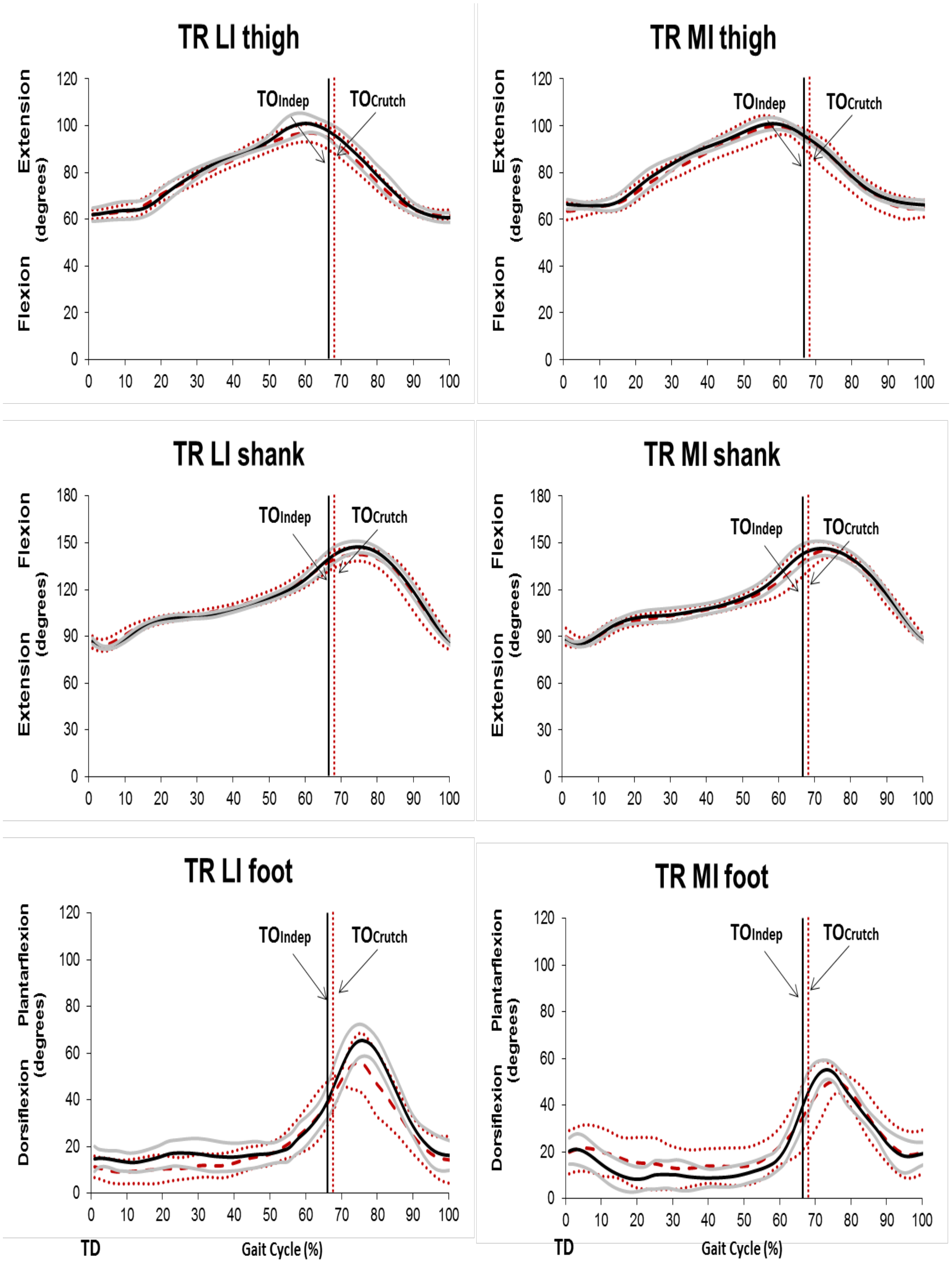


Figure C.1.1. (continued).

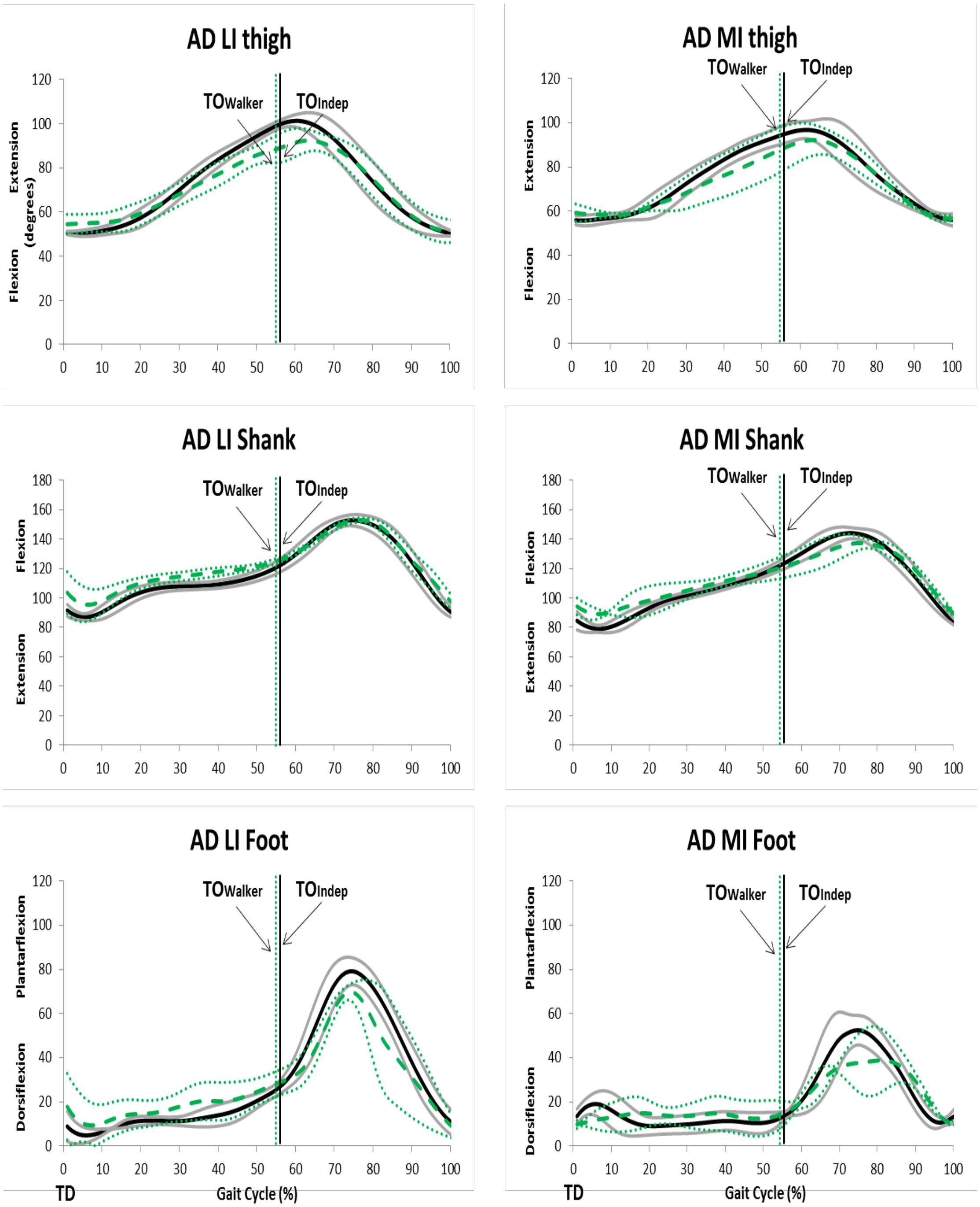


Figure C.1.2. Walker segmental angle ensemble average traces with ±1 standard deviation envelopes for individual par ticipants compared to independent walking. (In chart title: first 2 initials=participant's initials; MI=More involved leg; LI=Less involved leg; TO=Toe off; TD=Touch down; — Independent average; — Independent standard deviation; – –Walker average; ·····Walker standard deviation.)

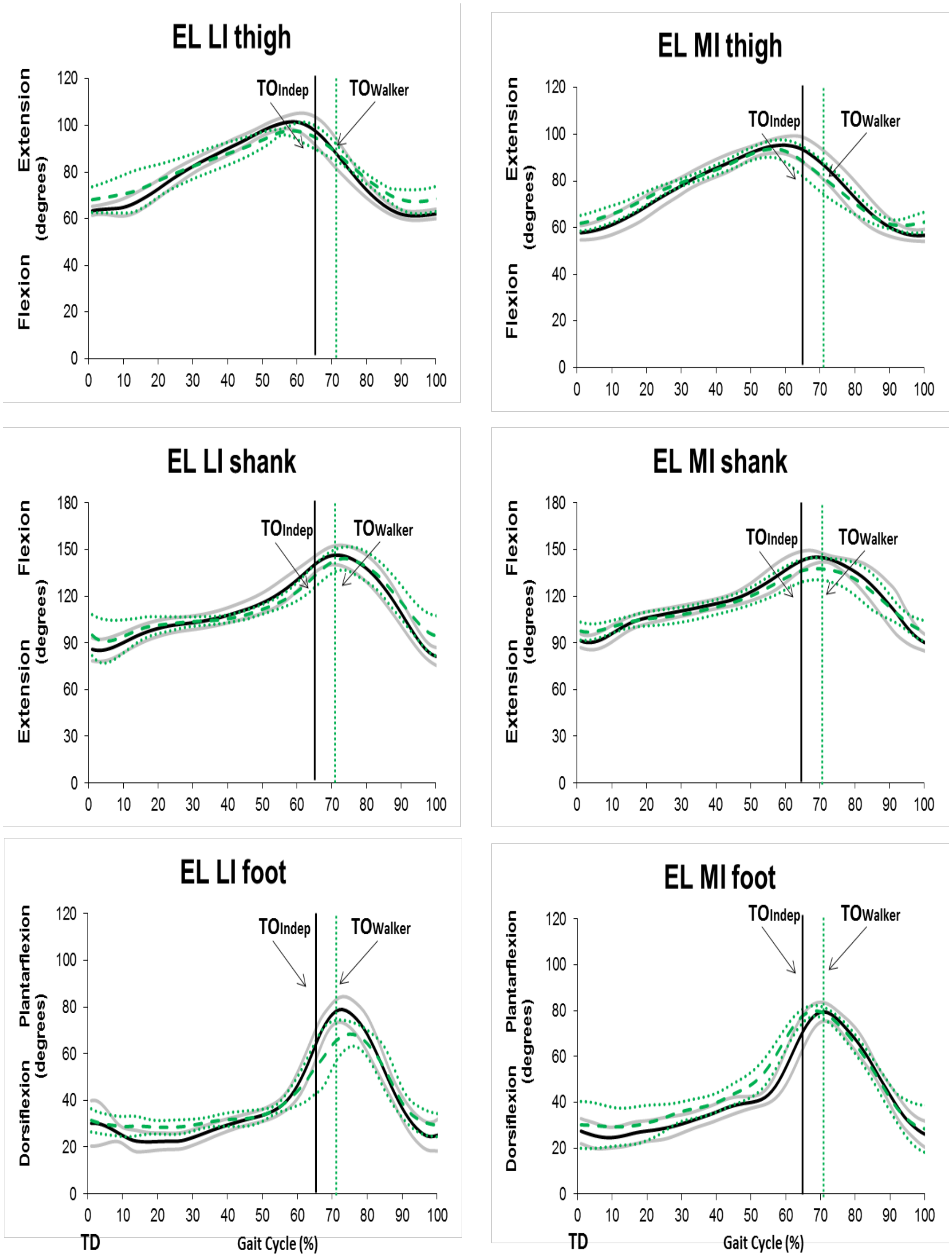


Figure C.1.2. (continued).

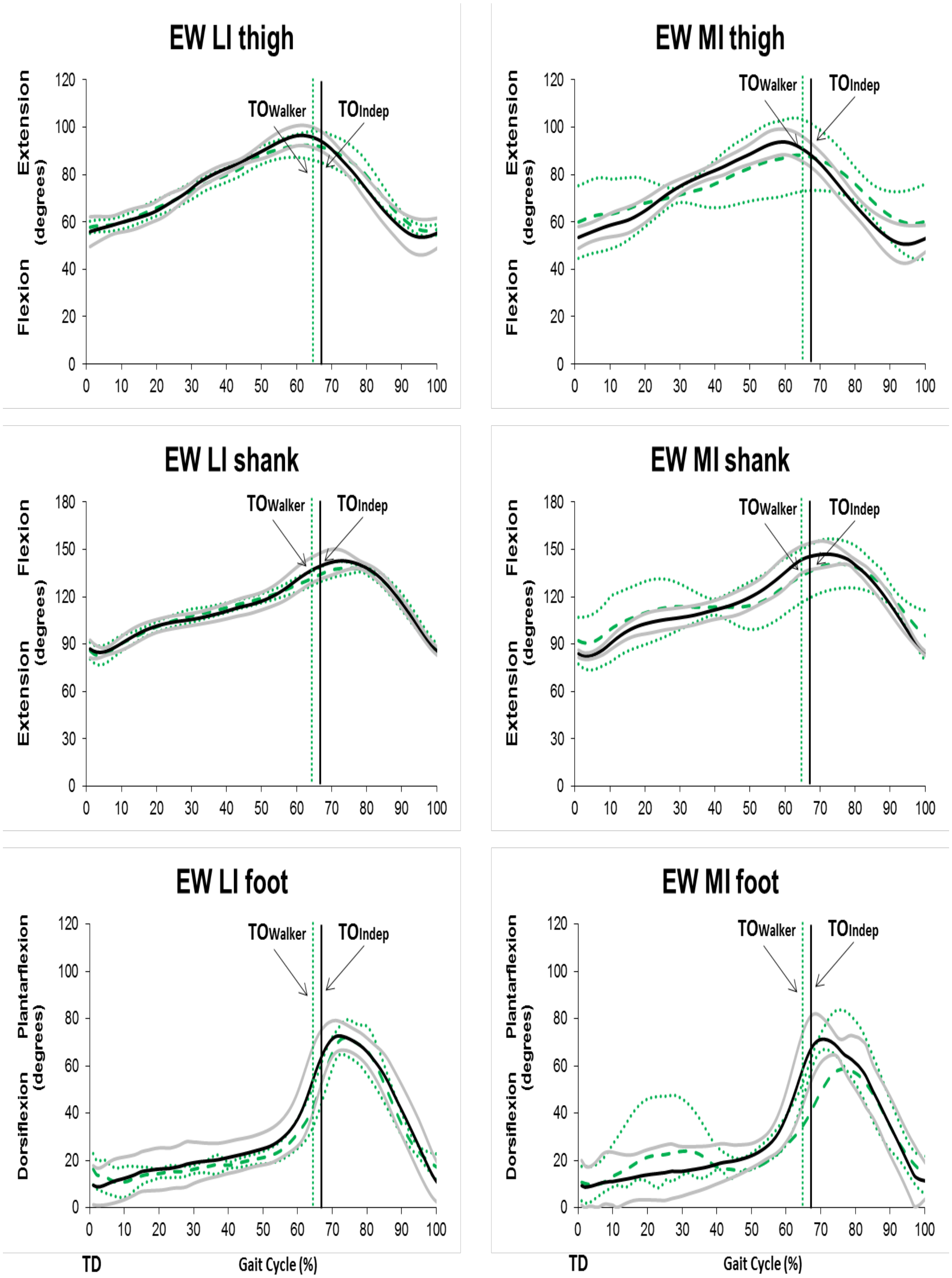


Figure C.1.2. (continued).

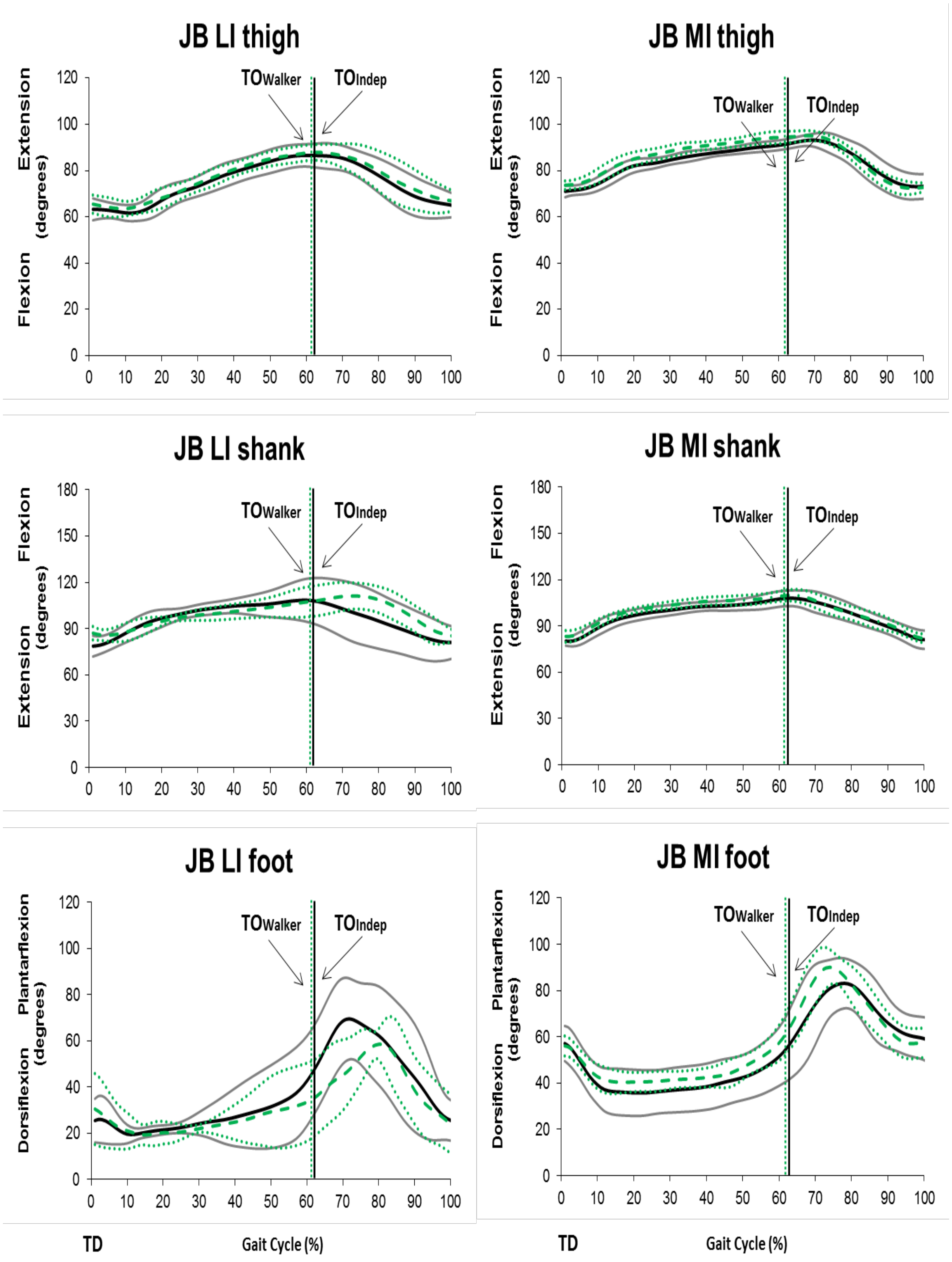


Figure C.1.2. (continued).

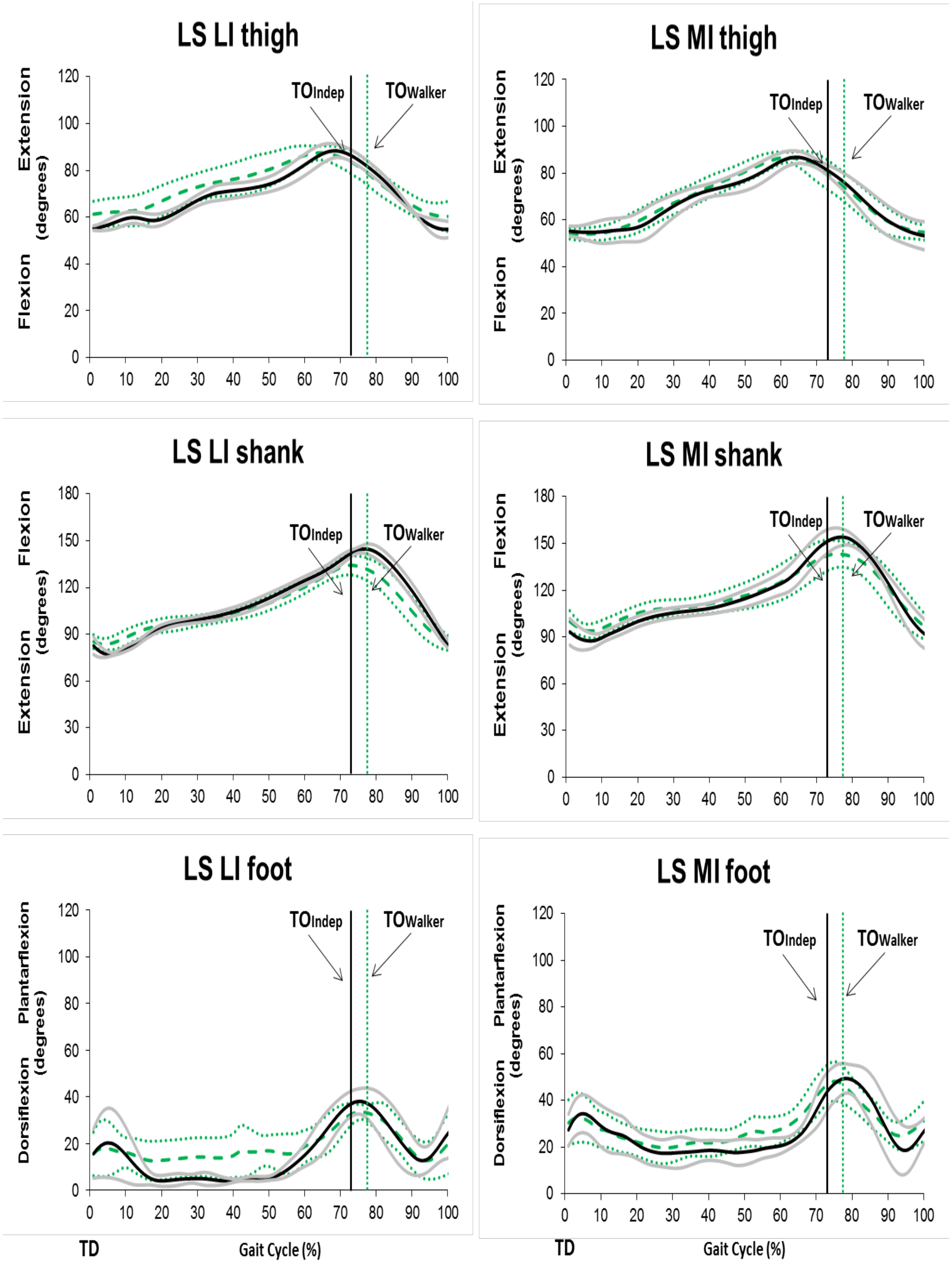


Figure C.1.2. (continued).

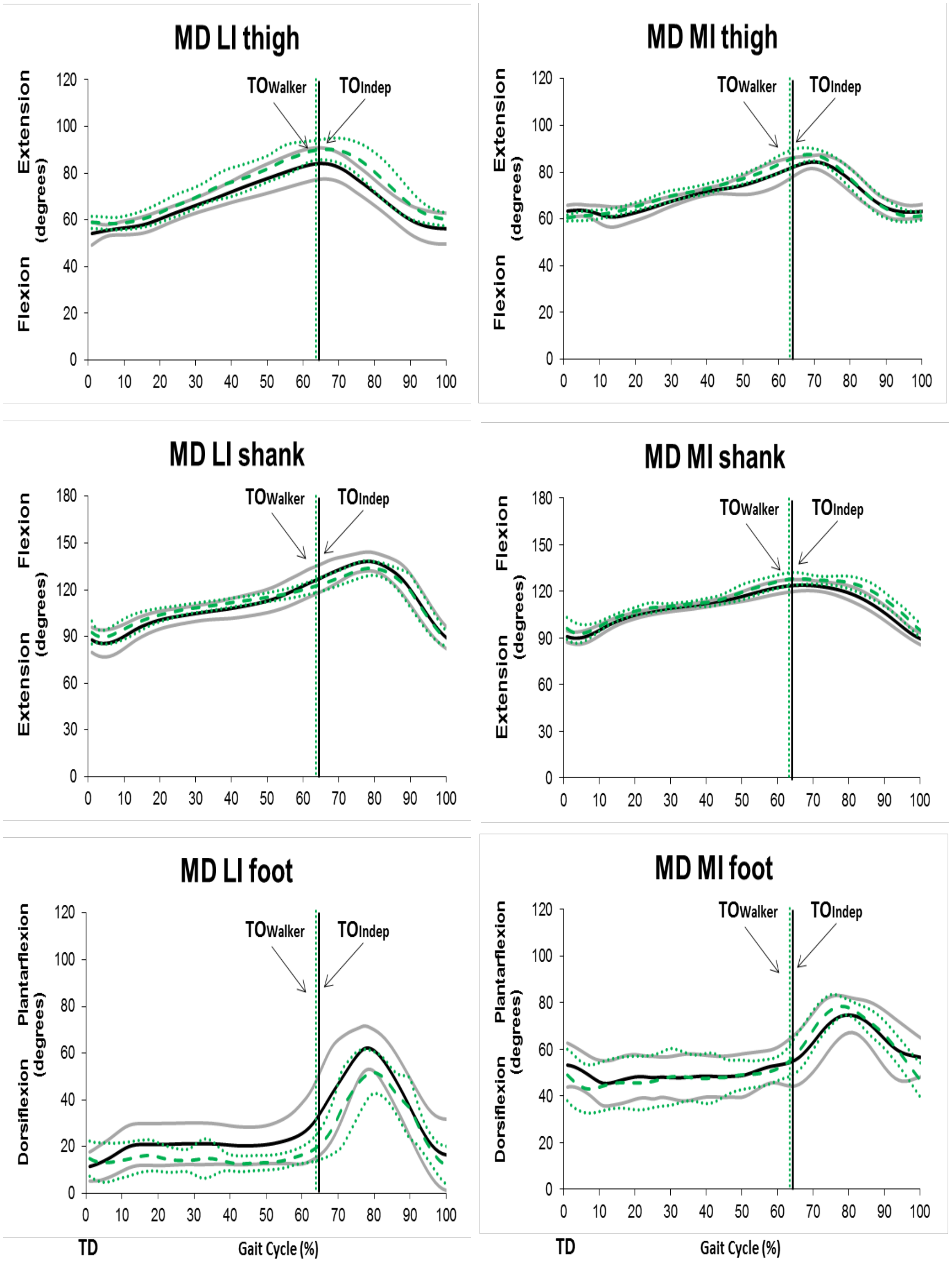


Figure C.1.2. (continued).

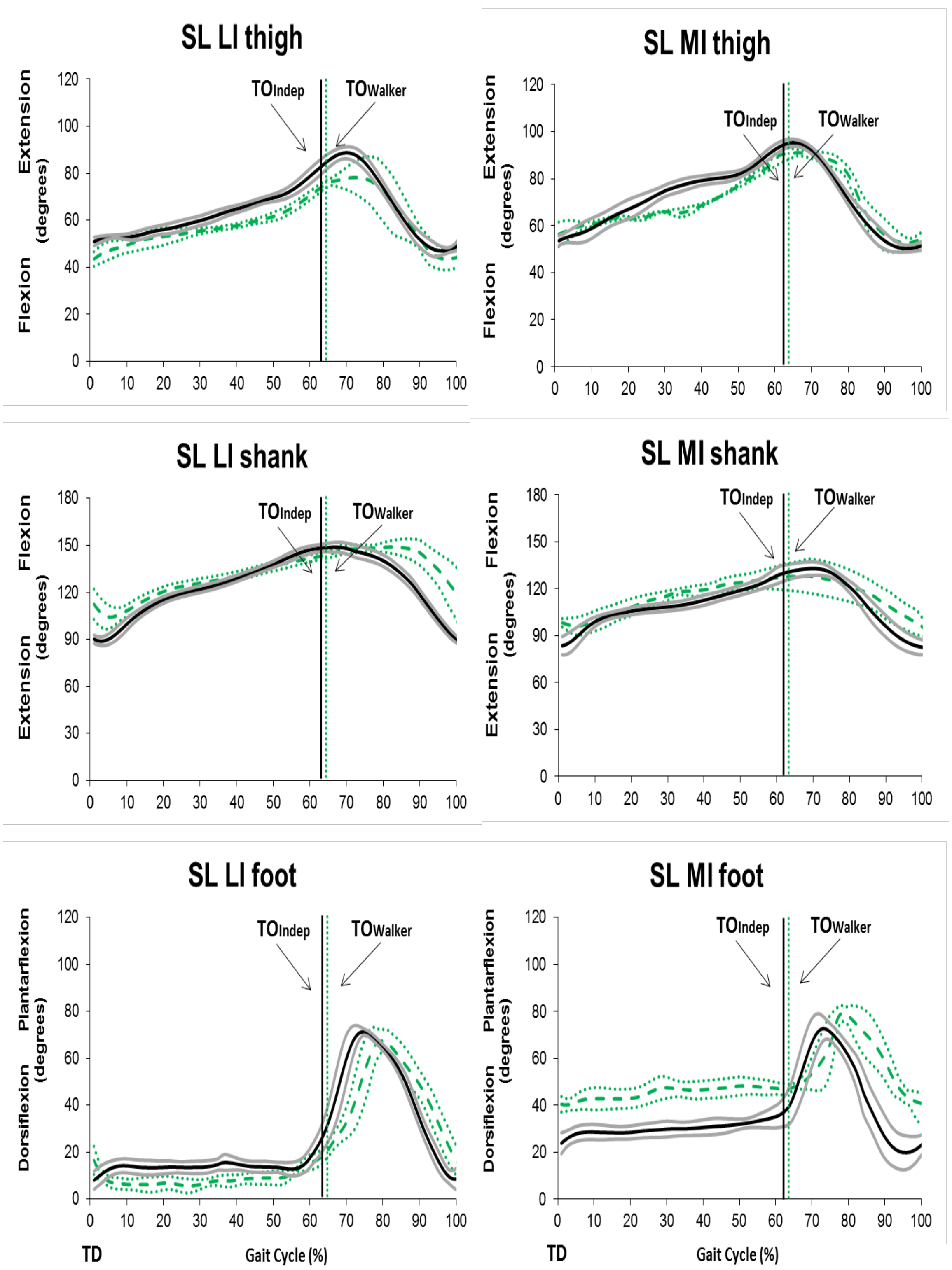


Figure C.1.2. (continued).

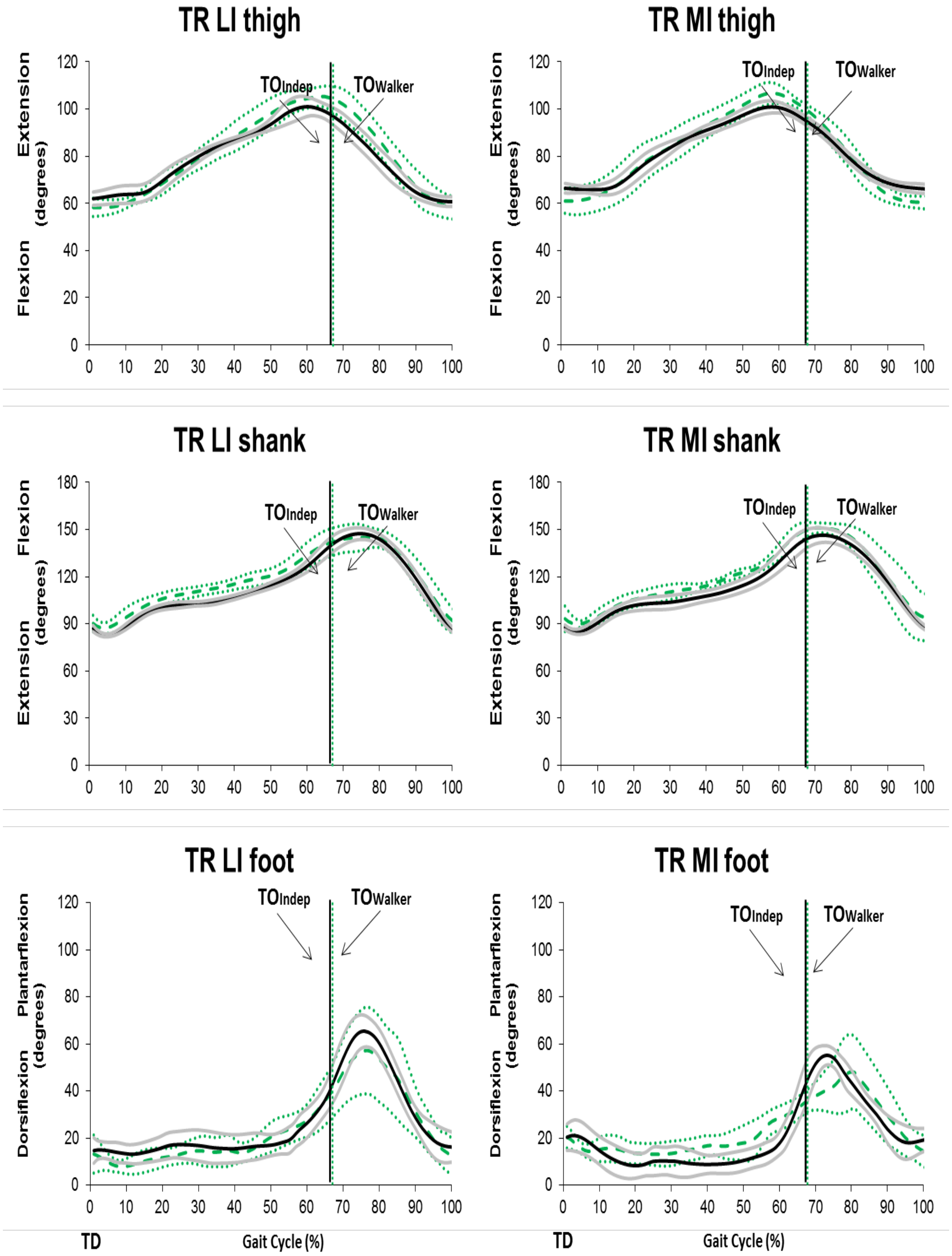


Figure C.1.2. (continued).

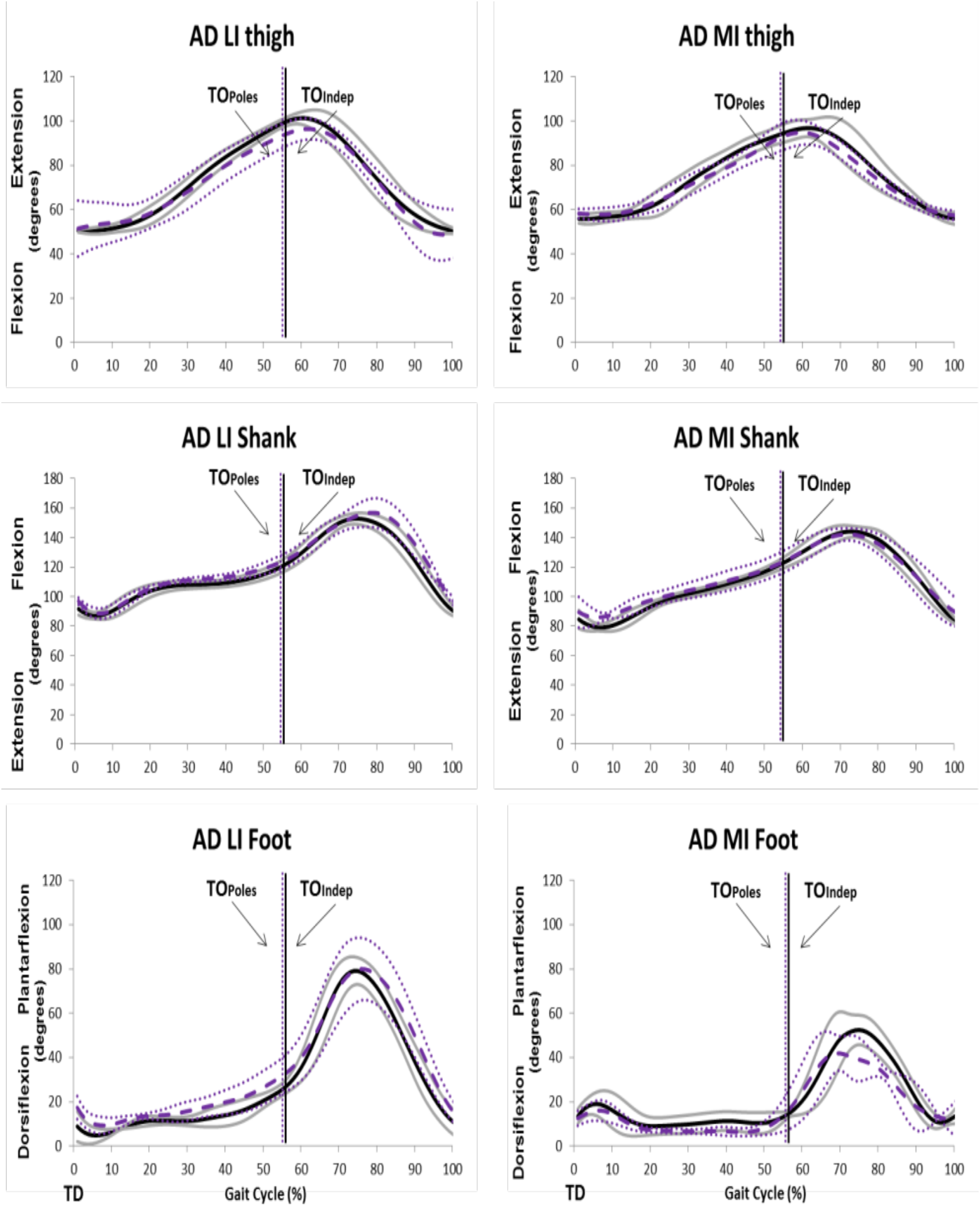


Figure C.1.3. Pole segmental angle ensemble average traces with ±1 standard deviation envelopes for individual participants compared to independent walking. (In chart title: first 2 initials=part icipant's initials; MI=More involved leg; LI=Less involved leg; TO=Toe off; TD=Touch down; — Independent average; — Independent standard deviation; – – Pole average; ····· Pole standard deviation.)

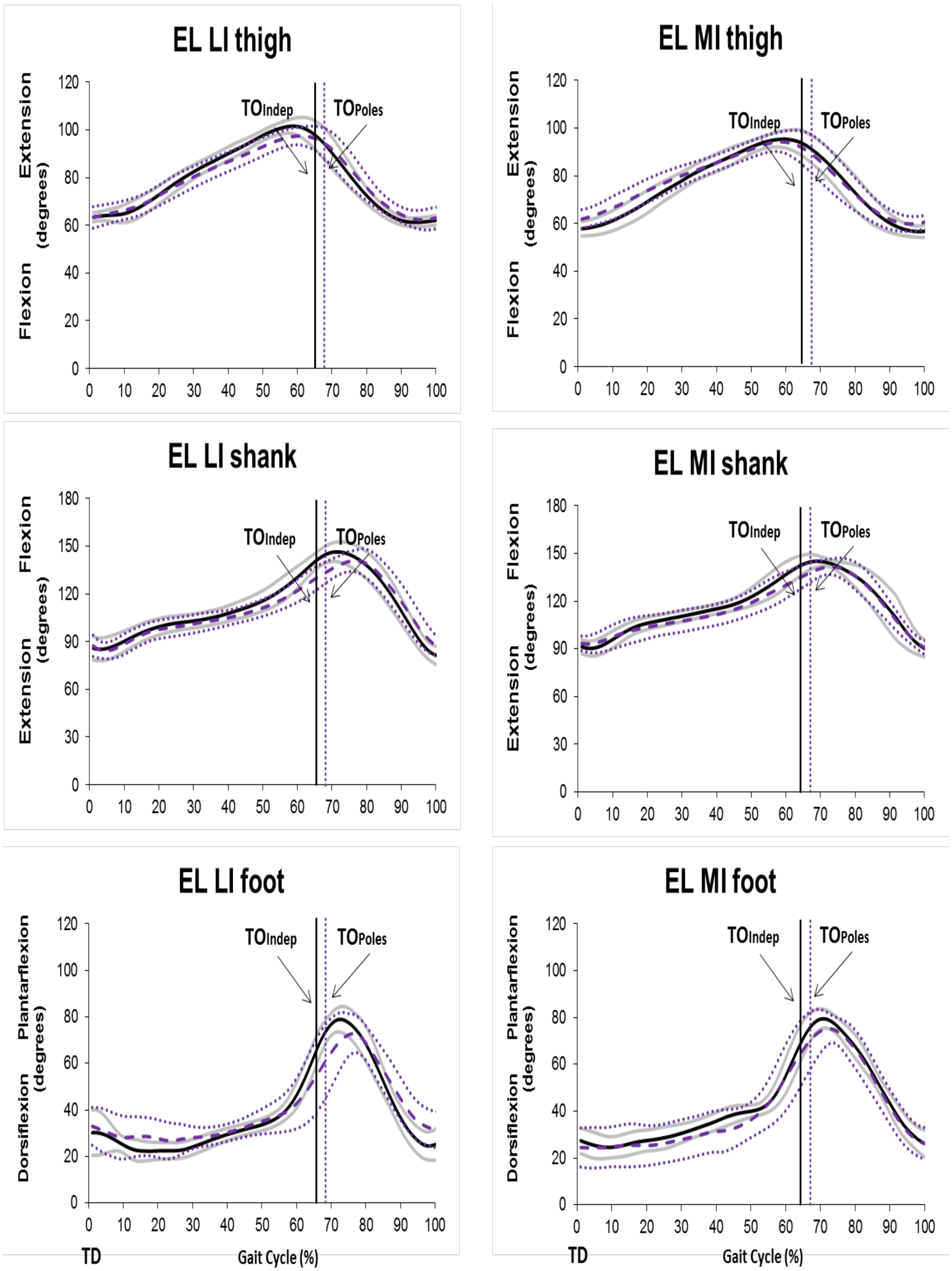


Figure C.1.3. (continued).

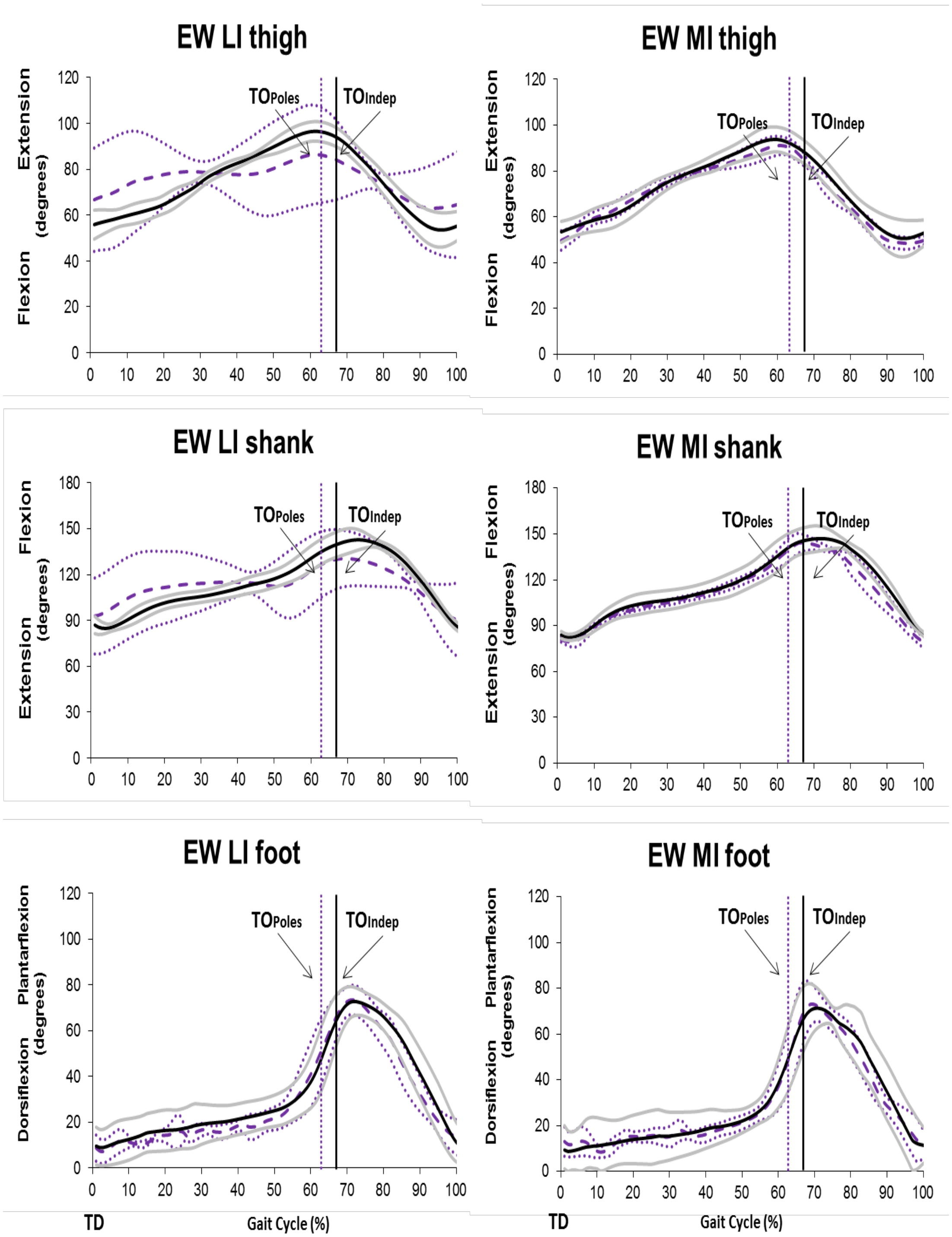


Figure C.1.3. (continued).

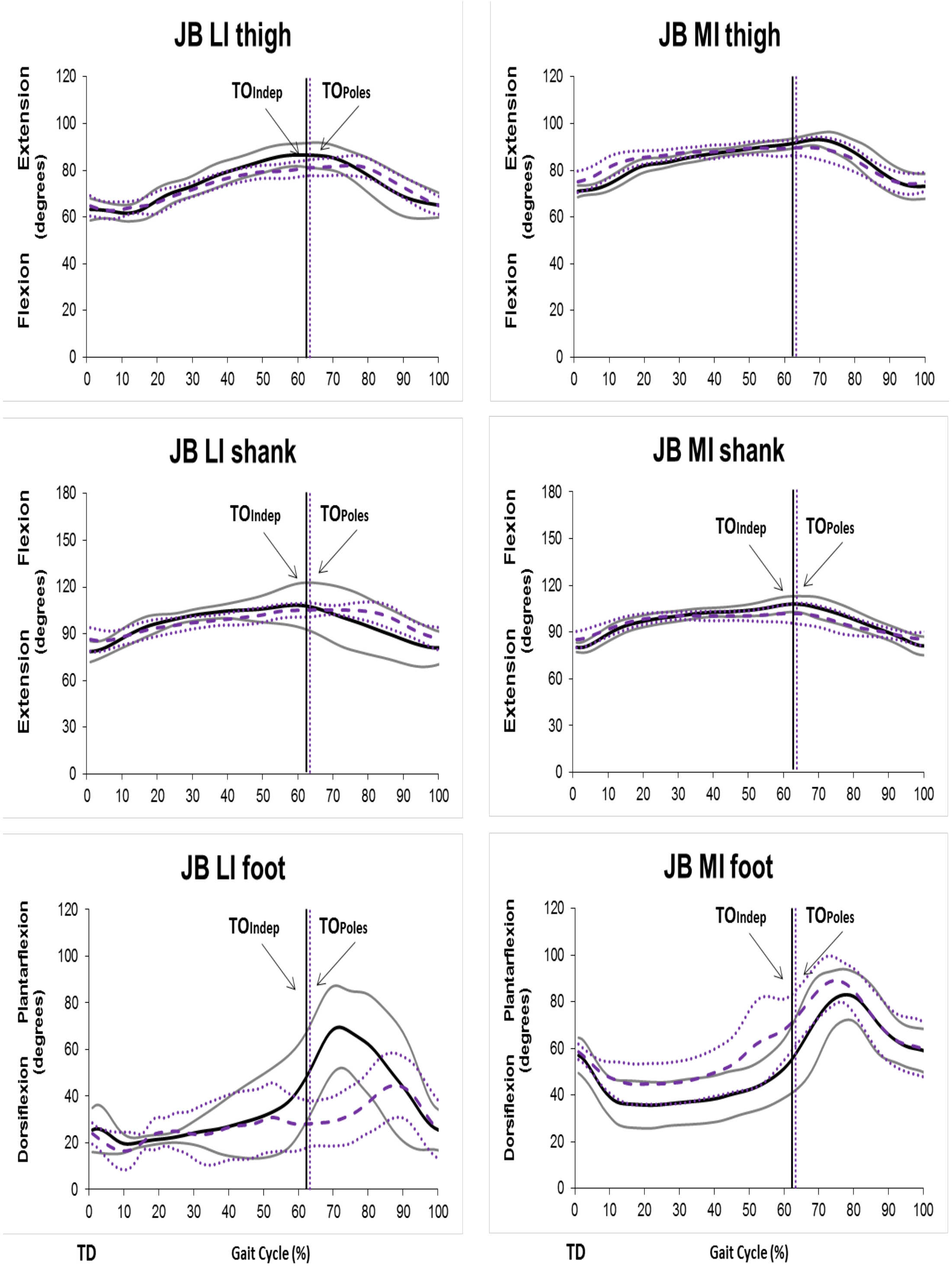


Figure C.1.3. (continued).

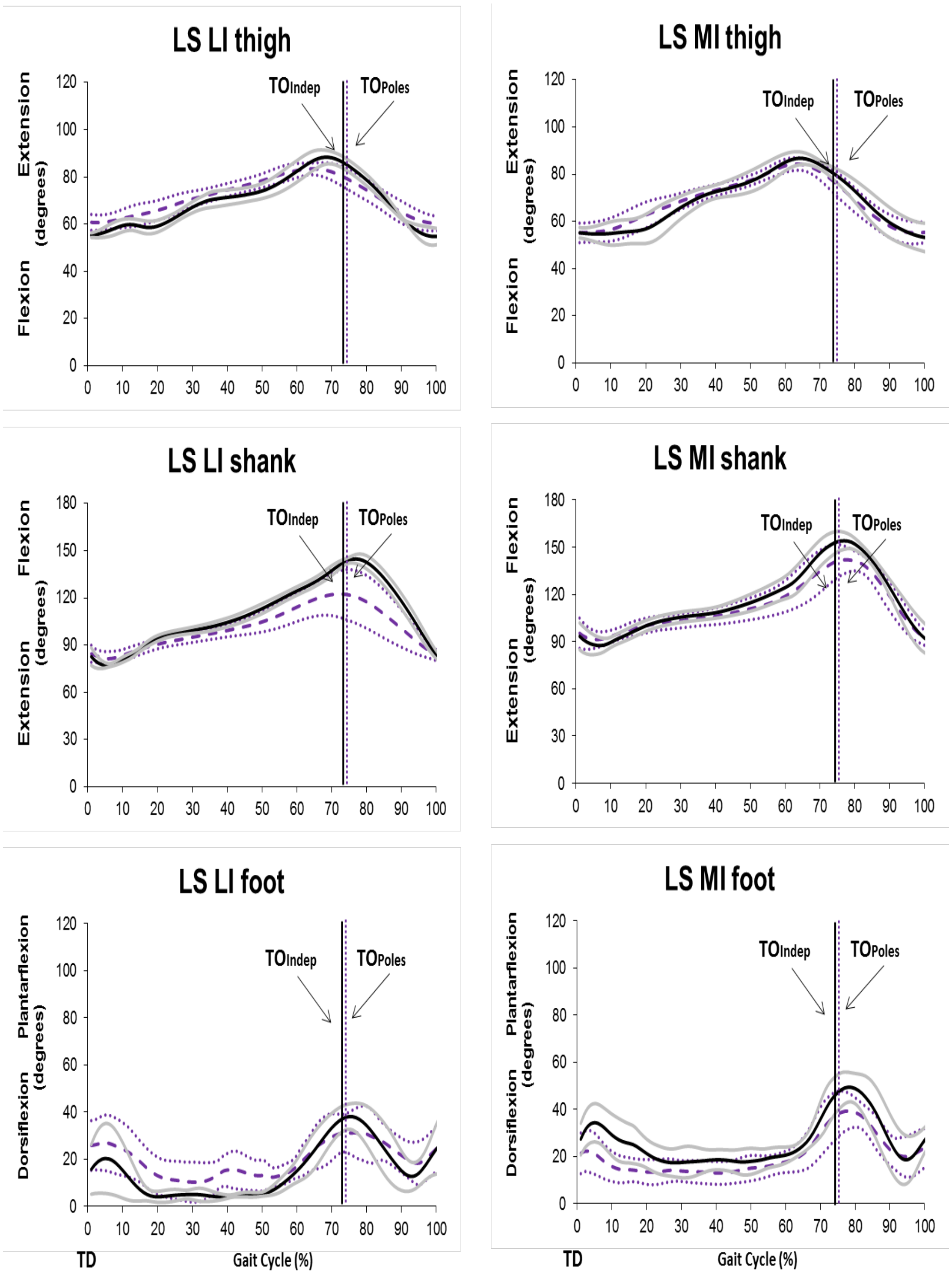
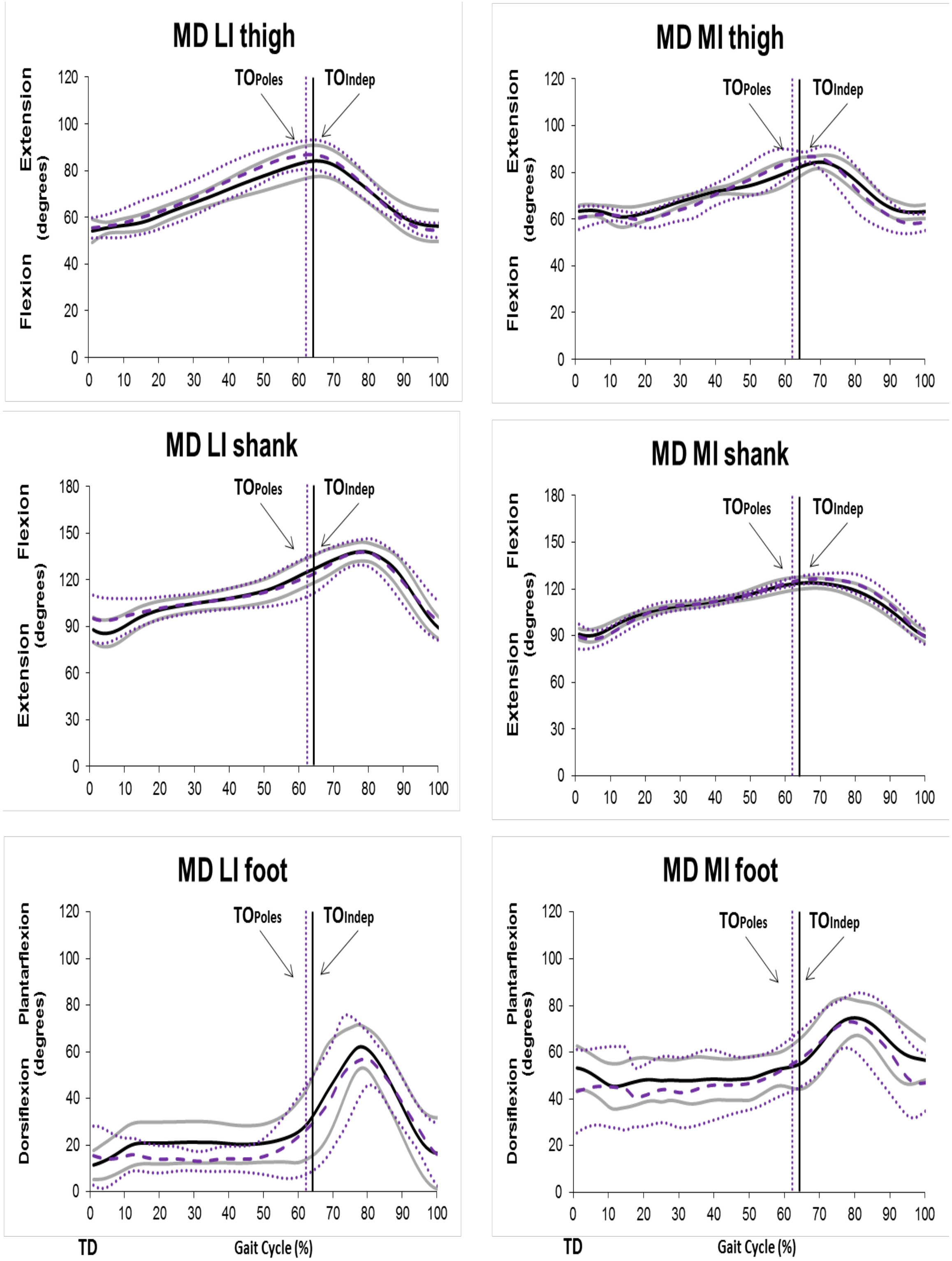


Figure C.1.3. (continued).



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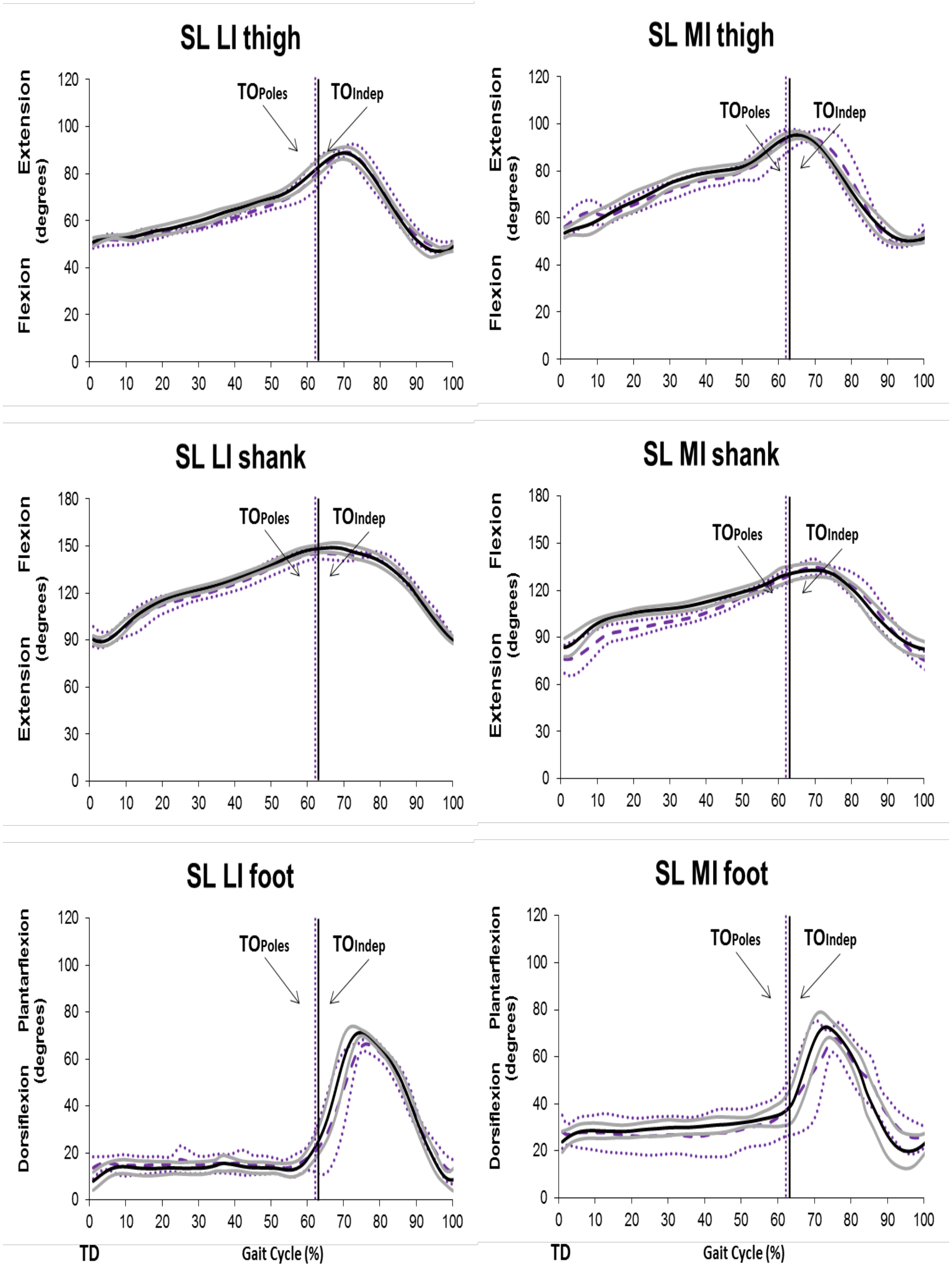


Figure C.1.3. (continued).

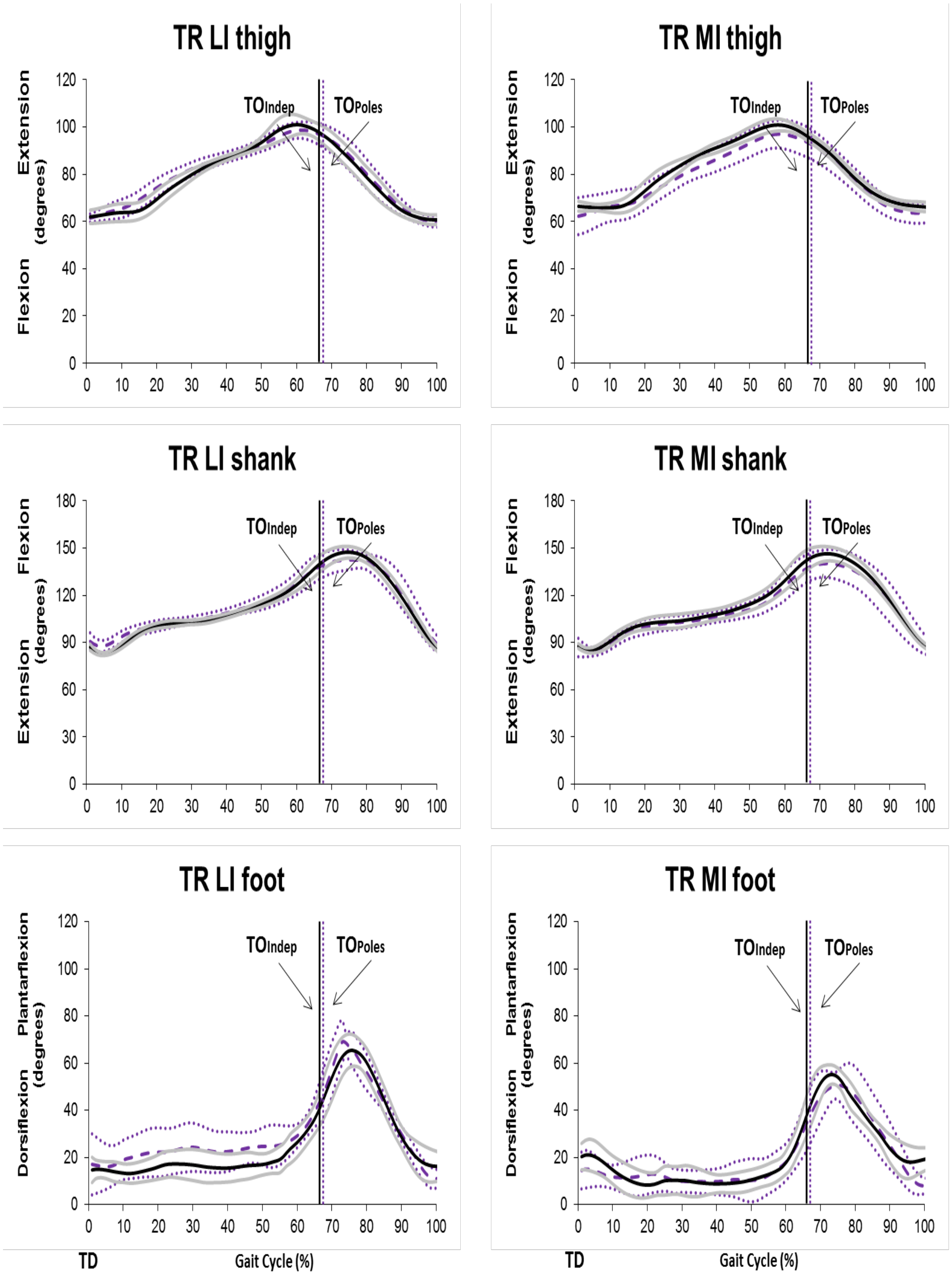


Figure C.1.3. (continued).

Appendix D.

Table D.1

Individual participant characteristics per visit.

Infant ID	Date of Birth	Gender	Visit	Mass (kg)	Standing Height (cm)	Ponderal Index (kg/m^3)	Full Leg Length (cm)	Bayley Score
AB	6/26/2011	M	-1	11.50	74.10	28.26	30.60	66
			0	11.48	76.90	27.21	30.80	71
CW	7/5/2011	F	-3	8.27	71.65	22.48	27.90	60
			-2	8.86	72.50	23.25	28.60	64
			-1	9.25	74.70	22.19	30.10	68
			0	9.22	76.60	20.51	30.80	71
CY	10/5/2011	F	-2	10.10	72.70	26.29	31.30	63
			-1	10.36	73.90	25.67	32.10	67
			0	10.78	74.80	25.76	32.60	71
KK	11/5/2011	F	-3	8.17	68.50	25.40	30.10	63
			-2	8.68	68.70	26.77	30.60	64
			-1	9.11	69.60	27.02	28.00	67
			0	9.09	70.90	25.50	31.30	71
KM	7/14/2011	F	-3	9.60	69.40	28.72	27.50	64
			-2	10.06	73.00	25.86	30.10	68
			-1	9.91	75.50	23.03	30.60	68
			0	10.22	77.00	22.39	31.50	71

Table D.1 (continued)

Infant ID	Date of Birth	Gender	Visit	Mass (kg)	Standing Height (cm)	Ponderal Index (kg/m^3)	Full Leg Length (cm)	Bayley Score
LB	11/22/2011	F	-2	8.79	69.80	25.85	27.60	63
			-1	9.06	72.50	23.77	28.00	67
			0	9.06	72.50	23.77	29.10	71
LH	6/30/2011	F	-4	9.31	67.60	30.14	24.30	62
			-3	9.75	68.70	30.07	27.30	59
			-2	10.14	69.50	30.21	28.20	64
			-1	10.84	72.40	28.56	28.80	70
			0	10.92	74.75	26.15	29.20	71
MV	8/7/2011	M	-1	11.28	71.60	30.73	28.60	62
			0	11.96	72.50	31.38	30.20	71
OM	11/29/2011	F	-4	9.51	68.40	29.72	27.50	64
			-3	9.72	68.80	29.85	29.00	66
			-2	10.12	70.40	29.00	27.20	67
			-1	10.64	71.70	28.90	30.20	68
			0	11.10	73.50	27.96	30.40	71
	OVERALL							
		***M* (SD)**	-4	9.41 (0.14)	68.00 (0.57)	29.93 (0.30)	25.90 (2.26)	63 (1)
			-3	9.10 (0.81)	69.41 (1.30)	27.30 (3.28)	28.36 (1.17)	62 (3)
			-2	9.54 (0.71)	70.94 (1.75)	26.75 (2.28)	29.43 (1.35)	65 (2)
			-1	10.22 (0.93)	72.89 (1.83)	26.46 (2.96)	30.00 (1.30)	67 (3)
			0	10.43 (1.09)	74.38 (2.19)	25.63 (3.19)	30.67 (1.10)	71 (0)

Table D.2

Participant Normalized Step Width and Stride Length with Coefficient of Variation (CV) values per condition per testing session.

Infant ID	**Visit**	**Normalized Step Width**		**CV Step Width**		**Normalized Stride Length**		**CV Stride Length**	
		C	***LG***	***C***	***LG***	***C***	***LG***	***C***	***LG***
		M (SD)	*M* (SD)	*M*	*M*	*M* (SD)	*M* (SD)	*M*	*M*
AB	-1	0.56 (0.14)	0.58 (0.08)	24.57	13.68	1.54 (0.20)	1.50 (0.20)	13.18	13.37
	0	0.45 (0.08)	0.47 (0.10)	18.20	21.68	1.36 (0.22)	1.62 (0.17)	16.24	10.18
CW	-3	0.49 (0.08)	0.48 (0.07)	16.73	14.27	0.72 (0.19)	0.76 (0.22)	26.60	29.15
	-2	0.47 (0.11)	0.54 (0.07)	22.99	13.38	1.18 (0.16)	1.21 (0.24)	13.64	19.52
	-1	0.53 (0.05)	0.44 (0.06)	9.56	13.41	1.12 (0.20)	1.03 (0.11)	17.99	10.73
	0	0.30 (0.06)	0.31 (0.05)	19.85	17.21	1.19 (0.19)	1.20 (0.10)	16.10	8.00
CY	-2	0.34 (0.06)	0.48 (0.07)	16.62	13.64	1.05 (0.29)	1.16 (0.08)	27.76	6.93
	-1	0.44 (0.07)	0.44 (0.05)	16.93	10.59	1.16 (0.11)	1.25 (0.11)	9.86	8.60
	0	0.49 (0.07)	0.52 (0.05)	13.43	10.03	1.32 (0.08)	1.51 (0.11)	6.29	7.54
KK	-3	0.37 (0.07)	0.43 (0.05)	18.57	12.16	1.01 (0.63)	0.74 (0.12)	0.00	15.88
	-2	0.24 (0.07)	0.30 (0.10)	28.01	33.94	1.23 (0.20)	1.20 (0.16)	15.97	13.07
	-1	0.23 (0.05)	0.28 (0.06)	24.11	20.69	1.32 (0.16)	1.44 (0.10)	12.00	6.91
	0	0.17 (0.12)	0.26 (0.08)	0.00	30.31	0.97 (0.10)	1.04 (0.21)	10.60	19.99
KM	-3	0.50 (0.07)	0.58 (0.07)	14.41	12.60	0.79 (0.19)	0.63 (0.09)	24.49	14.00
	-2	0.28 (0.05)	0.33 (0.06)	18.14	17.55	1.03 (0.16)	0.90 (0.08)	15.68	9.41
	-1	0.22 (0.06)	0.31 (0.07)	26.35	21.93	1.39 (0.15)	1.51 (0.09)	11.09	6.03
	0	0.21 (0.06)	0.26 (0.06)	28.18	24.74	1.17 (0.05)	1.06 (0.09)	4.53	8.79

Table D.2 (continued)

LB	-2	0.33 (0.07)	0.38 (0.07)	20.79	18.34	1.31 (0.25)	1.28 (0.25)	18.75	19.43
	-1	0.41 (0.12)	0.36 (0.07)	27.87	19.17	1.51 (0.20)	1.18 (0.13)	12.98	10.71
	0	0.39 (0.10)	0.42 (0.08)	24.80	19.51	1.19 (0.11)	1.18 (0.13)	9.28	11.08
LH	-4	0.52 (0.04)	0.62 (0.11)	7.98	17.64	1.18 (0.10)	1.21 (0.25)	8.49	20.56
	-3	0.29 (0.06)	0.32 (0.07)	19.32	21.45	1.32 (0.13)	1.39 (0.14)	9.64	9.72
	-2	0.24 (0.06)	0.32 (0.04)	23.81	12.29	1.27 (0.07)	1.05 (0.19)	5.89	18.31
	-1	0.39 (0.09)	0.38 (0.05)	22.00	12.91	1.69 (0.14)	1.57 (0.10)	8.20	6.29
	0	0.48 (0.06)	0.49 (0.08)	12.55	17.01	1.61 (0.15)	1.69 (0.13)	9.22	7.83
MV	-1	0.60 (0.09)	0.56 (0.07)	15.46	12.19	1.37 (0.11)	1.23 (0.17)	8.00	13.50
	0	0.41 (0.09)	0.42 (0.07)	21.63	17.66	1.81 (0.11)	1.69 (0.19)	6.01	10.94
OM	-4	0.39 (0.06)	0.45 (0.06)	15.77	13.24	0.95 (0.12)	1.01 (0.24)	12.69	24.03
	-3	0.34 (0.10)	0.39 (0.07)	28.04	18.31	1.04 (0.13)	1.19 (0.18)	12.88	14.78
	-2	0.49 (0.07)	0.49 (0.06)	14.59	13.21	1.32 (0.10)	1.39 (0.36)	7.80	25.72
	-1	0.39 (0.08)	0.39 (0.06)	20.61	15.92	1.25 (0.20)	1.34 (0.14)	16.08	10.62
	0	0.31 (0.07)	0.39 (0.07)	23.54	18.23	1.29 (0.15)	1.12 (0.13)	11.51	11.48
OVERALL									
***M* (SD)**	(n=2) -4	0.40 (0.07)	0.49 (0.10)	11.87 (5.51)	15.44 (3.11)	0.99 (0.15)	1.06 (0.26)	10.59 (2.97)	22.30 (2.45)
	(n=5) -3	0.40 (0.11)	0.45 (0.12)	19.41 (5.18)	15.76 (4.00)	0.99 (0.35)	0.99 (0.34)	18.40 (8.40)	16.71 (7.34)
	(n=7) -2	0.34 (0.12)	0.38 (0.11)	20.71 (4.64)	17.48 (7.62)	1.18 (0.22)	1.16 (0.25)	15.07 (7.24)	16.06 (6.56)
	(n=9) -1	0.40 (0.14)	0.40 (0.11)	20.83 (5.89)	15.61 (4.05)	1.34 (0.25)	1.33 (0.22)	12.15 (3.36)	9.64 (2.85)
	(n=9) 0	0.35 (0.13)	0.40 (0.11)	20.27 (5.43)	19.60 (5.65)	1.31 (0.26)	1.35 (0.29)	9.98 (4.18)	10.65 (3.82)

Table D.3

Participant Normalized Stride Frequency and Stride Velocity with Coefficient of Variation (CV) values per condition per session.

Infant ID	**Visit**	**Normalized Stride Frequency**		**CV Stride Frequency**		**Normalized Stride Velocity**		**CV Stride Velocity**	
		C	***LG***	***C***	***LG***	***C***	***LG***	***C***	***LG***
		M (SD)	*M* (SD)	*M*	*M*	*M* (SD)	*M* (SD)	*M*	*M*
AB	-1	0.24 (0.03)	0.19 (0.01)	12.41	7.48	0.36 (0.03)	0.29 (0.04)	7.22	12.39
	0	0.23 (0.02)	0.25 (0.02)	9.24	6.79	0.29 (0.07)	0.39 (0.07)	24.08	16.80
CW	-3	0.09 (0.03)	0.08 (0.02)	34.63	21.51	0.07 (0.03)	0.06 (0.02)	40.06	34.41
	-2	0.12 (0.01)	0.12 (0.01)	9.76	8.56	0.14 (0.02)	0.13 (0.03)	15.32	26.28
	-1	0.24 (0.03)	0.19 (0.03)	10.65	13.75	0.27 (0.07)	0.19 (0.03)	26.07	16.42
	0	0.24 (0.04)	0.23 (0.02)	18.11	8.88	0.31 (0.08)	0.28 (0.03)	26.45	10.87
CY	-2	0.18 (0.02)	0.19 (0.01)	10.31	5.69	0.18 (0.05)	0.22 (0.04)	29.71	15.88
	-1	0.22 (0.02)	0.20 (0.01)	7.52	5.69	0.26 (0.03)	0.25 (0.03)	12.71	13.42
	0	0.24 (0.01)	0.24 (0.01)	4.62	6.22	0.32 (0.03)	0.36 (0.04)	8.72	11.09
KK	-3	0.13 (0.02)	0.08 (0.02)	13.27	27.12	0.12 (0.07)	0.07 (0.02)	0.00	33.86
	-2	0.17 (0.03)	0.22 (0.03)	20.48	15.86	0.21 (0.08)	0.26 (0.07)	36.83	27.28
	-1	0.21 (0.03)	0.25 (0.04)	15.23	18.13	0.29 (0.08)	0.36 (0.09)	28.78	25.36
	0	0.17 (0.04)	0.20 (0.03)	22.85	17.06	0.17 (0.05)	0.21 (0.07)	29.04	34.77
KM	-3	0.13 (0.04)	0.14 (0.04)	27.45	27.41	0.10 (0.03)	0.09 (0.03)	28.08	37.14
	-2	0.21 (0.02)	0.17 (0.04)	10.78	21.01	0.22 (0.05)	0.17 (0.06)	24.12	33.40
	-1	0.27 (0.03)	0.28 (0.02)	10.44	8.31	0.37 (0.06)	0.43 (0.04)	15.97	9.83
	0	0.23 (0.03)	0.26 (0.04)	13.74	17.49	0.27 (0.04)	0.27 (0.05)	14.14	18.84

Table D.3 (continued)

LB	-2	0.16 (0.01)	0.13 (0.01)	6.71	10.39	0.20 (0.05)	0.17 (0.04)	23.35	24.15
	-1	0.14 (0.02)	0.10 (0.03)	16.39	26.92	0.21 (0.06)	0.12 (0.04)	28.47	34.76
	0	0.15 (0.02)	0.17 (0.02)	16.37	10.75	0.18 (0.03)	0.21 (0.03)	15.27	16.40
LH	-4	0.15 (0.03)	0.13 (0.02)	17.26	12.30	0.18 (0.04)	0.16 (0.04)	23.79	24.69
	-3	0.16 (0.01)	0.19 (0.02)	8.46	10.44	0.20 (0.03)	0.27 (0.04)	13.15	15.53
	-2	0.18 (0.03)	0.18 (0.01)	16.77	7.06	0.24 (0.05)	0.20 (0.02)	19.86	12.30
	-1	0.17 (0.02)	0.20 (0.02)	13.58	8.53	0.30 (0.06)	0.31 (0.03)	19.75	8.39
	0	0.24 (0.02)	0.25 (0.02)	6.60	9.88	0.38 (0.04)	0.42 (0.06)	11.23	13.28
MV	-1	0.20 (0.04)	0.15 (0.02)	18.89	15.44	0.28 (0.06)	0.19 (0.03)	20.45	15.07
	0	0.26 (0.02)	0.26 (0.01)	7.70	4.63	0.47 (0.03)	0.43 (0.04)	6.55	10.05
OM	-4	0.09 (0.03)	0.10 (0.02)	36.39	17.13	0.09 (0.04)	0.09 (0.02)	43.22	25.00
	-3	0.16 (0.02)	0.17 (0.02)	13.74	13.27	0.17 (0.03)	0.19 (0.03)	19.90	16.20
	-2	0.15 (0.02)	0.17 (0.01)	12.47	4.39	0.19 (0.03)	0.24 (0.07)	15.57	29.42
	-1	0.19 (0.02)	0.17 (0.02)	12.14	10.82	0.24 (0.05)	0.23 (0.05)	19.61	19.42
	0	0.17 (0.02)	0.13 (0.02)	12.56	12.93	0.21 (0.04)	0.15 (0.03)	19.56	20.90
OVERALL									
***M* (SD)**	(n=2) -4	0.10 (0.04)	0.11 (0.02)	26.82 (13.53)	14.71 (3.42)	0.10 (0.05)	0.11 (0.04)	33.50 (13.74)	24.84 (0.22)
	(n=5) -3	0.13 (0.04)	0.14 (0.05)	19.51 (11.02)	19.95 (7.82)	0.14 (0.06)	0.16 (0.08)	25.30 (11.58)	27.43 (10.63)
	(n=7) -2	0.17 (0.04)	0.18 (0.04)	12.47 (4.67)	10.42 (5.99)	0.20 (0.06)	0.21 (0.07)	23.54 (7.74)	24.10 (7.49)
	(n=9) -1	0.21 (0.04)	0.20 (0.06)	13.03 (3.44)	12.79 (6.68)	0.28 (0.07)	0.26 (0.11)	19.89 (7.23)	17.23 (8.32)
	(n=9) 0	0.21 (0.05)	0.22 (0.05)	12.42 (5.97)	10.52 (4.57)	0.28 (0.10)	0.30 (0.11)	17.23 (8.01)	17.00 (7.66)

Table D.4

Gait cycle phase values for individual infants per condition per testing session.

Infant ID	**Visit**	**Swing Phase (%)**		**Stance Phase (%)**		**Double Support Phase (%)**	
		C	***LG***	***C***	***LG***	***C***	***LG***
		M (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)
AB	-1	31.70 (5.52)	28.55 (8.92)	68.30 (5.52)	71.45 (8.92)	54.86 (8.49)	59.91 (7.35)
	0	27.54 (4.70)	29.87 (4.48)	72.46 (4.70)	70.13 (4.48)	61.51 (11.36)	60.32 (8.26)
CW	-3	16.55 (8.19)	16.97 (8.13)	83.45 (8.19)	83.03 (8.13)	66.49 (15.09)	65.69 (15.07)
	-2	20.65 (3.70)	18.52 (4.49)	79.35 (3.70)	81.48 (4.49)	74.96 (7.07)	71.61 (6.55)
	-1	26.76 (4.52)	26.36 (5.00)	73.24 (4.52)	73.64 (5.00)	63.48 (8.10)	62.04 (10.03)
	0	29.52 (5.17)	28.80 (5.42)	70.48 (5.17)	71.20 (5.42)	58.60 (7.88)	59.69 (8.21)
CY	-2	23.73 (4.14)	22.41 (4.15)	76.27 (4.14)	77.59 (4.15)	69.46 (8.42)	74.40 (8.20)
	-1	24.95 (3.27)	27.13 (3.19)	75.05 (3.27)	72.87 (3.19)	67.46 (7.46)	61.89 (8.52)
	0	31.87 (4.67)	31.84 (6.20)	68.13 (4.67)	68.16 (6.20)	54.47 (11.54)	54.20 (7.97)
KK	-3	15.25 (2.54)	11.32 (2.49)	84.99 (2.59)	88.68 (2.49)	76.80 (18.40)	77.56 (28.49)
	-2	25.24 (8.05)	30.33 (8.21)	74.76 (8.05)	69.67 (8.21)	62.33 (14.19)	57.00 (14.20)
	-1	24.74 (5.70)	27.55 (4.68)	76.31 (7.63)	72.45 (4.68)	69.78 (16.67)	60.59 (22.43)
	0	24.94 (6.06)	26.44 (4.45)	75.06 (6.06)	73.56 (4.45)	76.71 (18.11)	64.35 (15.75)
KM	-3	17.48 (8.50)	16.10 (7.28)	82.52 (8.50)	83.90 (7.28)	58.55 (17.74)	72.61 (13.51)
	-2	27.03 (4.98)	24.55 (5.99)	72.97 (4.98)	75.45 (5.99)	66.55 (8.25)	69.47 (10.89)
	-1	26.99 (3.75)	29.25 (3.40)	73.01 (3.75)	70.75 (3.40)	67.57 (8.94)	62.59 (7.14)
	0	32.74 (4.75)	39.31 (5.62)	67.26 (4.75)	60.69 (5.62)	53.85 (9.63)	51.04 (11.10)

Table D.4 (continued)

LB	-2	25.39 (4.52)	25.34 (4.94)	75.27 (5.06)	74.66 (4.94)	68.91 (14.11)	71.83 (13.38)
	-1	28.67 (7.78)	21.63 (8.45)	71.33 (7.78)	78.37 (8.45)	56.75 (13.74)	64.89 (18.01)
	0	25.18 (5.73)	27.56 (3.31)	74.82 (5.73)	72.44 (3.31)	59.36 (12.72)	64.31 (15.10)
LH	-4	24.15 (10.90)	25.43 (8.29)	75.85 (10.90)	74.57 (8.29)	71.78 (8.37)	69.85 (19.04)
	-3	30.58 (4.05)	32.74 (6.14)	69.42 (4.05)	67.26 (6.14)	57.82 (15.61)	47.82 (14.24)
	-2	33.48 (6.76)	28.73 (3.18)	66.52 (6.76)	71.27 (3.18)	47.28 (13.33)	60.84 (9.38)
	-1	32.72 (7.27)	32.77 (3.72)	67.28 (7.27)	67.23 (3.72)	55.38 (16.90)	50.72 (9.09)
	0	32.63 (3.98)	32.61 (3.41)	67.37 (3.98)	67.39 (3.41)	58.30 (11.06)	52.40 (8.75)
MV	-1	28.57 (6.91)	24.41 (8.81)	71.43 (6.91)	75.59 (8.81)	64.00 (17.09)	65.65 (11.56)
	0	34.91 (5.28)	37.43 (3.81)	65.09 (5.28)	62.57 (3.81)	49.16 (5.24)	43.10 (4.87)
OM	-4	15.17 (5.95)	13.48 (3.86)	84.83 (5.95)	86.52 (3.86)	73.55 (13.62)	73.72 (12.95)
	-3	24.83 (6.49)	24.98 (7.10)	75.17 (6.49)	75.02 (7.10)	62.11 (8.23)	65.76 (13.66)
	-2	25.72 (4.31)	26.89 (8.47)	74.28 (4.31)	73.11 (8.47)	67.40 (13.25)	61.47 (14.64)
	-1	27.67 (4.86)	29.64 (5.41)	72.33 (4.86)	70.36 (5.41)	65.08 (13.36)	59.04 (16.39)
	0	22.46 (3.43)	21.00 (4.02)	77.54 (3.43)	79.00 (4.02)	69.88 (7.13)	70.82 (20.32)
OVERALL							
***M* (SD)**	(n=2) -4	15.92 (6.63)	17.19 (7.84)	84.08 (6.63)	82.81 (7.84)	73.34 (12.93)	72.29 (15.06)
	(n=5) -3	22.58 (8.38)	22.86 (9.80)	77.54 (8.41)	77.14 (9.80)	63.44 (15.15)	63.87 (17.45)
	(n=7) -2	25.96 (6.21)	25.96 (7.04)	74.13 (6.26)	74.04 (7.04)	65.83 (13.02)	65.55 (13.19)
	(n=9) -1	27.71 (5.92)	27.24 (6.48)	72.47 (6.28)	72.76 (6.48)	63.60 (13.63)	60.61 (14.23)
	(n=9) 0	28.66 (6.29)	30.51 (6.79)	71.34 (6.29)	69.49 (6.79)	61.90 (14.41)	58.13 (14.36)

Table D.5

Wrist width and foot rotation ratio values for individual infants per condition per testing session.

Infant ID	**Visit**	**Wrist Width (mm)**		**Foot Rotation Ratio**	
		C	***LG***	***C***	***LG***
		M (SD)	*M* (SD)	*M*	*M*
AB	-1	228.15 (55.66)	203.38 (70.77)	0.82	0.76
	0	275.91 (47.74)	251.24 (29.42)	0.49	0.26
CW	-3	222.44 (14.45)	221.02 (26.46)	0.58	0.65
	-2	231.20 (43.65)	214.58 (13.92)	0.35	0.33
	-1	236.71 (41.58)	228.63 (46.04)	0.59	0.92
	0	214.24 (68.37)	239.14 (7.84)	0.66	0.92
CY	-2	207.93 (25.07)	219.87 (41.62)	0.59	0.90
	-1	240.62 (23.25)	270.16 (12.74)	0.45	0.93
	0	203.92 (32.34)	232.18 (21.12)	0.58	0.49
KK	-3	247.10 (35.06)	231.08 (35.01)	0.50	0.69
	-2	176.84 (4.32)	171.68 (17.80)	0.55	0.64
	-1	201.93 (33.34)	216.45 (19.28)	0.90	0.55
	0	206.66 (44.26)	248.17 (26.00)	0.41	0.41
KM	-3	215.89 (38.70)	195.32 (31.52)	0.23	0.16
	-2	262.02 (7.04)	284.32 (19.97)	0.39	0.44
	-1	239.81 (23.57)	233.88 (30.88)	0.77	0.58
	0	248.73 (38.24)	257.65 (61.43)	0.65	0.77

Table D.5 (continued)

LB	-2	195.29 (56.07)	195.58 (0.00)	0.34	0.35
	-1	147.96 (18.70)	176.13 (21.75)	0.60	0.57
	0	230.50 (118.36)	235.75 (71.93)	0.73	0.52
LH	-4	177.13 (20.38)	230.08 (44.95)	0.84	0.62
	-3	261.45 (13.77)	237.07 (33.18)	0.78	0.65
	-2	128.66 (26.80)	139.40 (11.33)	0.25	0.43
	-1	254.48 (33.06)	242.89 (22.51)	0.29	0.45
	0	244.95 (3.66)	276.92 (12.25)	0.61	0.93
MV	-1	272.40 (47.37)	174.81 (52.45)	0.54	0.85
	0	198.80 (22.21)	283.76 (21.45)	0.90	0.14
OM	-4	195.35 (28.84)	73.08 (137.73)	0.93	0.84
	-3	173.11 (19.48)	183.64 (29.63)	0.18	0.63
	-2	232.02 (18.97)	235.22 (69.26)	0.47	0.34
	-1	174.58 (3.00)	256.23 (32.04)	0.97	0.47
	0	243.85 (31.37)	262.95 (37.24)	0.87	0.83
OVERALL					
***M* (SD)**	(n=2) -4	188.06 (24.88)	135.88 (131.85)	0.89 (0.06)	0.73 (0.15)
	(n=5) -3	223.90 (37.42)	213.63 (34.05)	0.46 (0.25)	0.56 (0.22)
	(n=7) -2	204.85 (48.90)	208.37 (53.73)	0.42 (0.12)	0.49 (0.21)
	(n=9) -1	218.88 (48.77)	223.44 (43.84)	0.66 (0.22)	0.68 (0.19)
	(n=9) 0	230.32 (50.94)	253.54 (38.11)	0.66 (0.16)	0.59 (0.29)

Table D.6

Participant Anchoring Indices per condition per testing session.

Infant ID	**Visit**	**Roll Head**		**Roll Shoulders**		**Roll Trunk**		**Pitch Head**	
		C	***LG***	***C***	***LG***	***C***	***LG***	***C***	***LG***
		M (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)	*M* (SD)
AB	-1	-0.03 (0.14)	-0.21 (0.00)	0.02 (0.01)	-0.17 (0.06)	0.66 (0.04)	0.54 (0.22)	0.05 (0.21)	0.10 (0.15)
	0	-0.04 (0.17)	-0.11 (0.09)	-0.04 (0.10)	0.05 (0.25)	0.74 (0.08)	0.70 (0.06)	-0.28 (0.11)	-0.07 (0.23)
CW	-3	0.16 (0.09)	0.24 (0.24)	0.01 (0.17)	-0.02 (0.04)	0.53 (0.12)	0.46 (0.09)	0.28 (0.09)	0.31 (0.07)
	-2	0.03 (0.14)	-0.07 (0.05)	0.22 (0.23)	0.03 (0.27)	0.31 (0.50)	0.47 (0.61)	0.23 (0.02)	0.10 (0.12)
	-1	0.06 (0.02)	0.06 (0.05)	-0.04 (0.02)	-0.05 (0.08)	0.64 (0.12)	0.62 (0.04)	0.28 (0.08)	0.22 (0.06)
	0	0.15 (0.15)	0.19 (0.22)	0.01 (0.02)	0.08 (0.05)	0.78 (0.08)	0.75 (0.02)	0.21 (0.21)	0.07 (0.32)
CY	-2	0.04 (0.15)	-0.02 (0.16)	0.13 (0.18)	0.04 (0.05)	0.69 (0.12)	0.62 (0.14)	0.13 (0.30)	0.38 (0.25)
	-1	-0.05 (0.02)	-0.09 (0.07)	0.10 (0.13)	0.01 (0.23)	0.59 (0.17)	0.46 (0.17)	0.21 (0.36)	0.27 (0.13)
	0	0.03 (0.23)	-0.12 (0.20)	-0.06 (0.18)	-0.16 (0.11)	0.55 (0.03)	0.52 (0.07)	0.24 (0.11)	0.41 (0.20)
KK	-3	0.00 (0.02)	0.02 (0.22)	0.21 (0.20)	0.15 (0.18)	0.57 (0.12)	0.60 (0.08)	0.24 (0.27)	0.13 (0.21)
	-2	-0.21 (0.01)	-0.18 (0.18)	-0.22 (0.04)	-0.18 (0.14)	0.60 (0.11)	0.56 (0.04)	0.05 (0.13)	0.34 (0.18)
	-1	0.04 (0.22)	0.06 (0.23)	-0.07 (0.10)	-0.14 (0.02)	0.70 (0.13)	0.71 (0.06)	0.10 (0.14)	-0.03 (0.03)
	0	-0.03 (0.08)	-0.07 (0.15)	-0.05 (0.07)	0.12 (0.12)	0.72 (0.04)	0.74 (0.04)	0.22 (0.22)	0.39 (0.19)
KM	-3	-0.04 (0.05)	0.03 (0.03)	0.01 (0.07)	0.24 (0.09)	0.26 (0.34)	0.54 (0.38)	0.19 (0.19)	0.05 (0.03)
	-2	0.22 (0.20)	0.11 (0.07)	0.04 (0.10)	0.11 (0.07)	0.73 (0.05)	0.64 (0.13)	-0.05 (0.18)	0.21 (0.05)
	-1	0.09 (0.18)	-0.06 (0.05)	0.08 (0.22)	0.06 (0.13)	0.75 (0.03)	0.75 (0.03)	-0.01 (0.40)	0.36 (0.27)
	0	0.10 (0.35)	-0.04 (0.08)	-0.11 (0.18)	-0.11 (0.24)	0.79 (0.04)	0.70 (0.09)	0.17 (0.14)	0.26 (0.09)

Table D.6 (continued)

LB	-2	-0.11 (0.09)	-0.02 (0.00)	-0.09 (0.13)	-0.16 (0.00)	0.64 (0.13)	0.63 (0.00)	0.07 (0.08)	0.17 (0.00)
	-1	-0.12 (0.13)	-0.08 (0.07)	-0.19 (0.16)	-0.12 (0.08)	0.54 (0.15)	0.70 (0.03)	0.01 (0.11)	0.20 (0.46)
	0	0.18 (0.12)	0.07 (0.08)	-0.10 (0.09)	0.02 (0.10)	0.45 (0.13)	0.48 (0.09)	0.11 (0.05)	0.21 (0.17)
LH	-4	-0.09 (0.37)	0.03 (0.04)	0.03 (0.42)	0.08 (0.28)	0.07 (0.08)	0.52 (0.51)	0.14 (0.08)	0.42 (0.42)
	-3	0.10 (0.11)	-0.05 (0.11)	0.05 (0.16)	-0.14 (0.07)	0.68 (0.08)	0.70 (0.11)	0.13 (0.09)	0.18 (0.38)
	-2	0.21 (0.04)	0.12 (0.21)	0.15 (0.07)	0.08 (0.08)	0.71 (0.06)	0.58 (0.14)	0.13 (0.28)	0.23 (0.20)
	-1	0.05 (0.17)	0.11 (0.24)	-0.05 (0.09)	-0.06 (0.09)	0.80 (0.03)	0.85 (0.04)	0.10 (0.03)	0.26 (0.09)
	0	0.17 (0.15)	0.09 (0.08)	0.35 (0.17)	0.03 (0.18)	0.63 (0.08)	0.72 (0.02)	-0.22 (0.15)	0.17 (0.18)
MV	-1	-0.07 (0.16)	-0.08 (0.14)	0.17 (0.14)	0.14 (0.18)	0.59 (0.38)	0.80 (0.04)	0.25 (0.14)	0.26 (0.06)
	0	0.23 (0.36)	-0.09 (0.08)	0.24 (0.15)	0.00 (0.05)	0.84 (0.02)	0.61 (0.05)	0.34 (0.25)	-0.02 (0.27)
OM	-4	-0.05 (0.11)	0.22 (0.18)	-0.03 (0.06)	0.16 (0.14)	0.63 (0.20)	0.19 (0.31)	0.26 (0.17)	0.05 (0.14)
	-3	0.06 (0.08)	0.00 (0.11)	-0.03 (0.07)	-0.04 (0.11)	0.63 (0.10)	0.59 (0.24)	-0.21 (0.09)	0.05 (0.26)
	-2	0.23 (0.54)	-0.12 (0.06)	0.30 (0.45)	0.04 (0.05)	0.76 (0.05)	0.50 (0.04)	0.20 (0.20)	0.11 (0.18)
	-1	0.08 (0.11)	0.12 (0.14)	0.05 (0.12)	0.08 (0.15)	0.52 (0.02)	0.56 (0.13)	0.20 (0.10)	-0.05 (0.17)
	0	U*	U*	U*	U*	0.67 (0.11)	0.68 (0.14)	U*	U*
OVERALL									
***M* (SD)**	(n=2) -4	-0.07 (0.23)	0.15 (0.16)	0.00 (0.21)	0.13 (0.17)	0.40 (0.34)	0.32 (0.38)	0.22 (0.14)	0.20 (0.31)
	(n=5) -3	0.06 (0.10)	0.05 (0.17)	0.05 (0.15)	0.04 (0.17)	0.54 (0.21)	0.58 (0.20)	0.14 (0.22)	0.14 (0.22)
	(n=7) -2	0.06 (0.25)	-0.02 (0.16)	0.08 (0.25)	-0.01 (0.14)	0.64 (0.23)	0.58 (0.17)	0.11 (0.18)	0.23 (0.22)
	(n=9) -1	0.00 (0.14)	-0.01 (0.15)	0.00 (0.16)	-0.02 (0.15)	0.64 (0.17)	0.67 (0.15)	0.13 (0.20)	0.18 (0.21)
	(n=9) 0	0.10 (0.21)	0.00 (0.15)	0.03 (0.20)	0.00 (0.15)	0.68 (0.14)	0.65 (0.12)	0.09 (0.26)	0.17 (0.24)

U*: Data unavailable due to repeated removal of headstrap with retro-reflective markers by infant.

Appendix E.

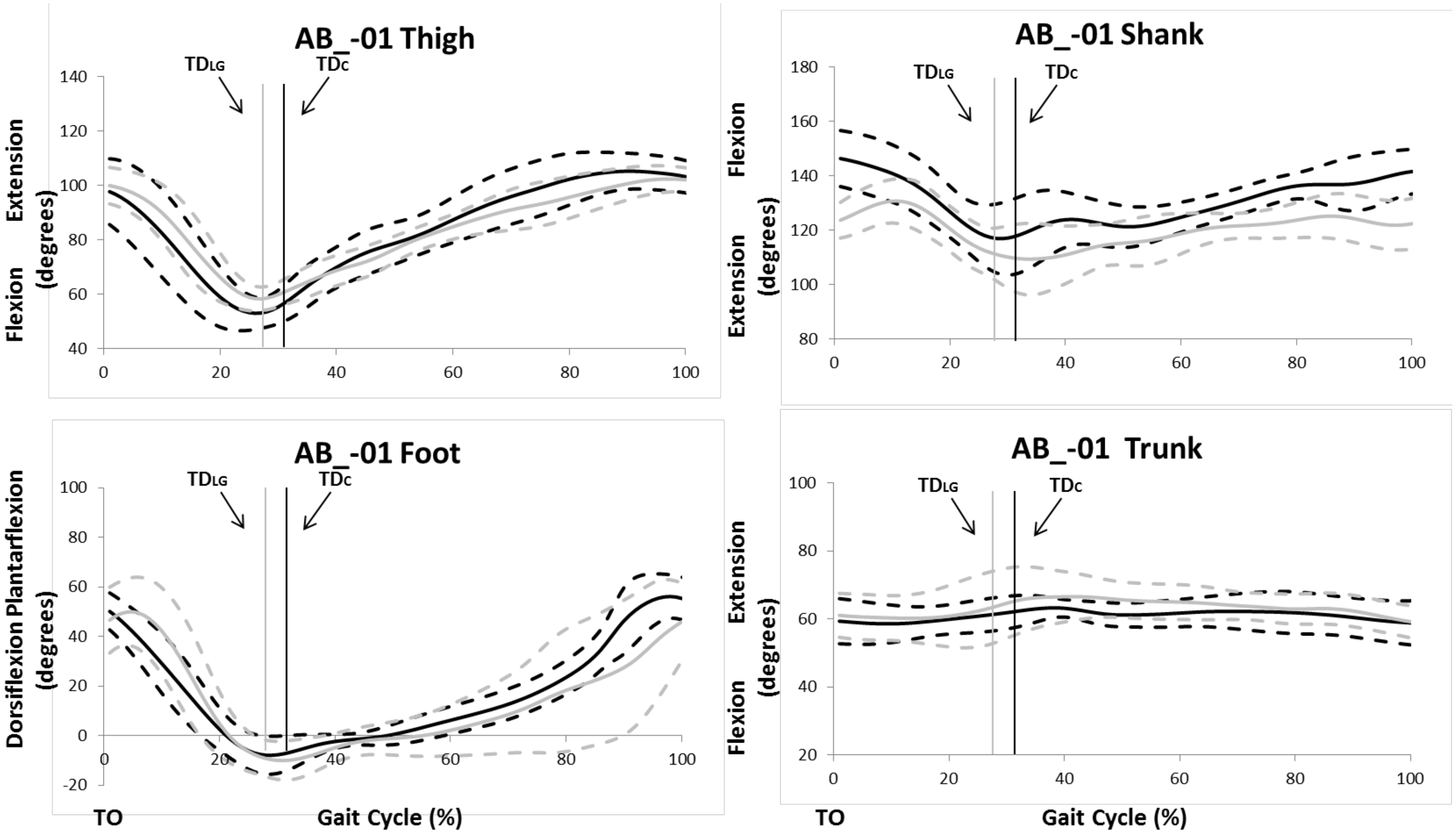


Figure E.1. Segmental angles for individual infants ±1 standard deviation envelope of the thigh, shank, foot, and trunk across visits. TO=Toe off; TD_{C}=Touchdown during the Control (diaper-only) condition; $\mathrm{TD}_{\mathrm{LG}}$=Touchdown during the Lycra Garment + diaper condition. — Control (diaper-only); — Lycra Garment.

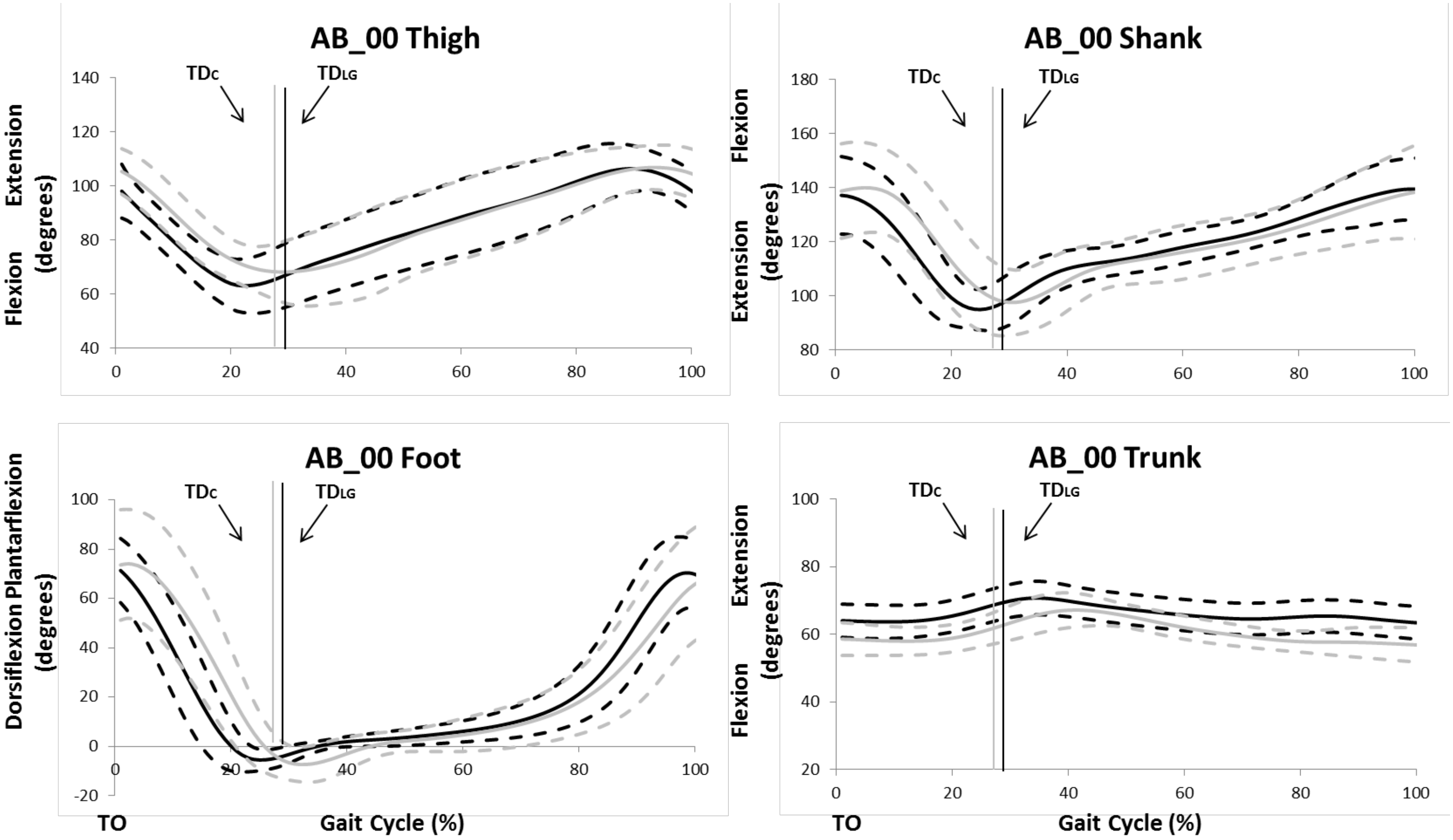


Figure E.1. (continued)

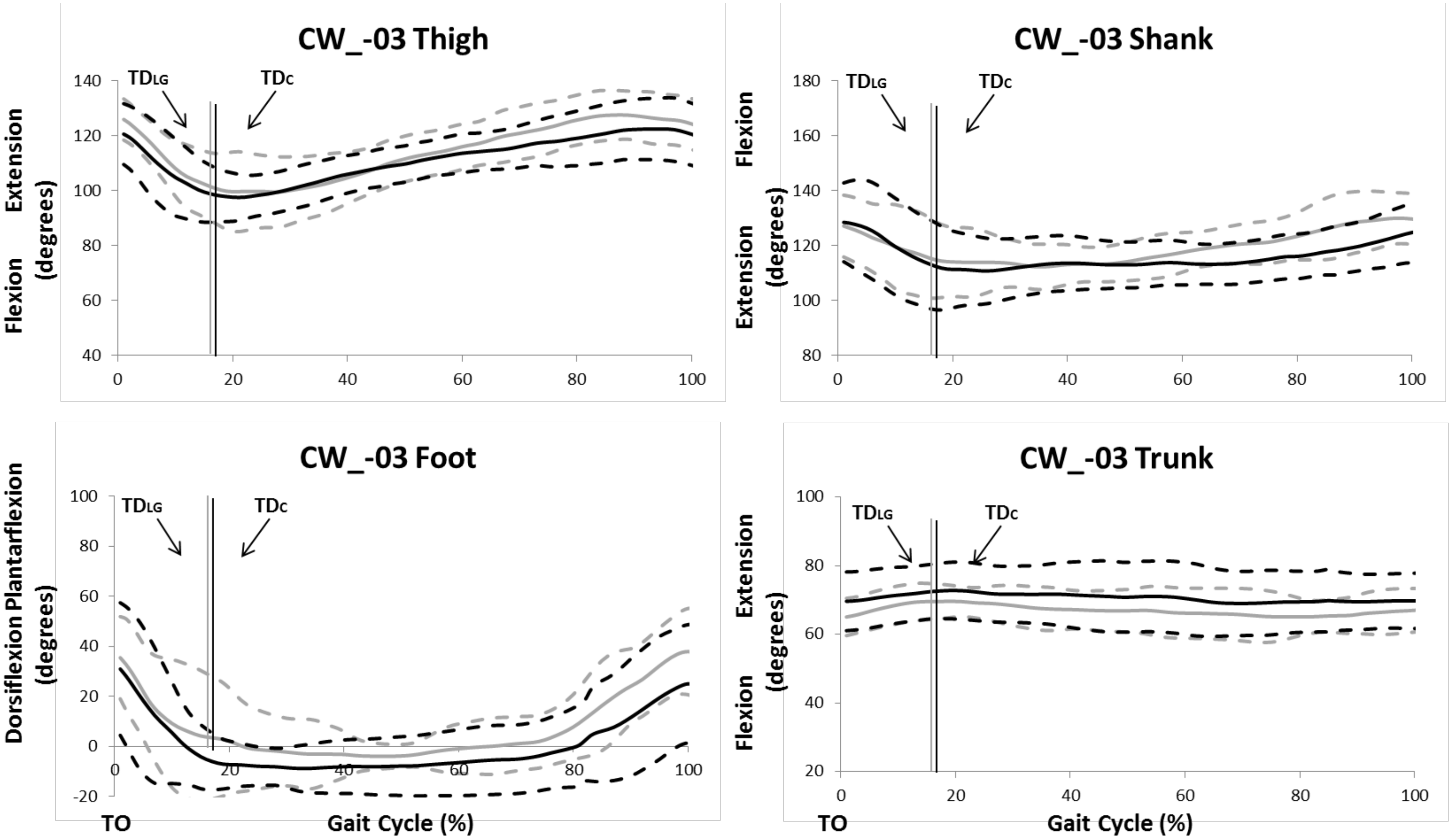
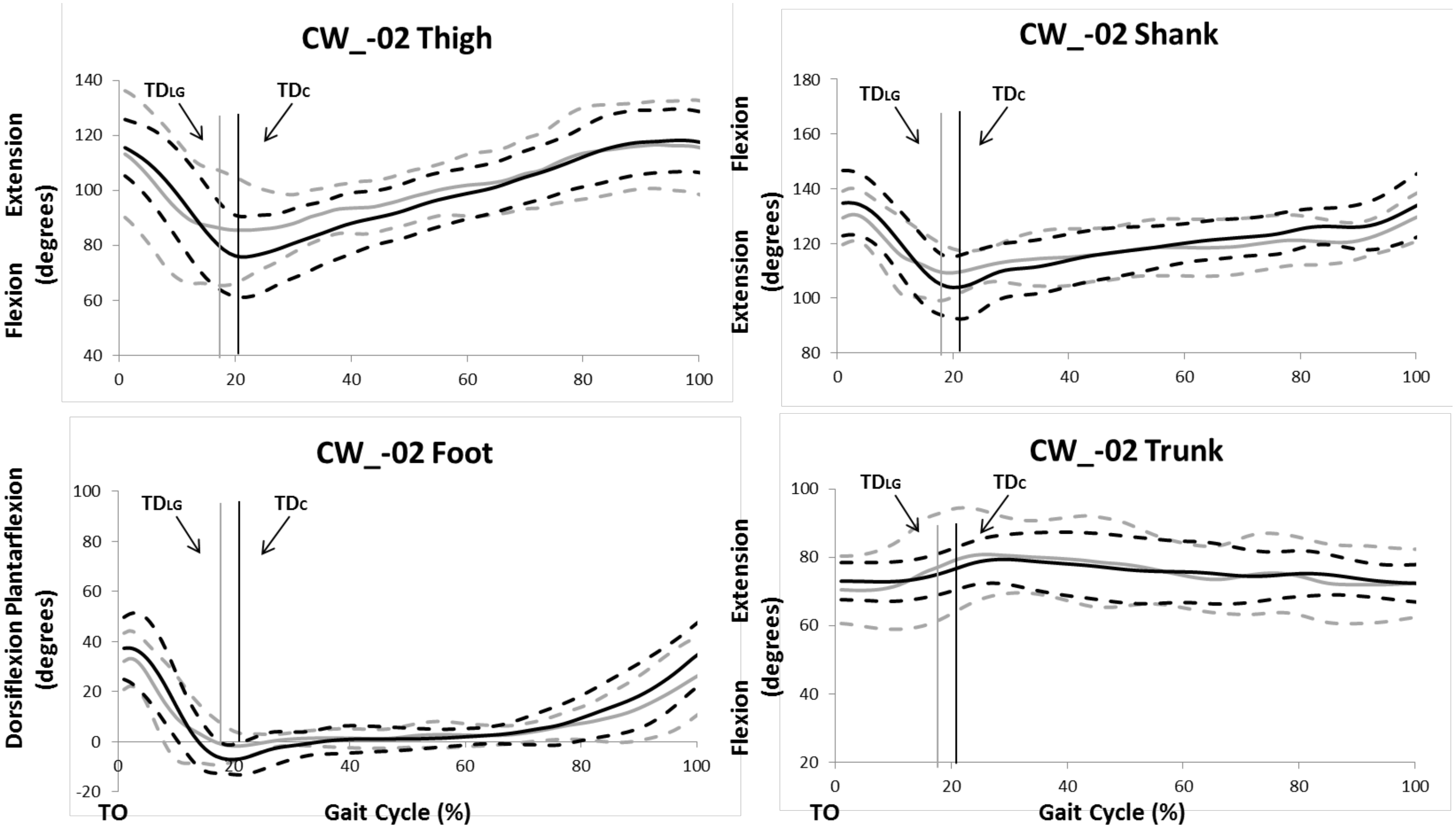


Figure E.1. (continued)



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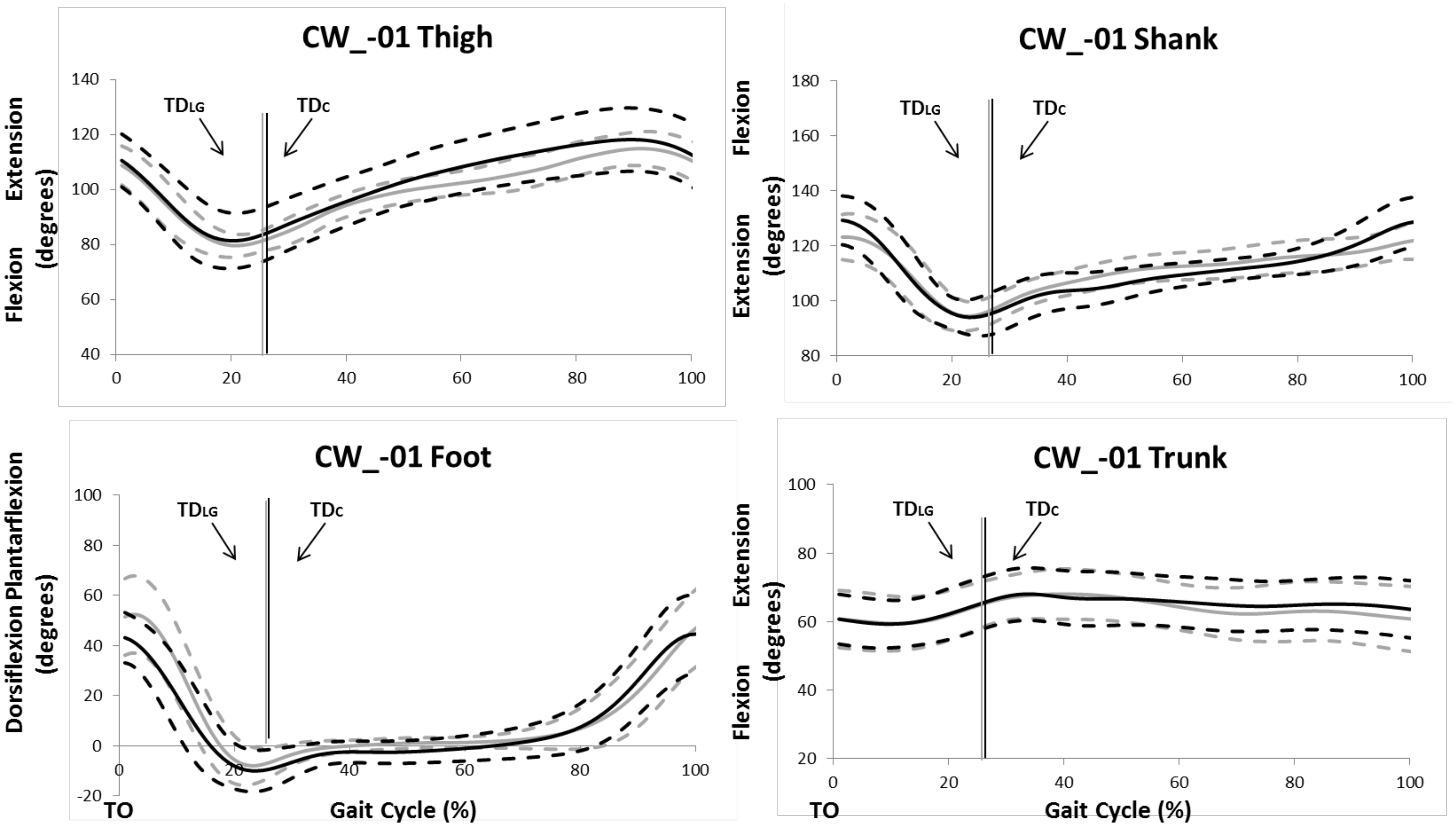

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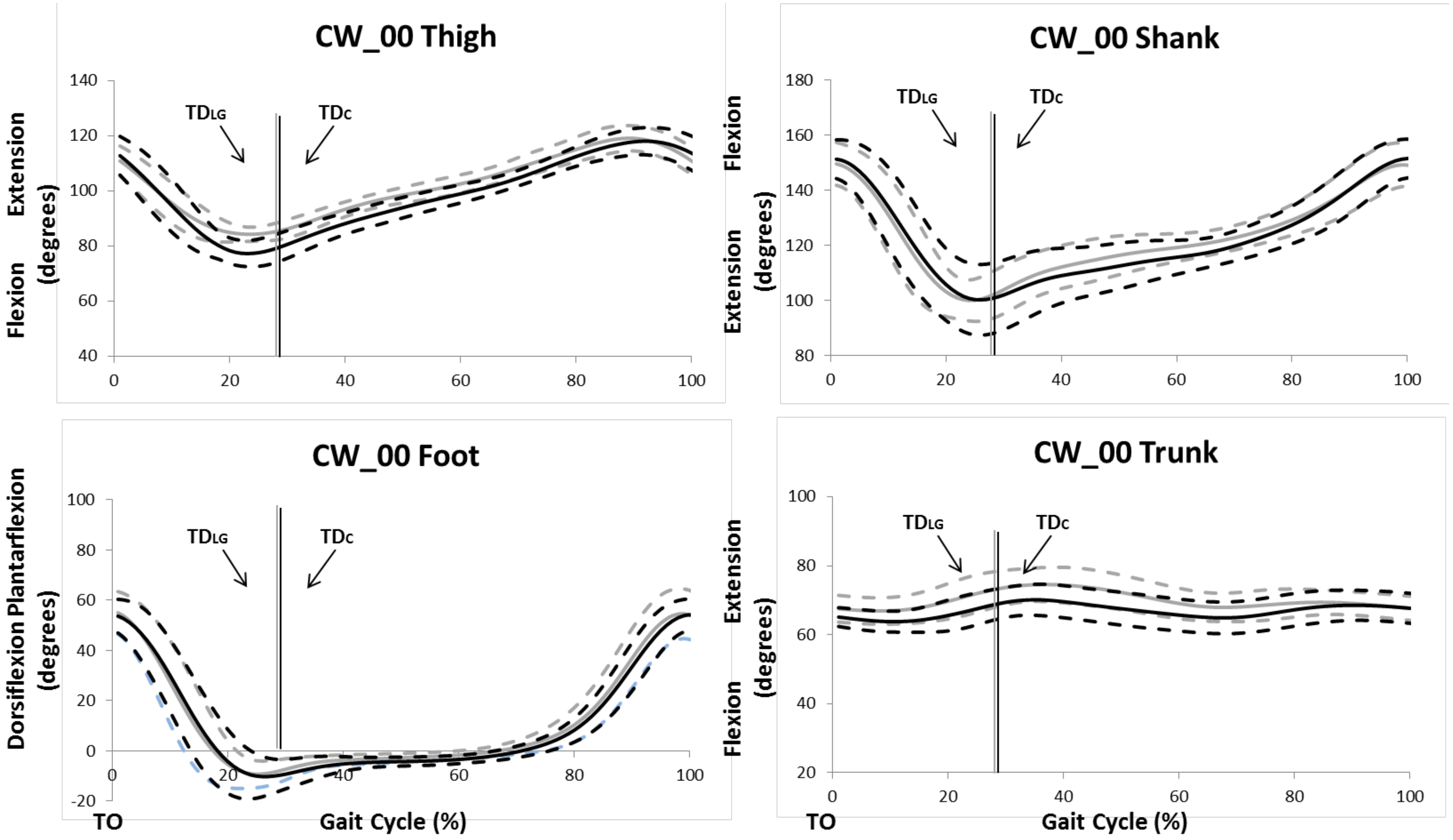

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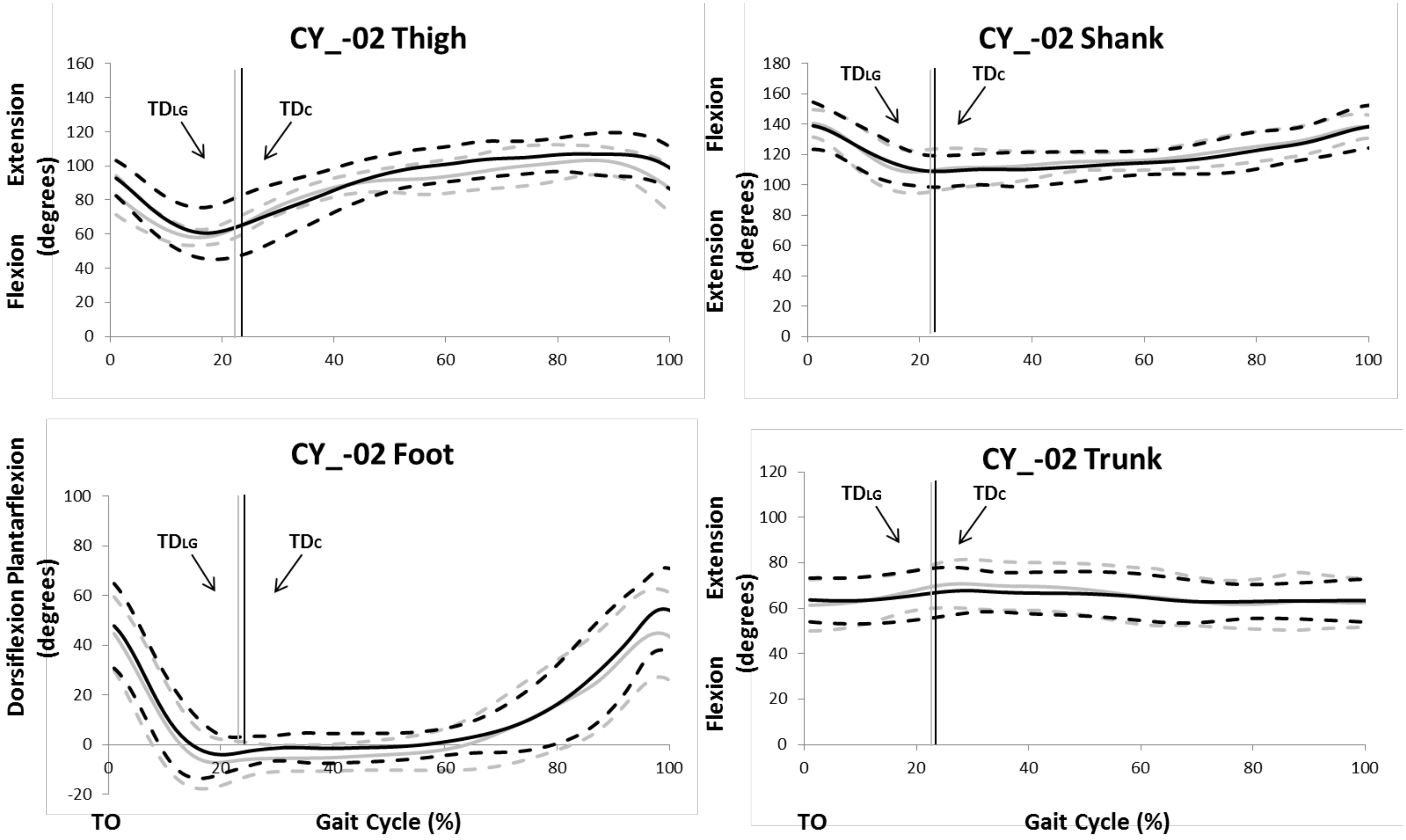


Figure E.1. (continued)

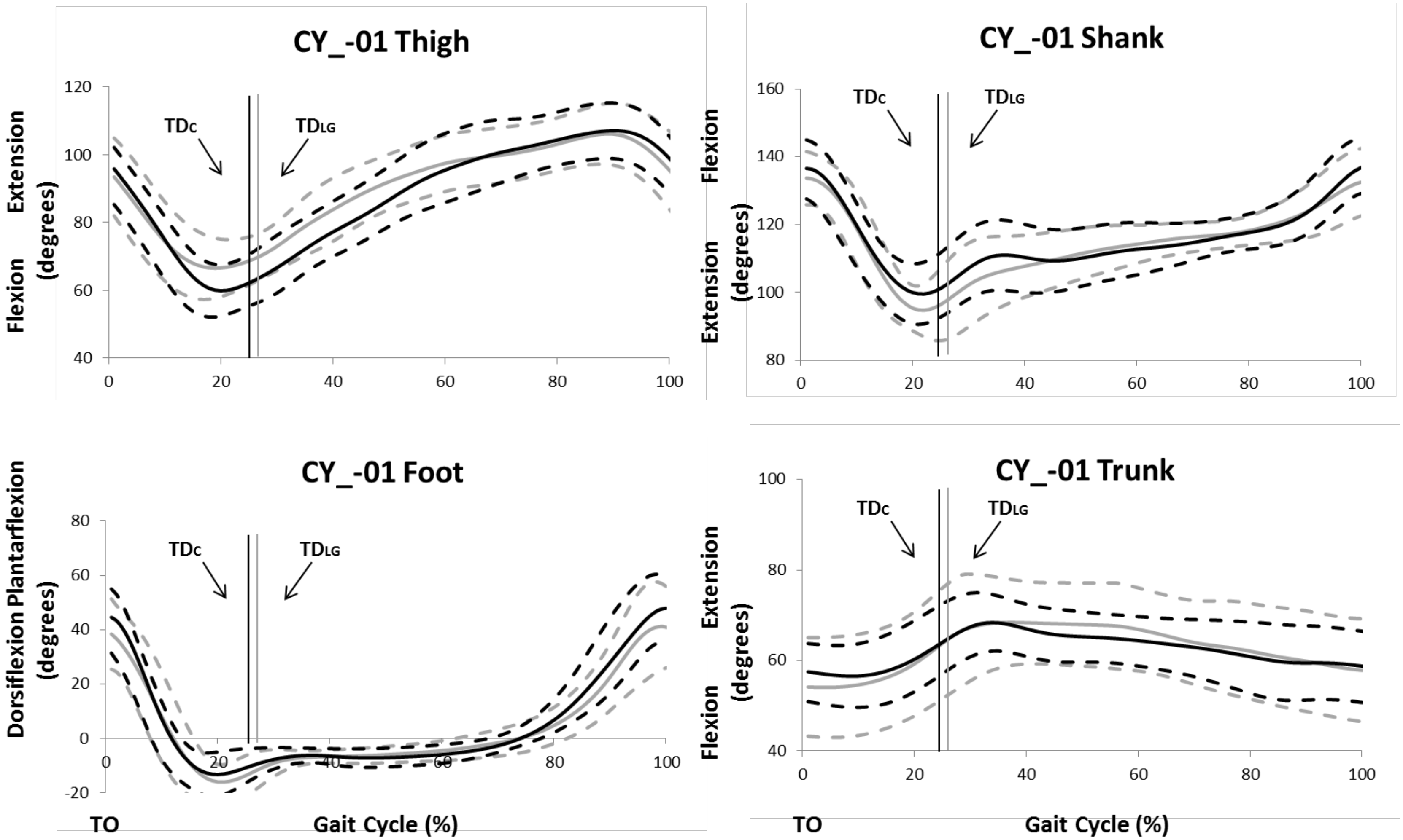


Figure E.1. (continued)

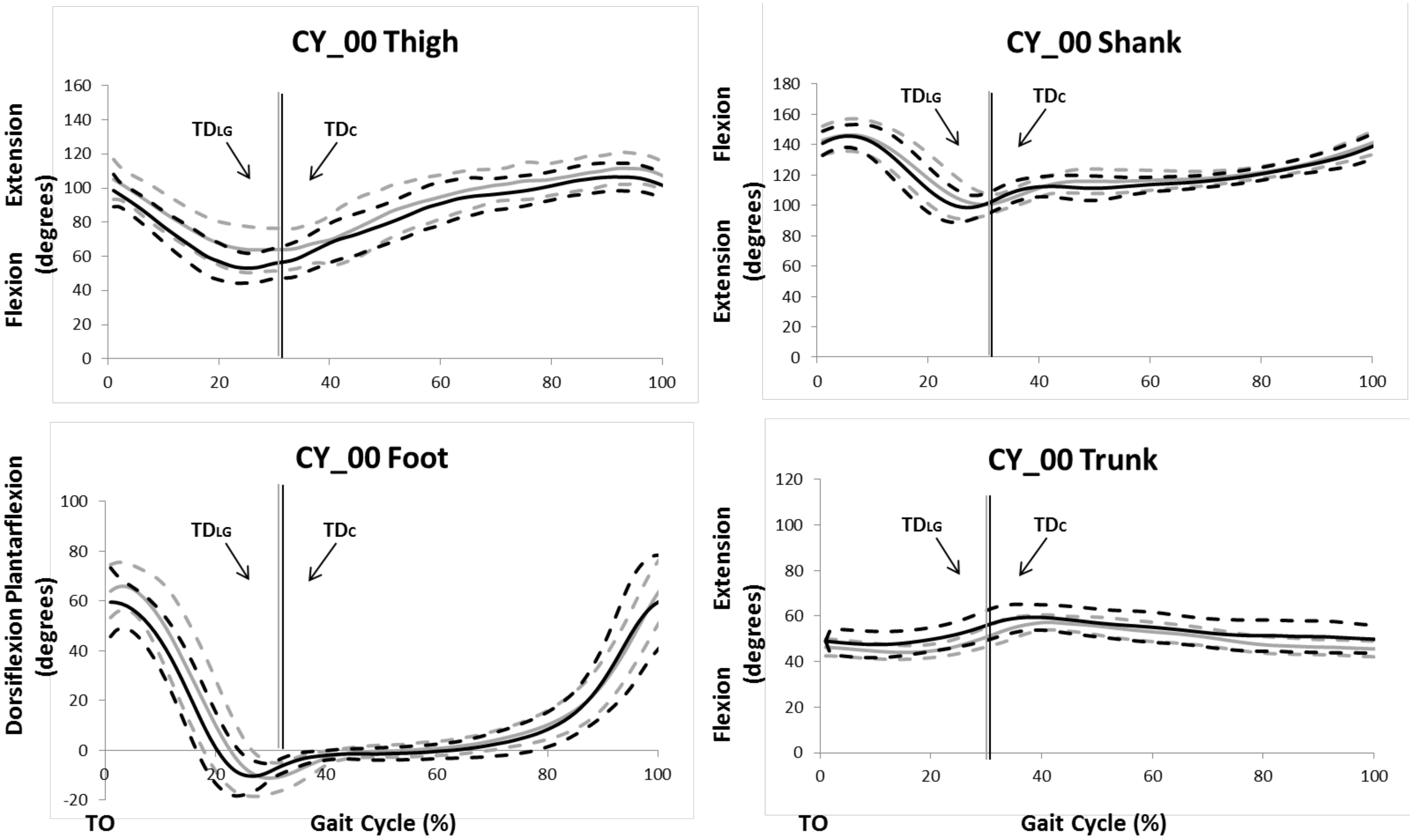


Figure E.1. (continued)

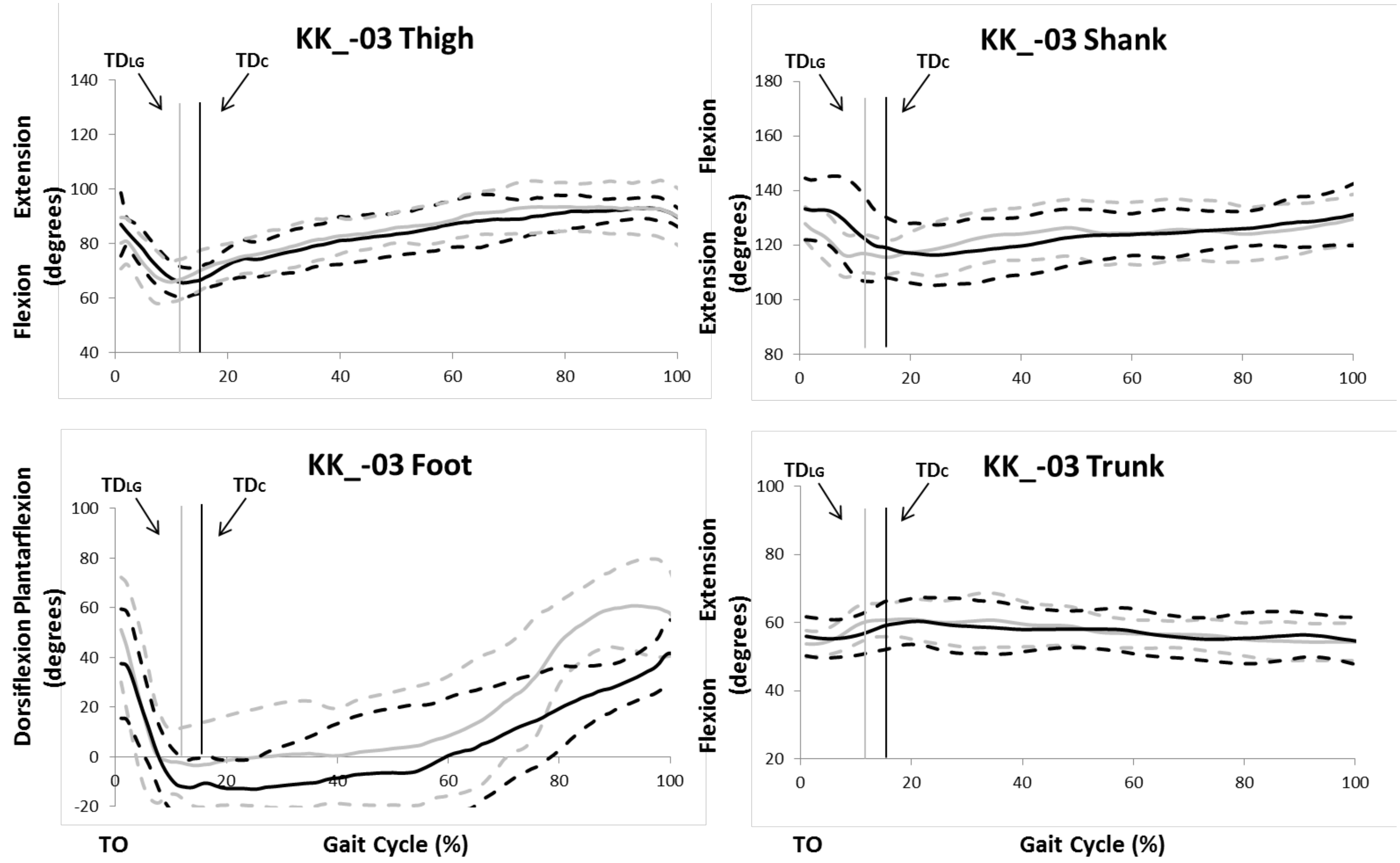


Figure E.1. (continued)

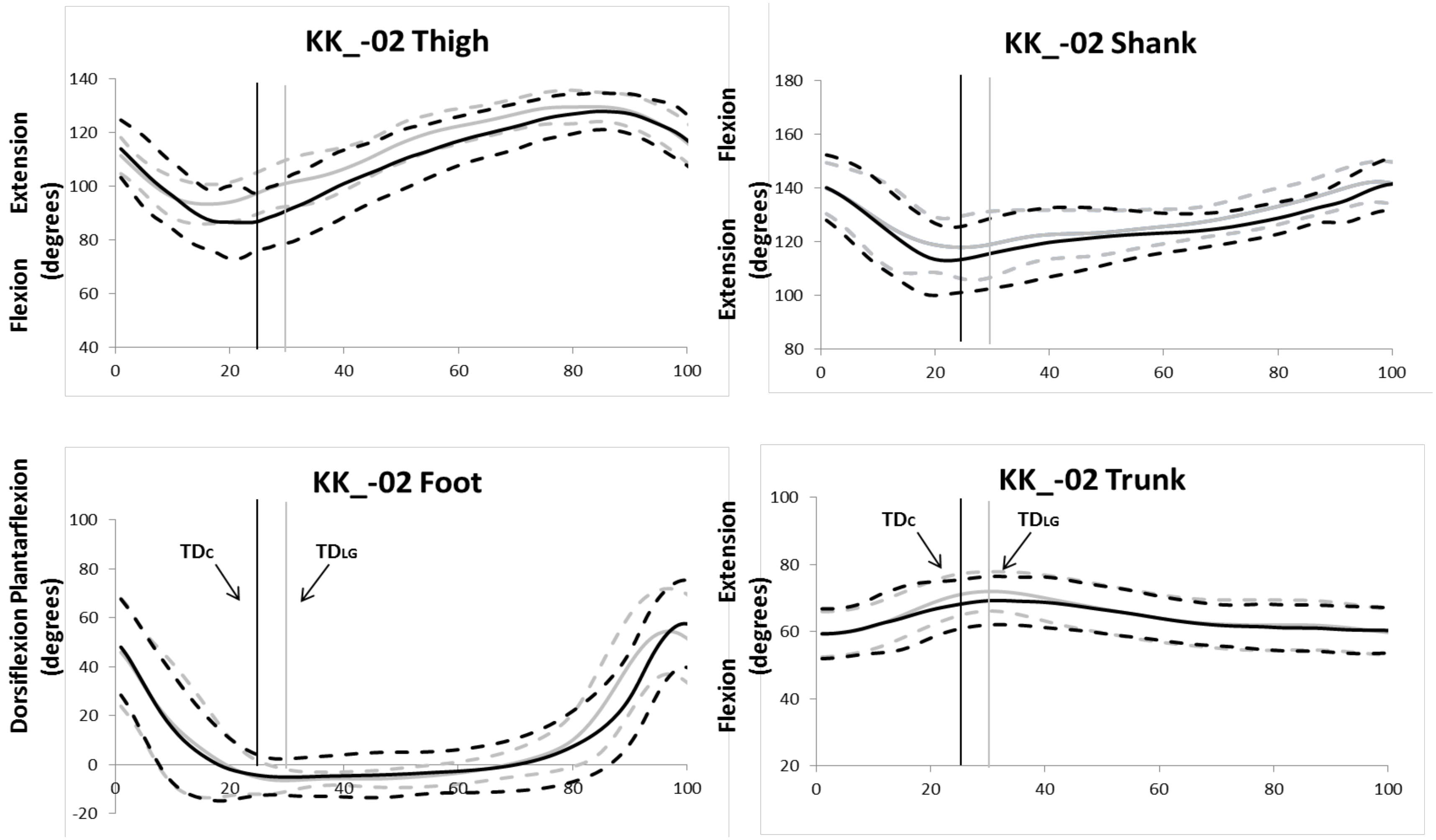


Figure E.1. (continued)

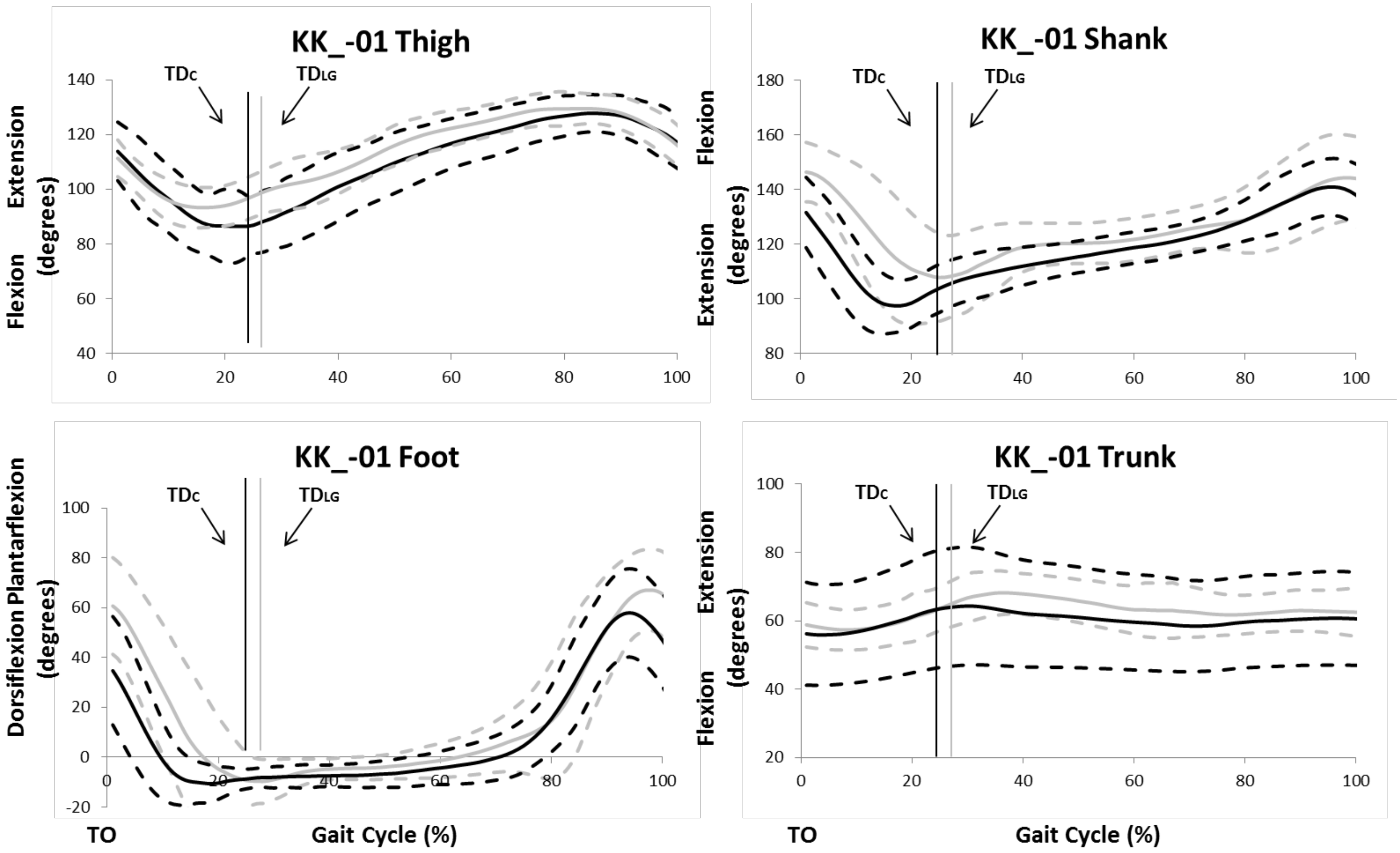
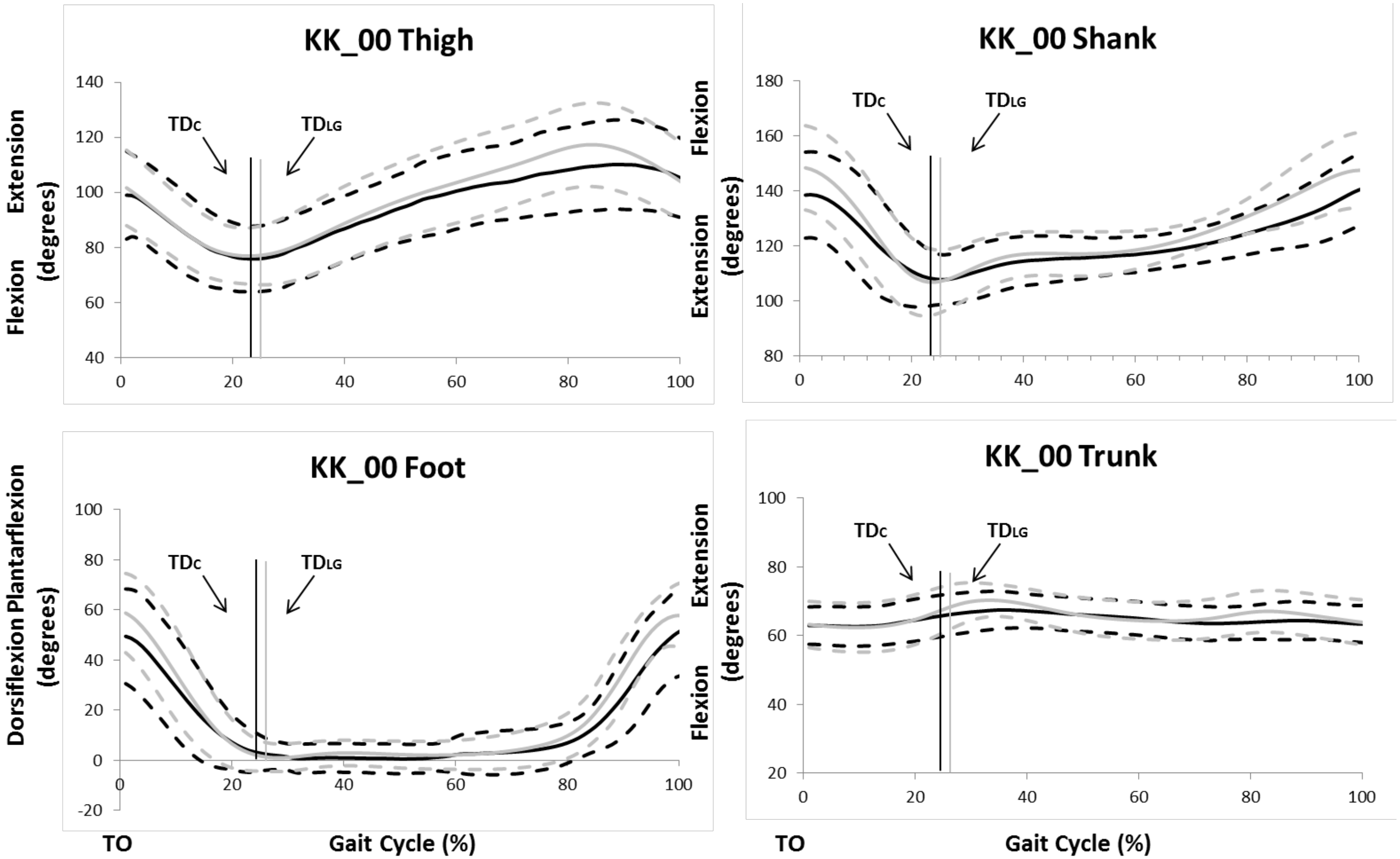


Figure E.1. (continued)



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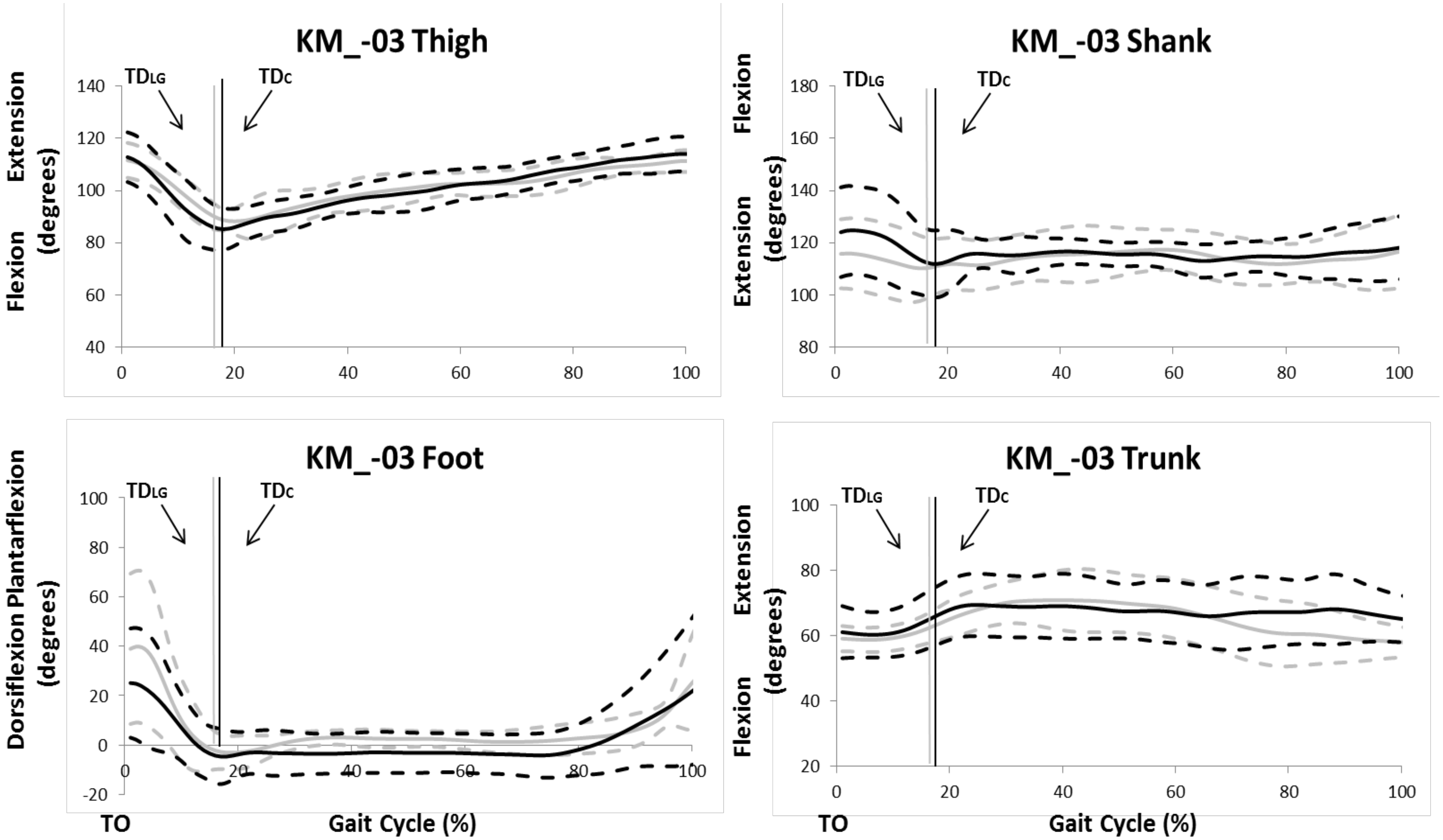

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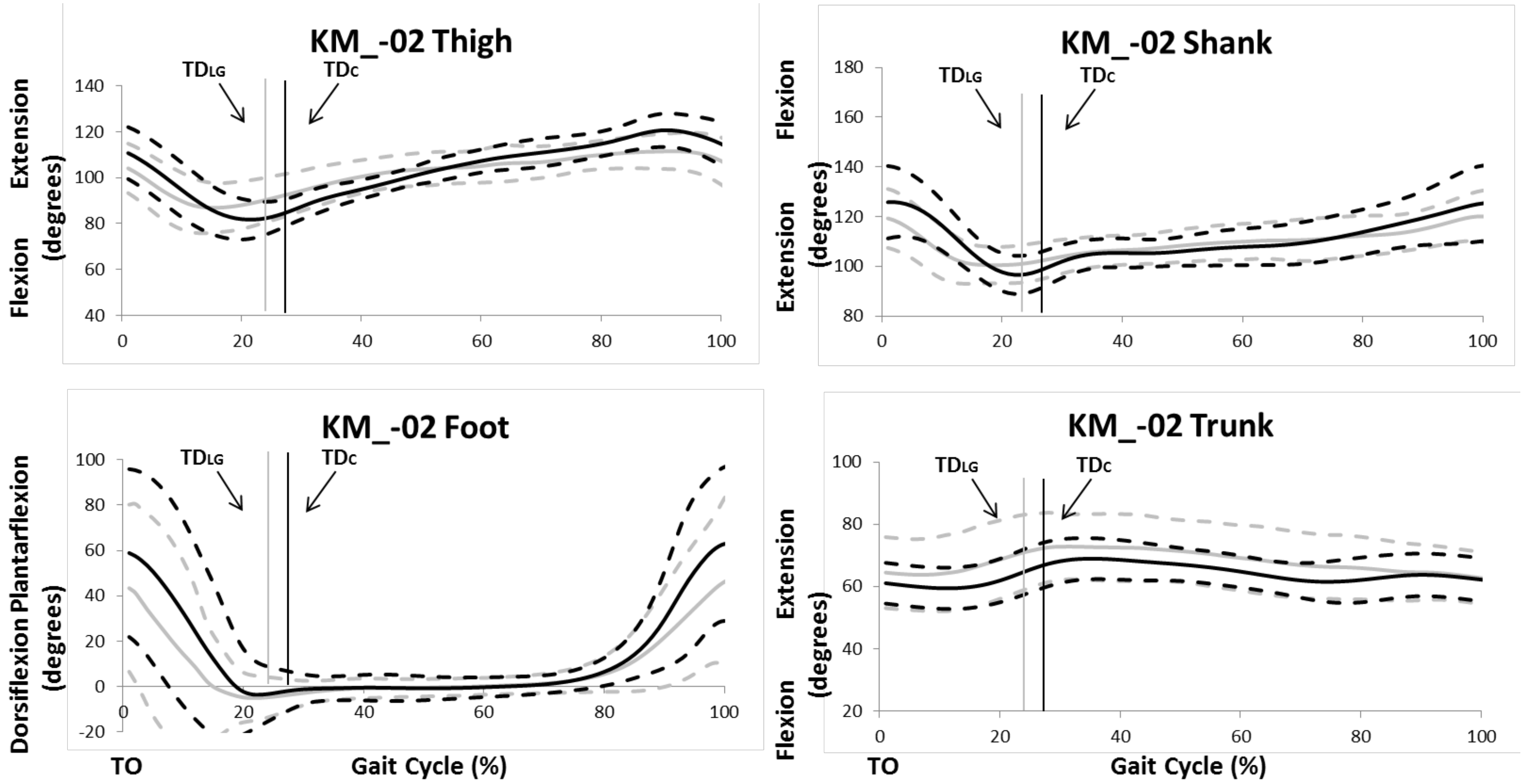

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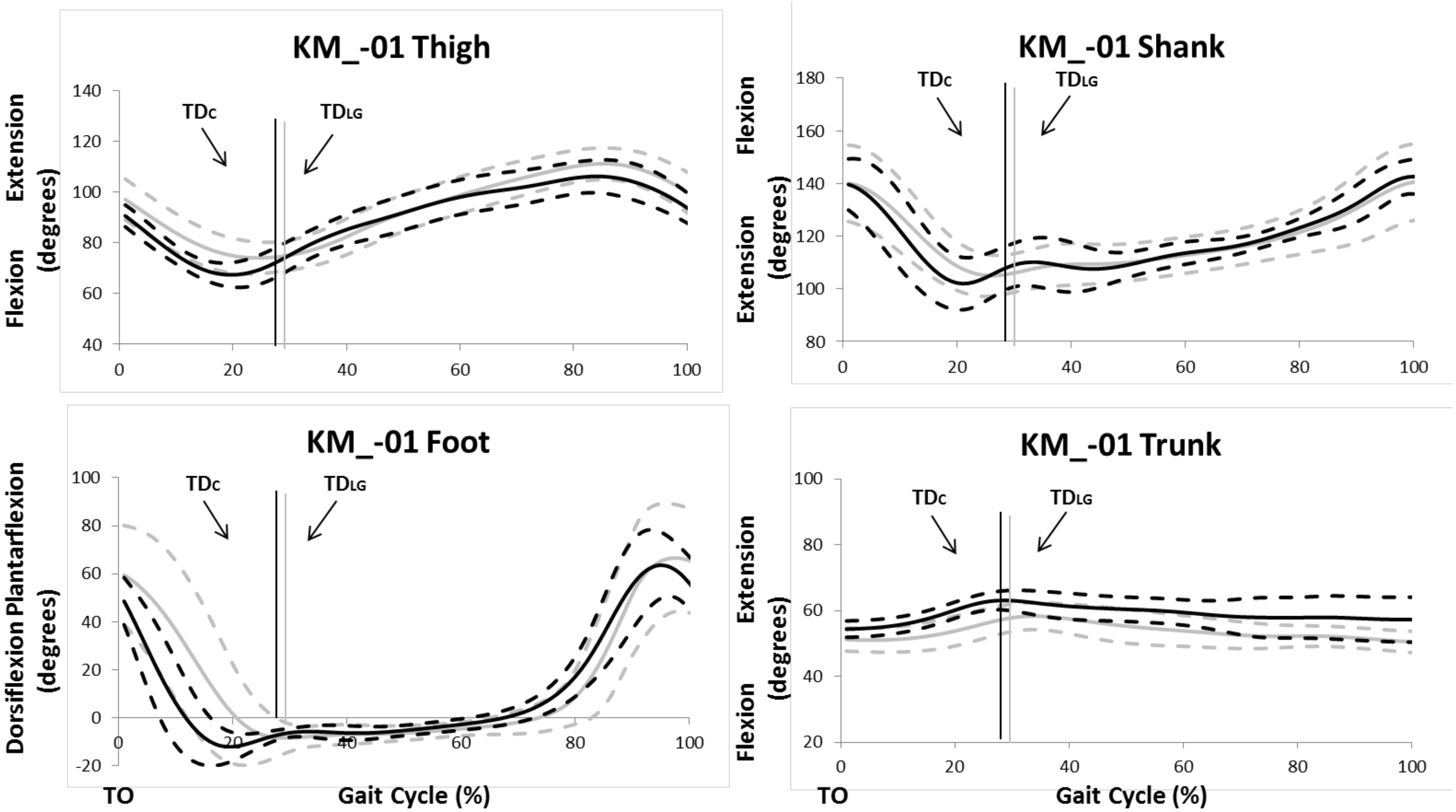

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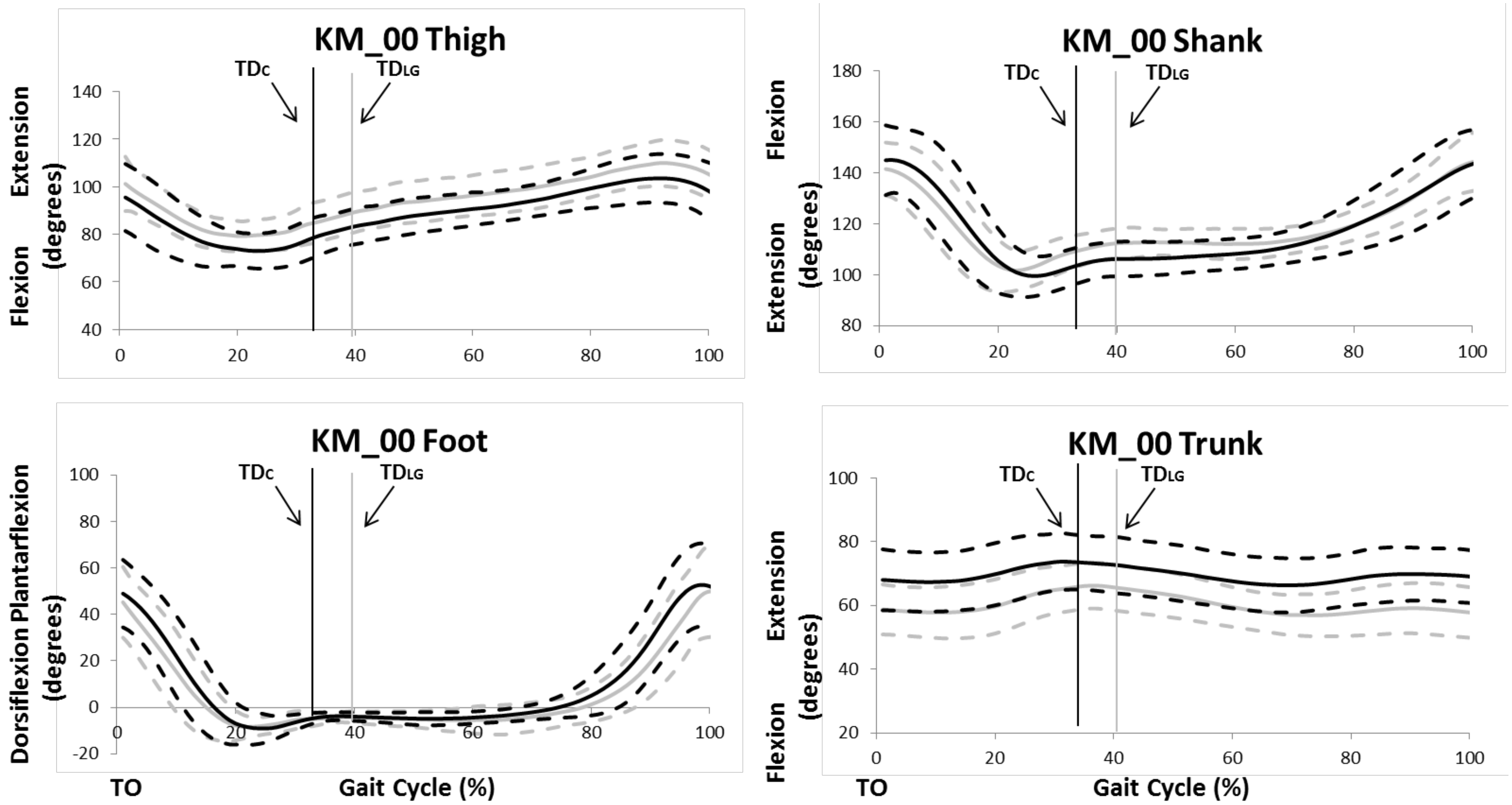

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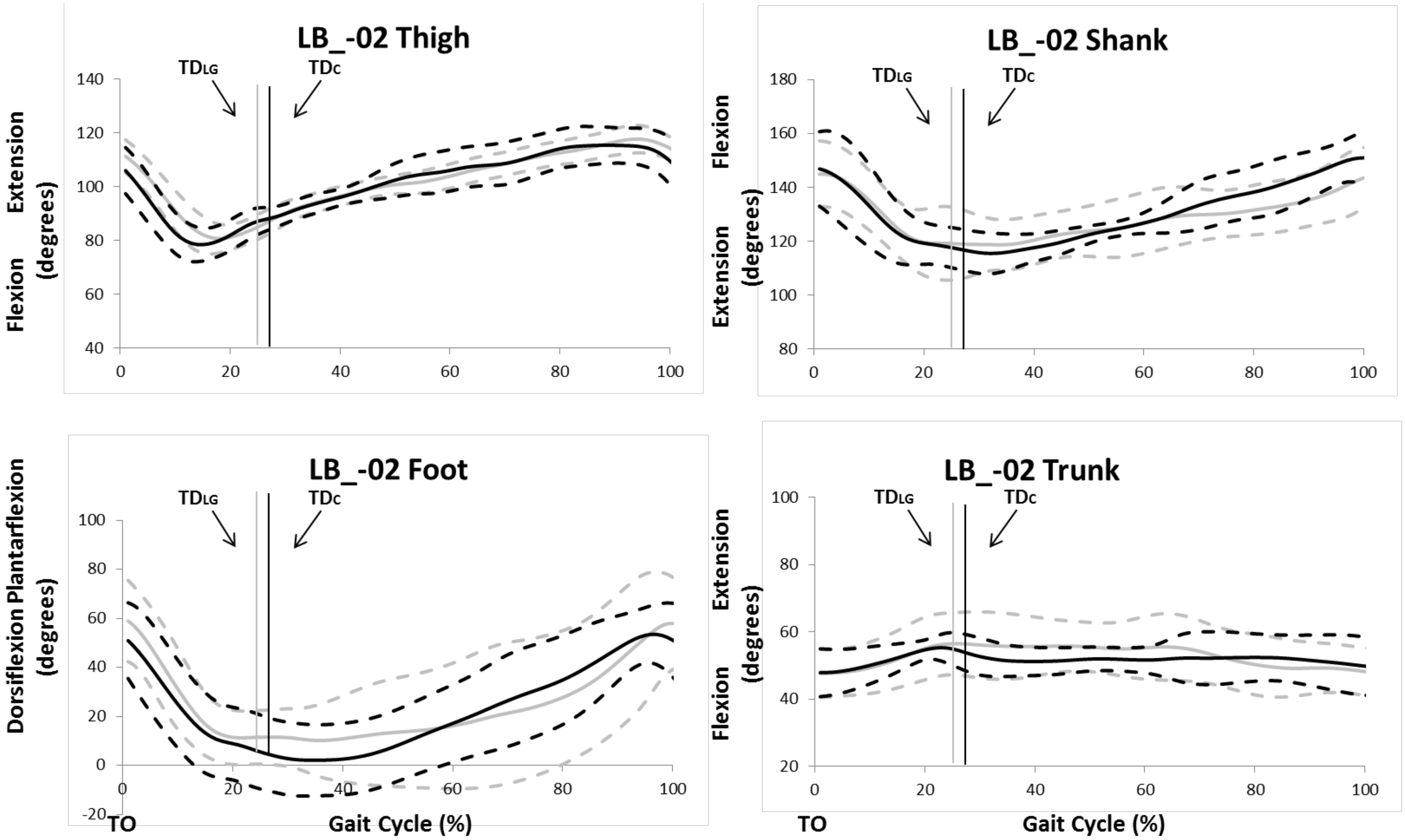

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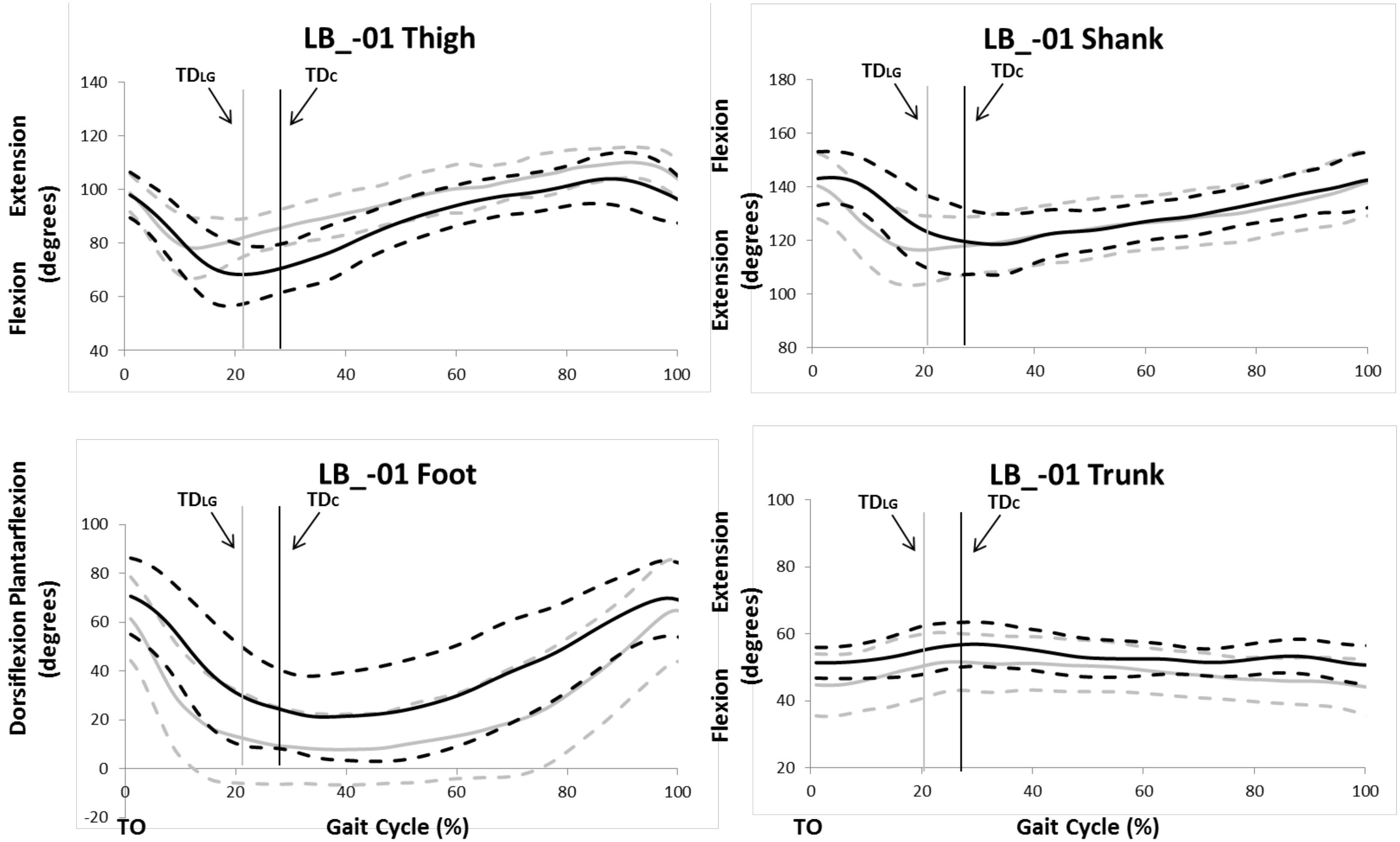


Figure E.1. (continued)

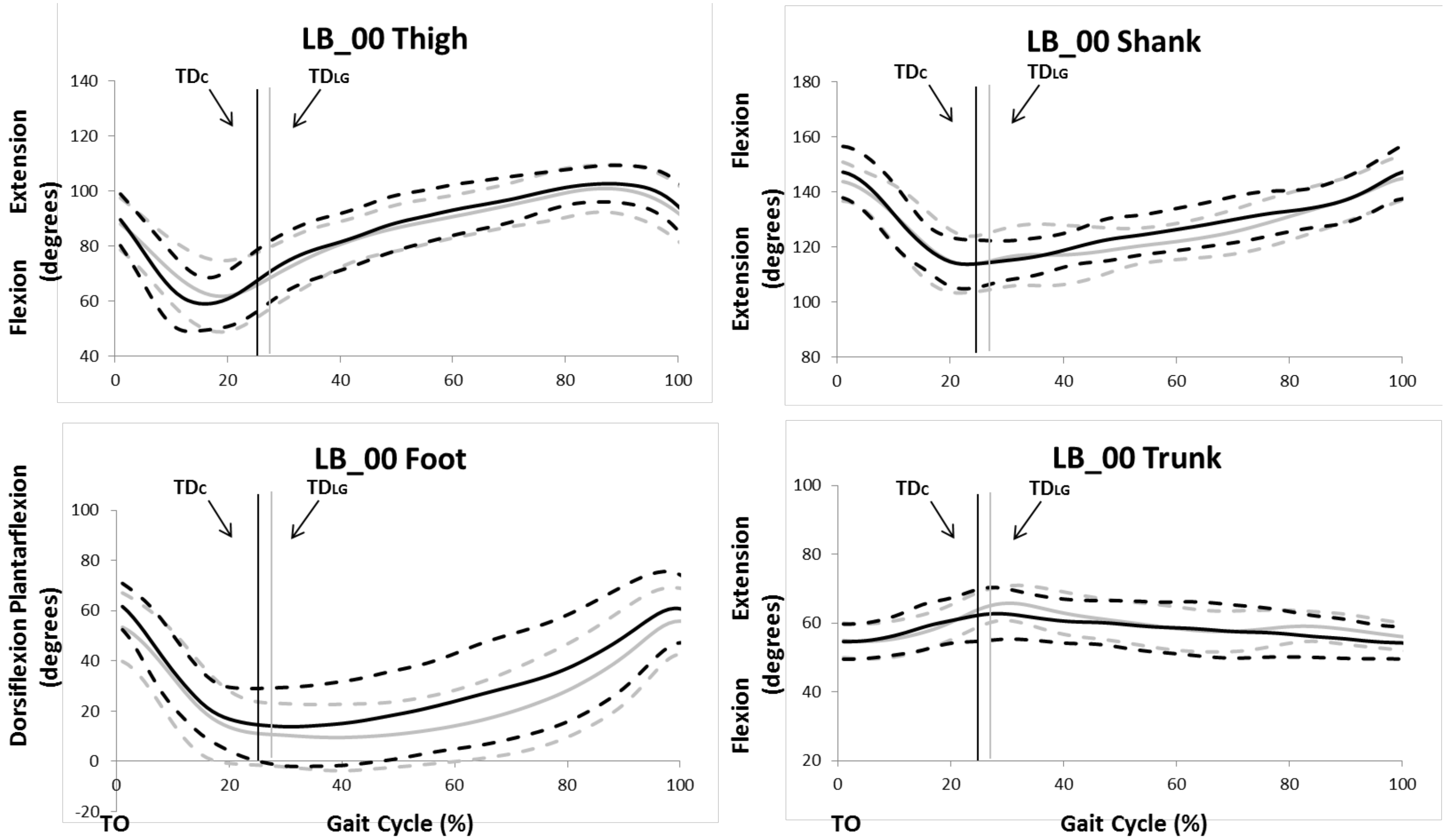
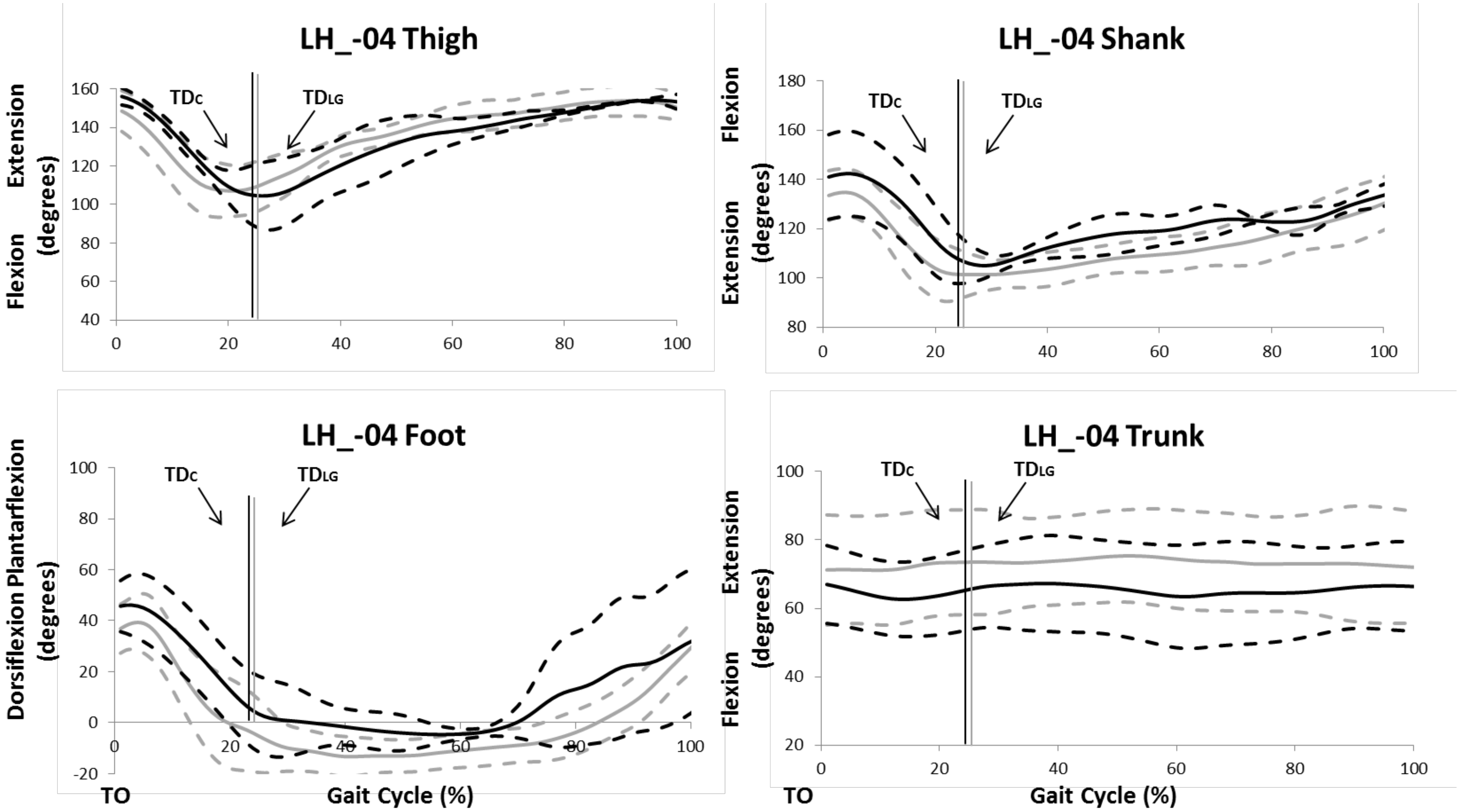


Figure E.1. (continued)



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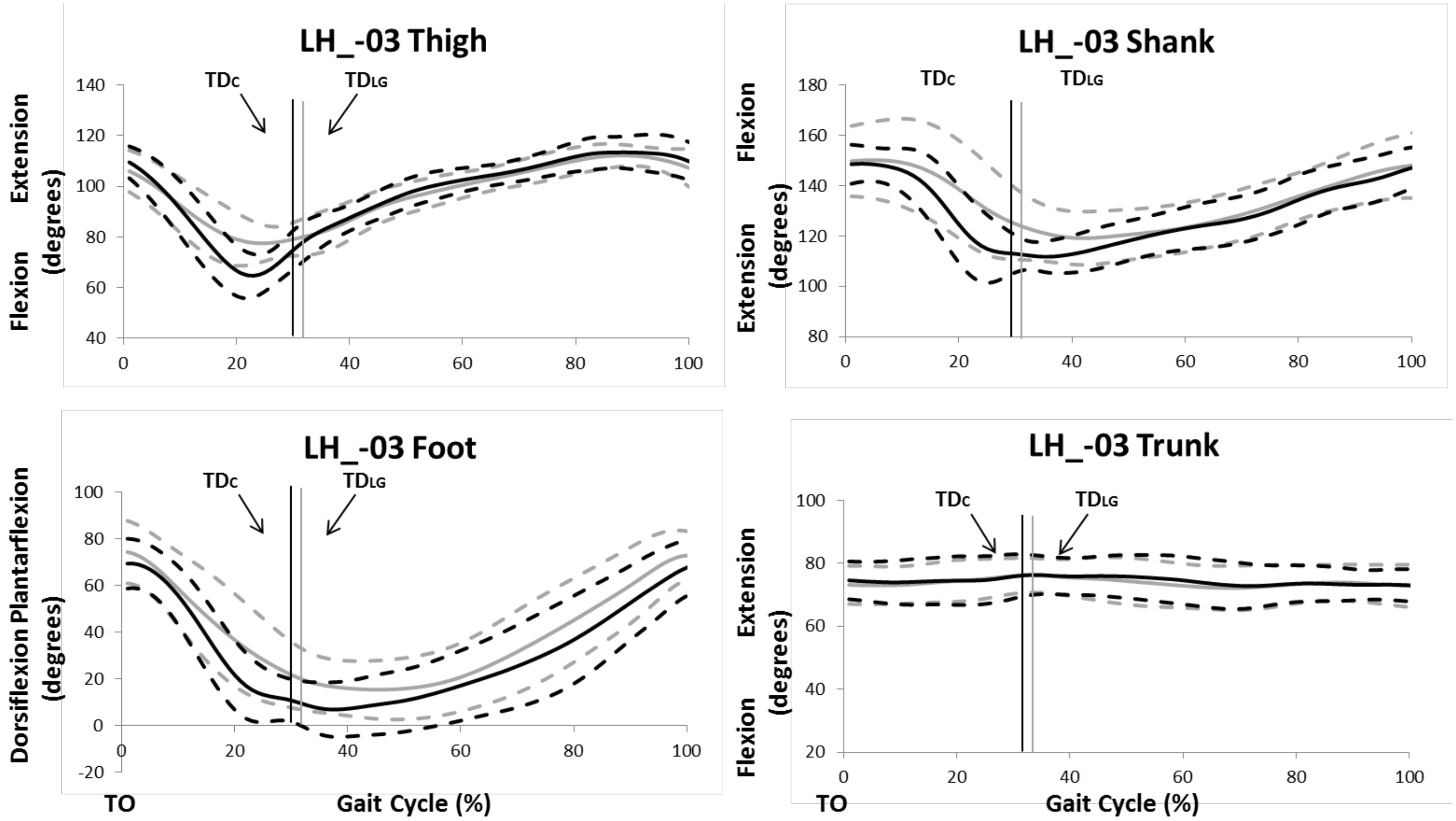
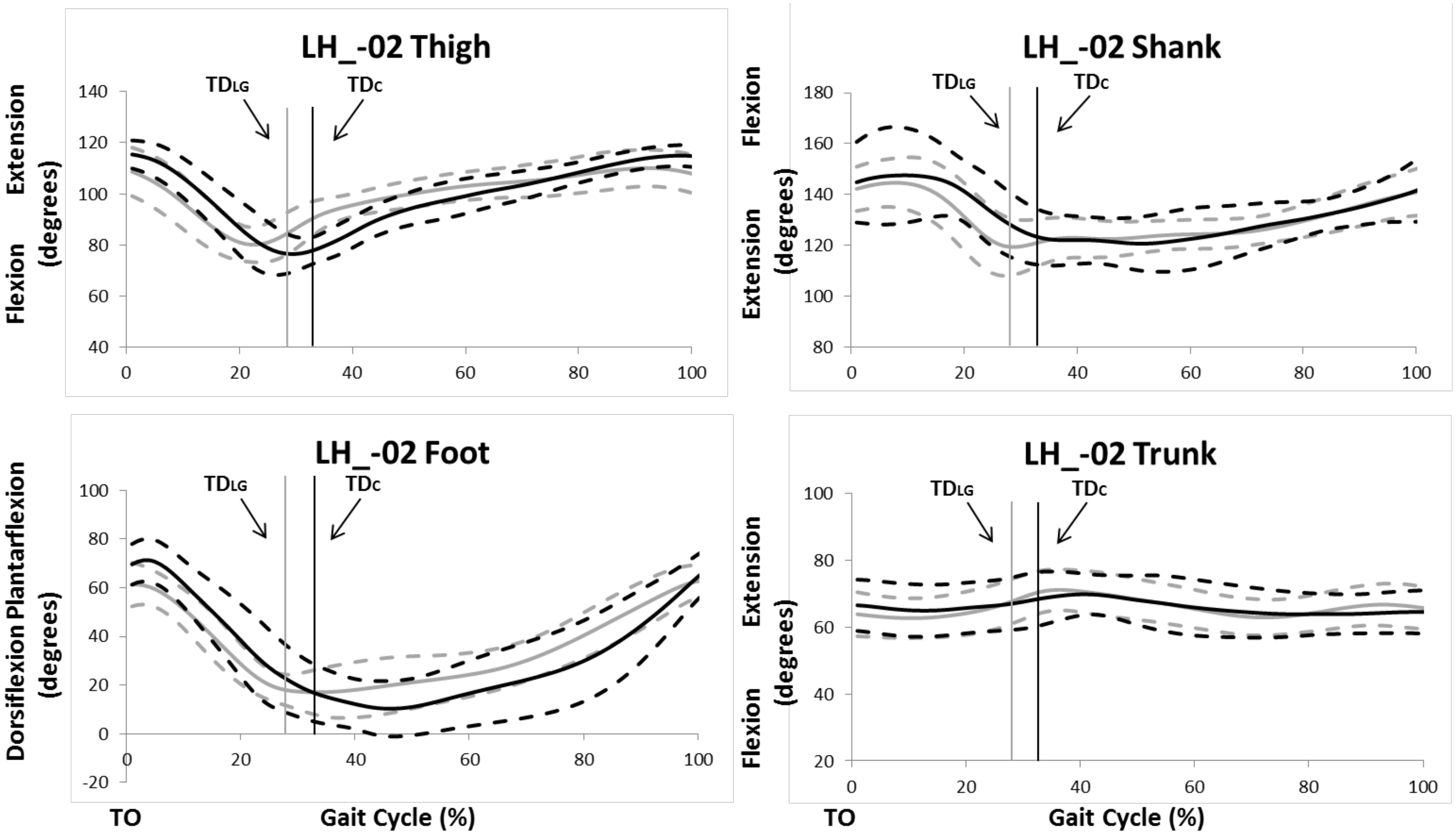


Figure E.1. (continued)



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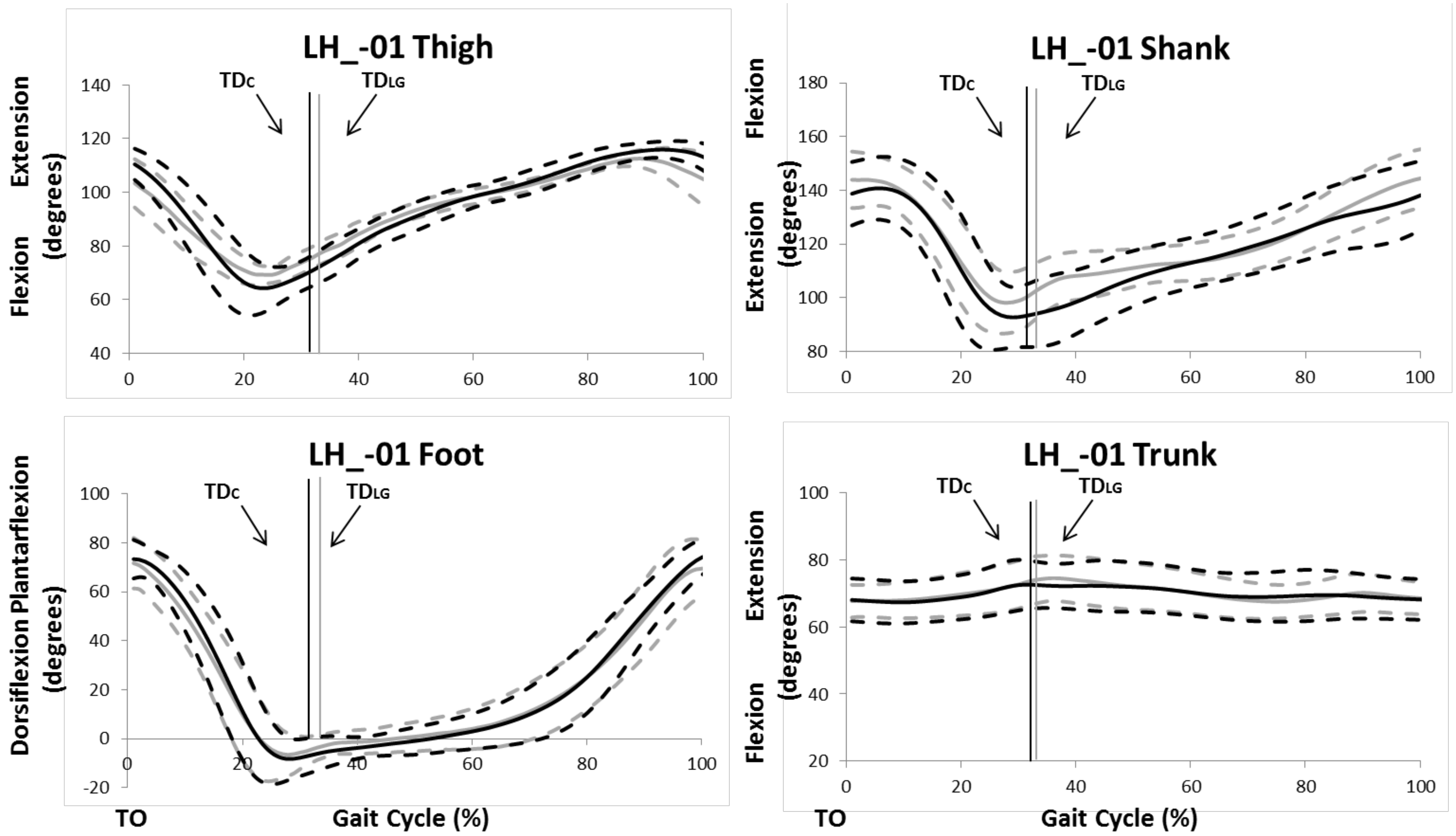

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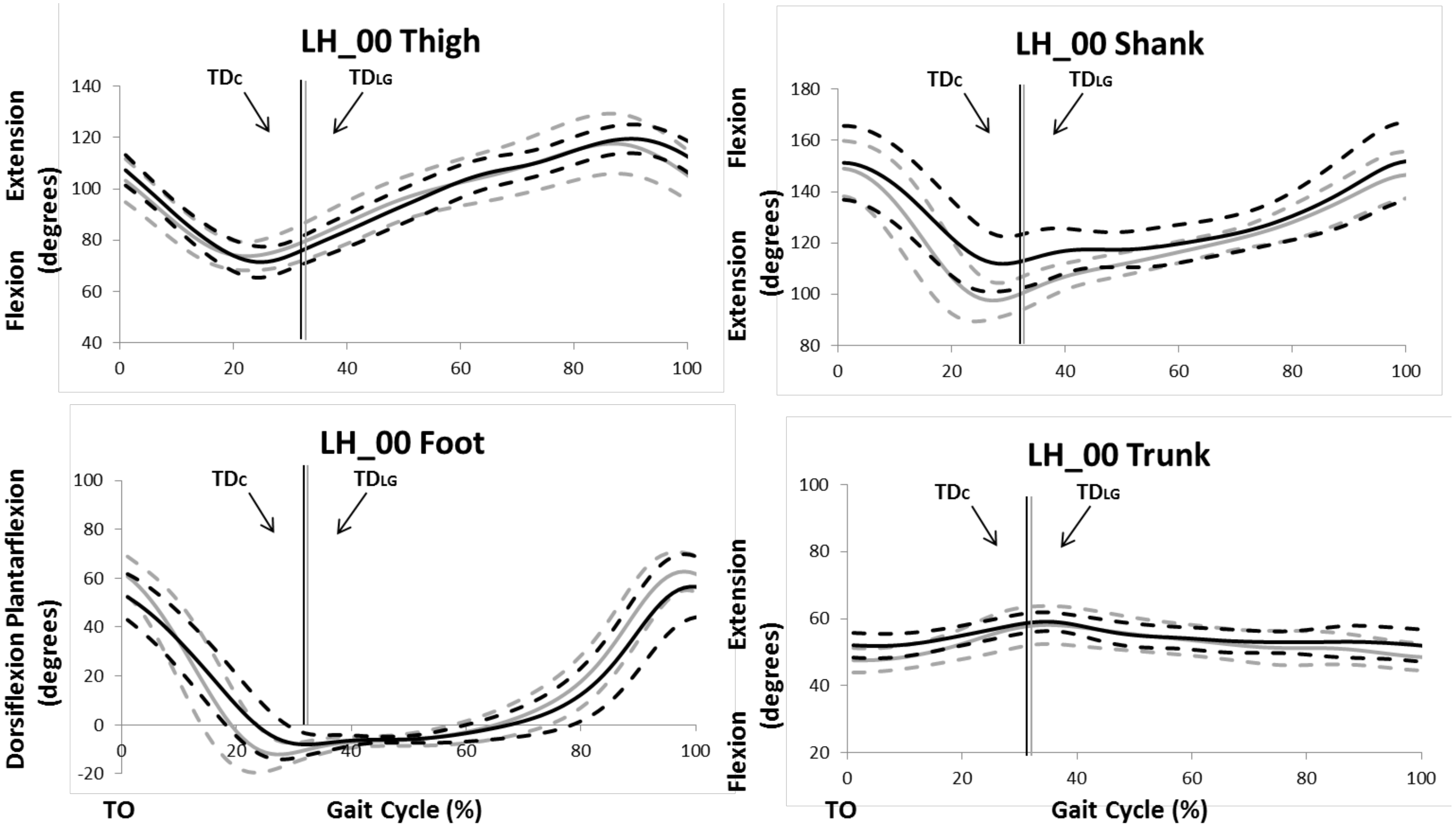

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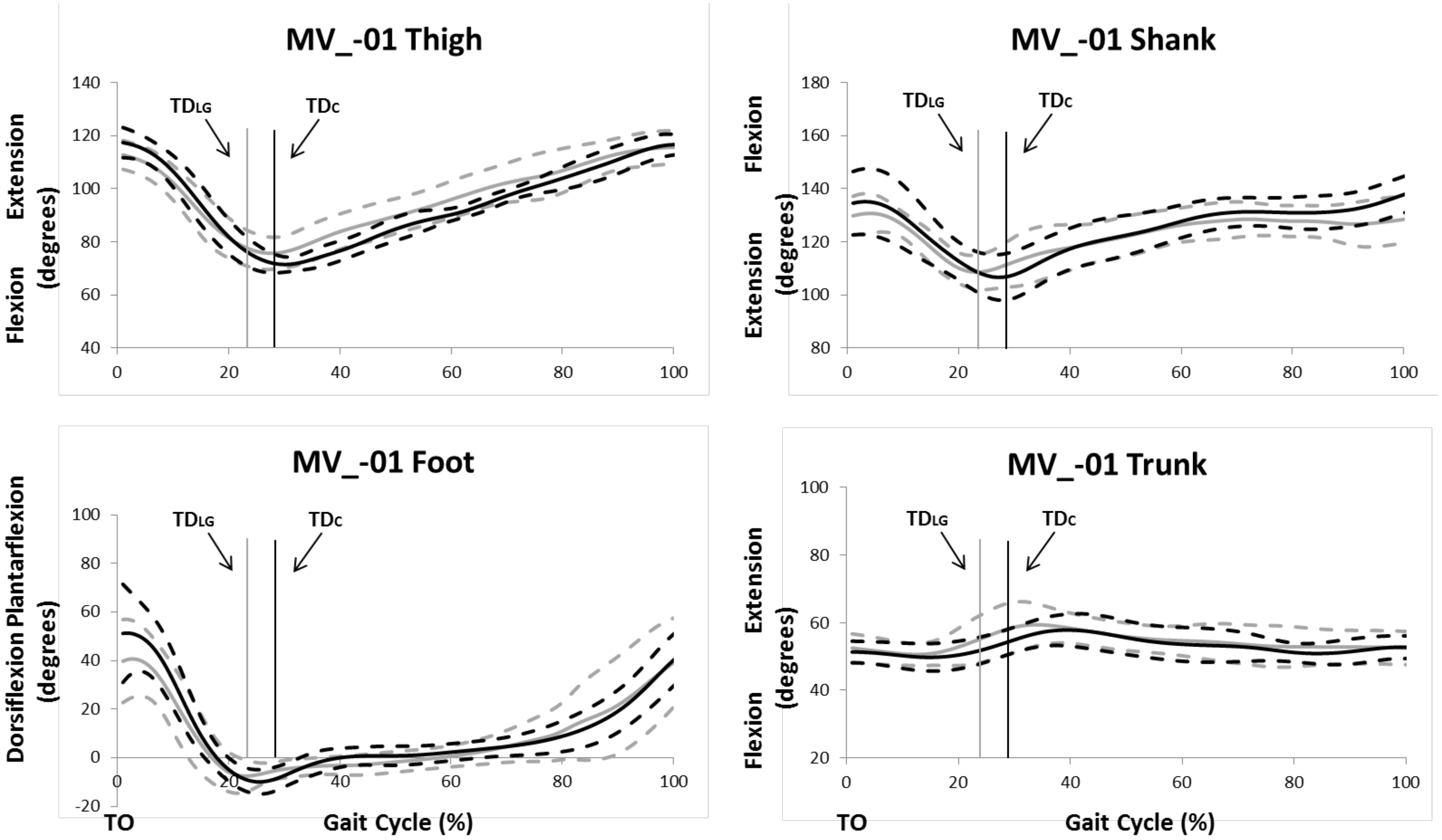

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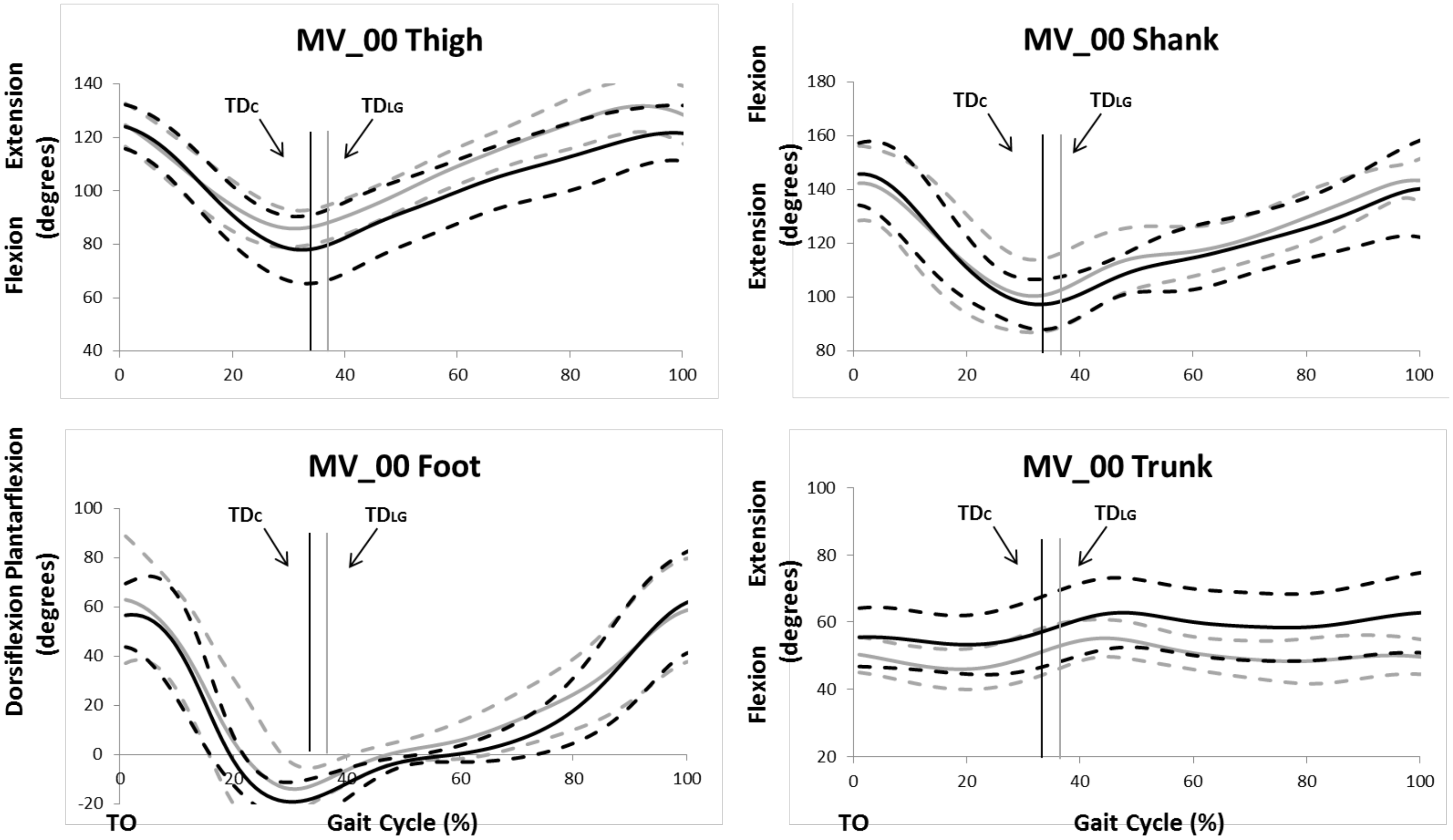

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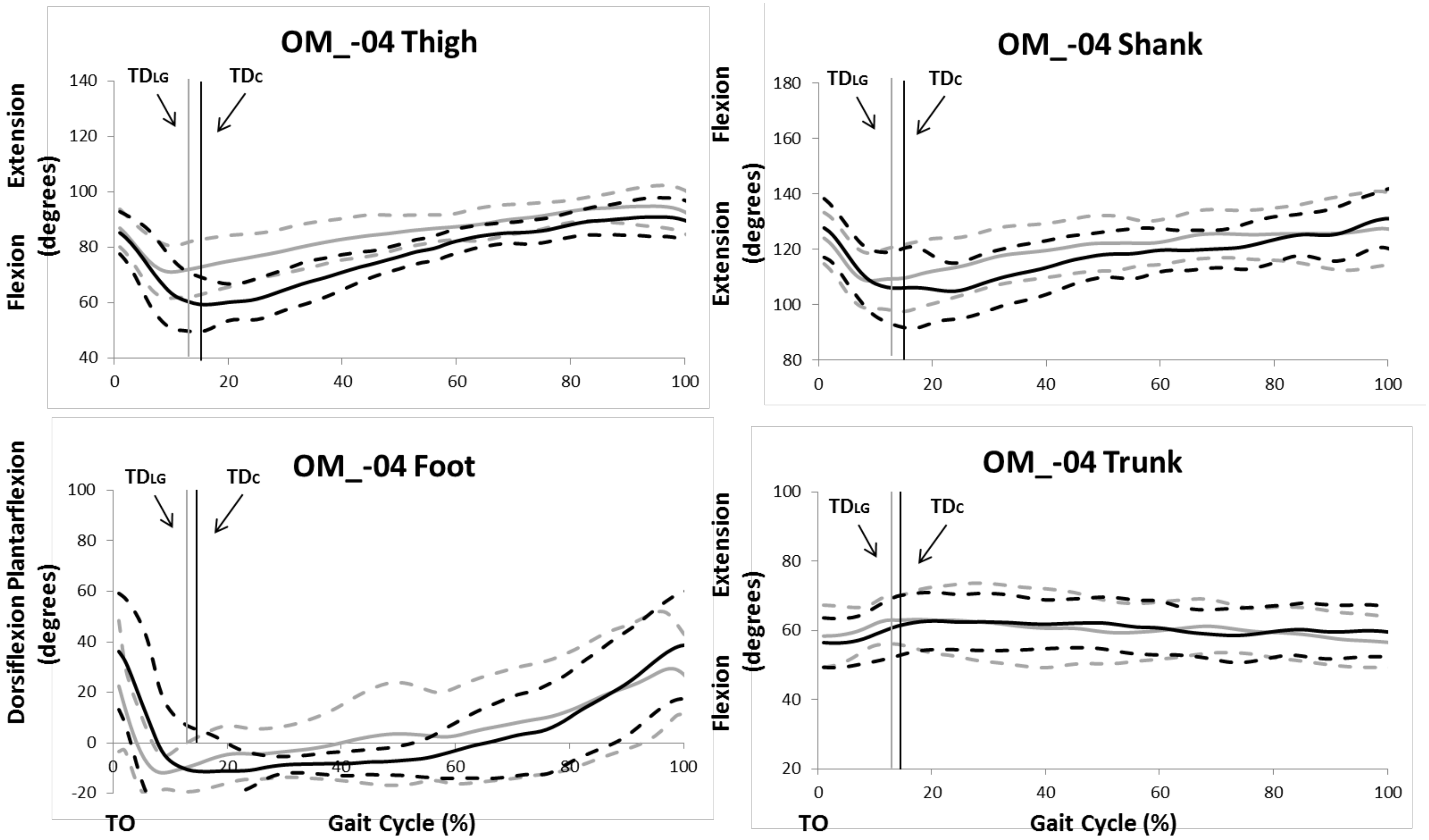

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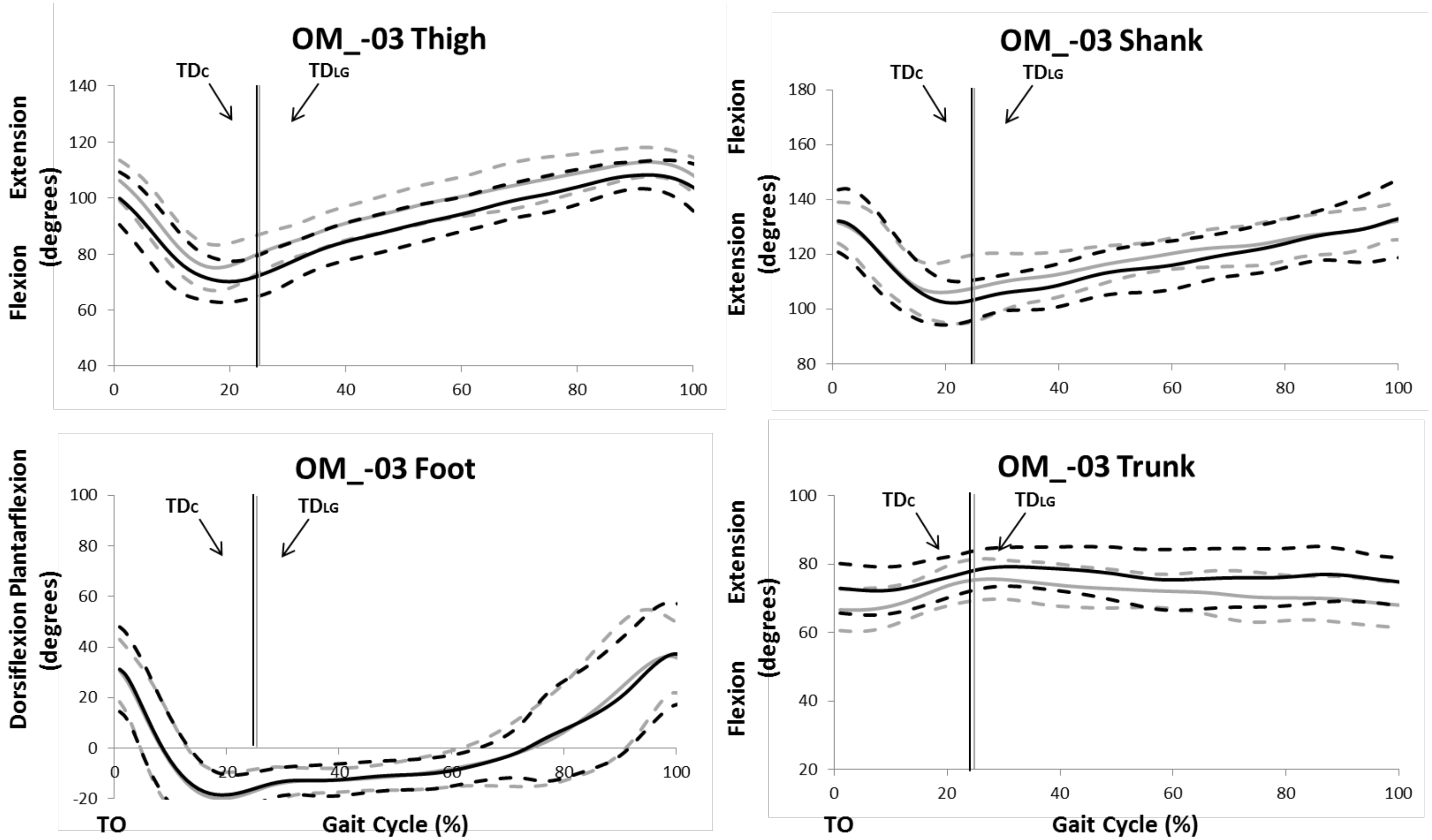

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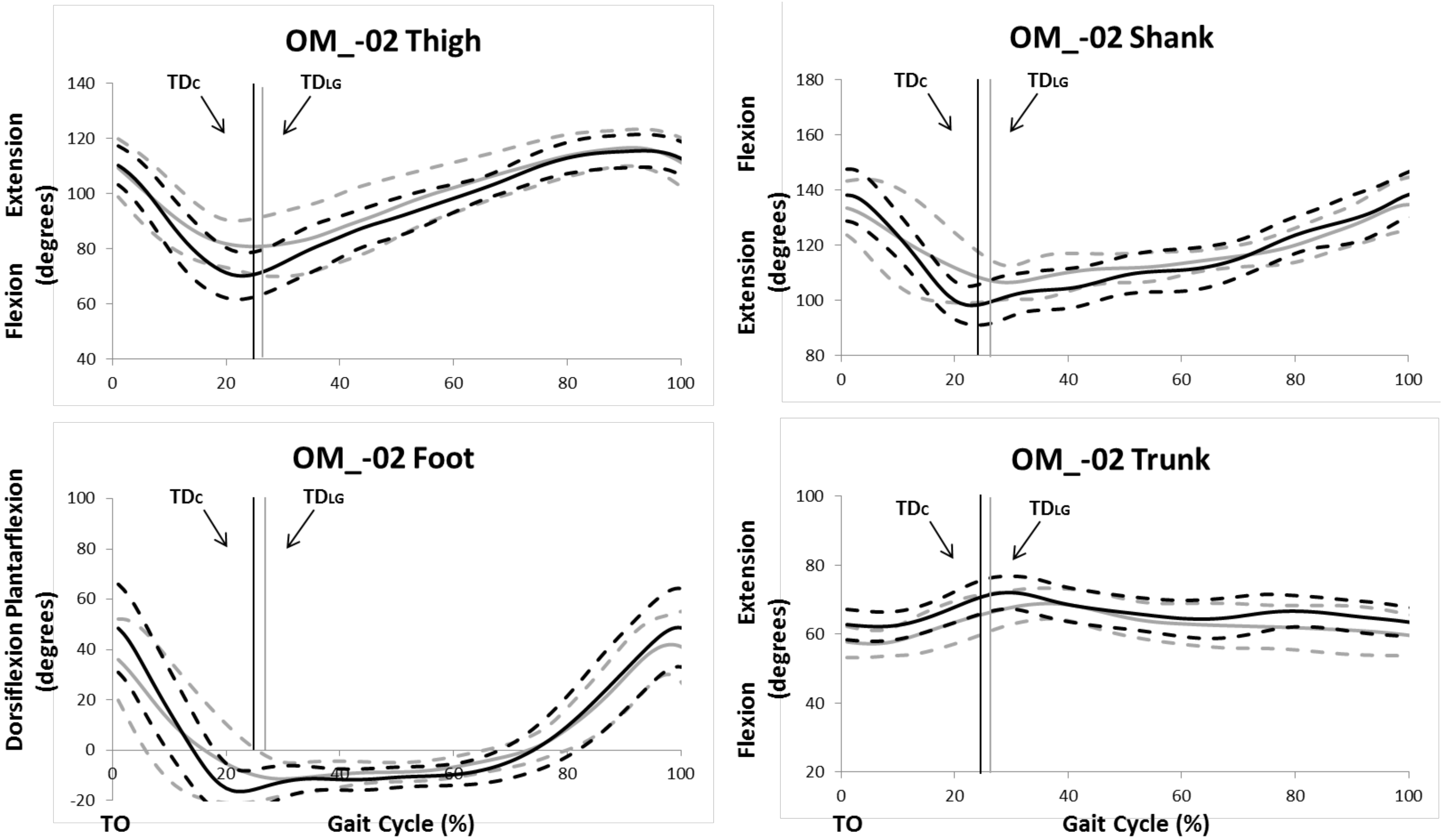

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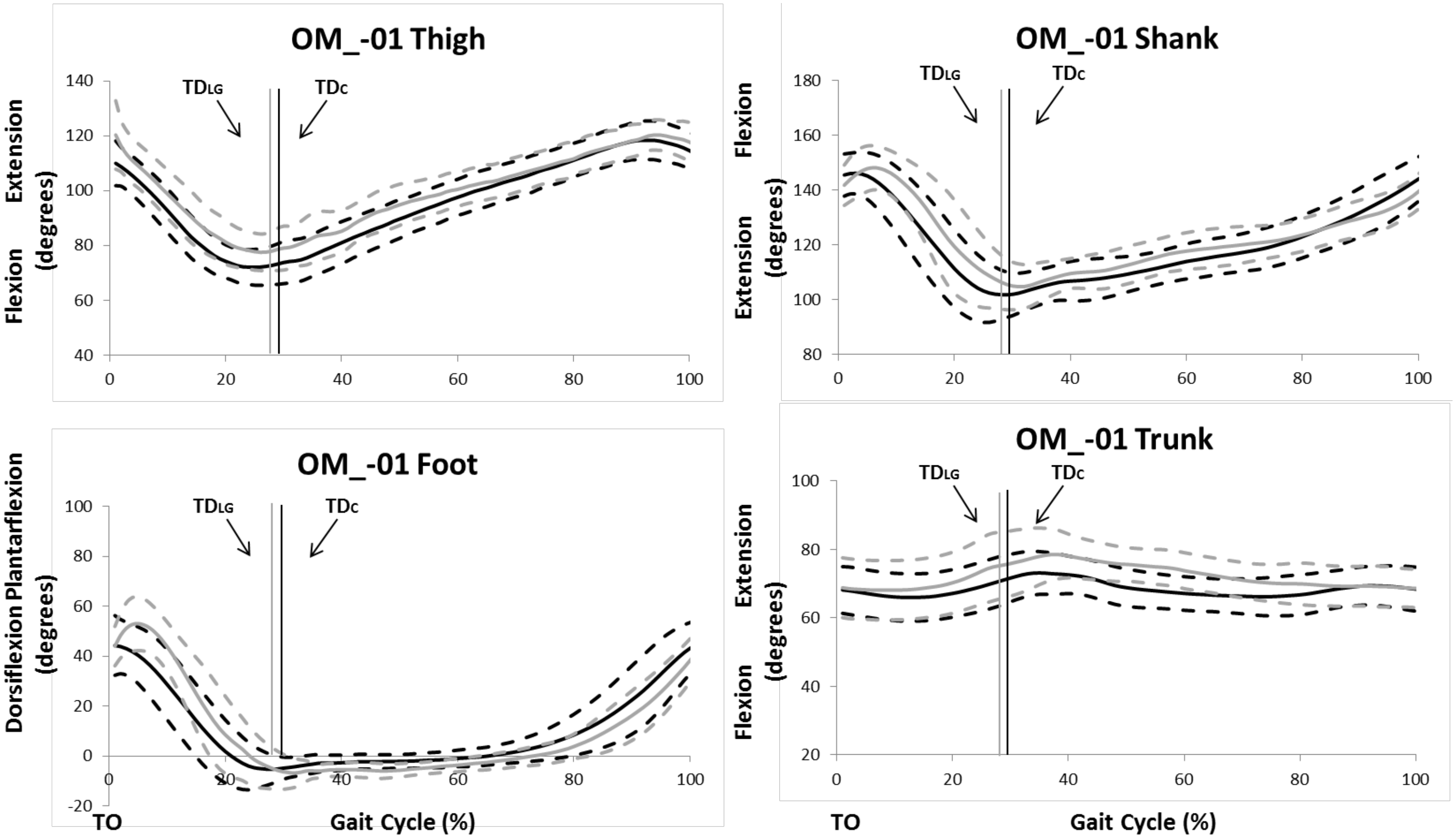

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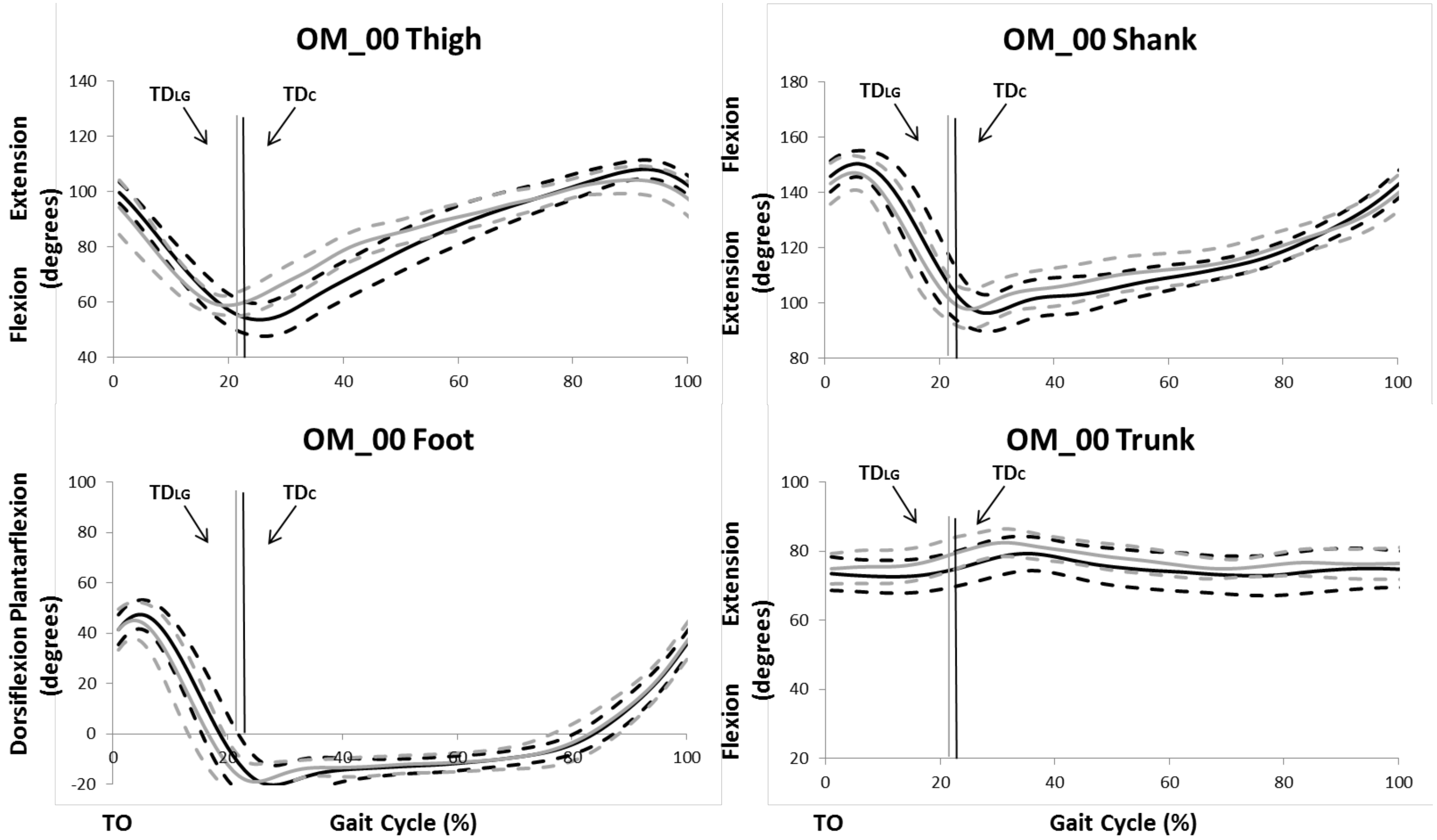


Figure E.1. (continued)

Appendix F.

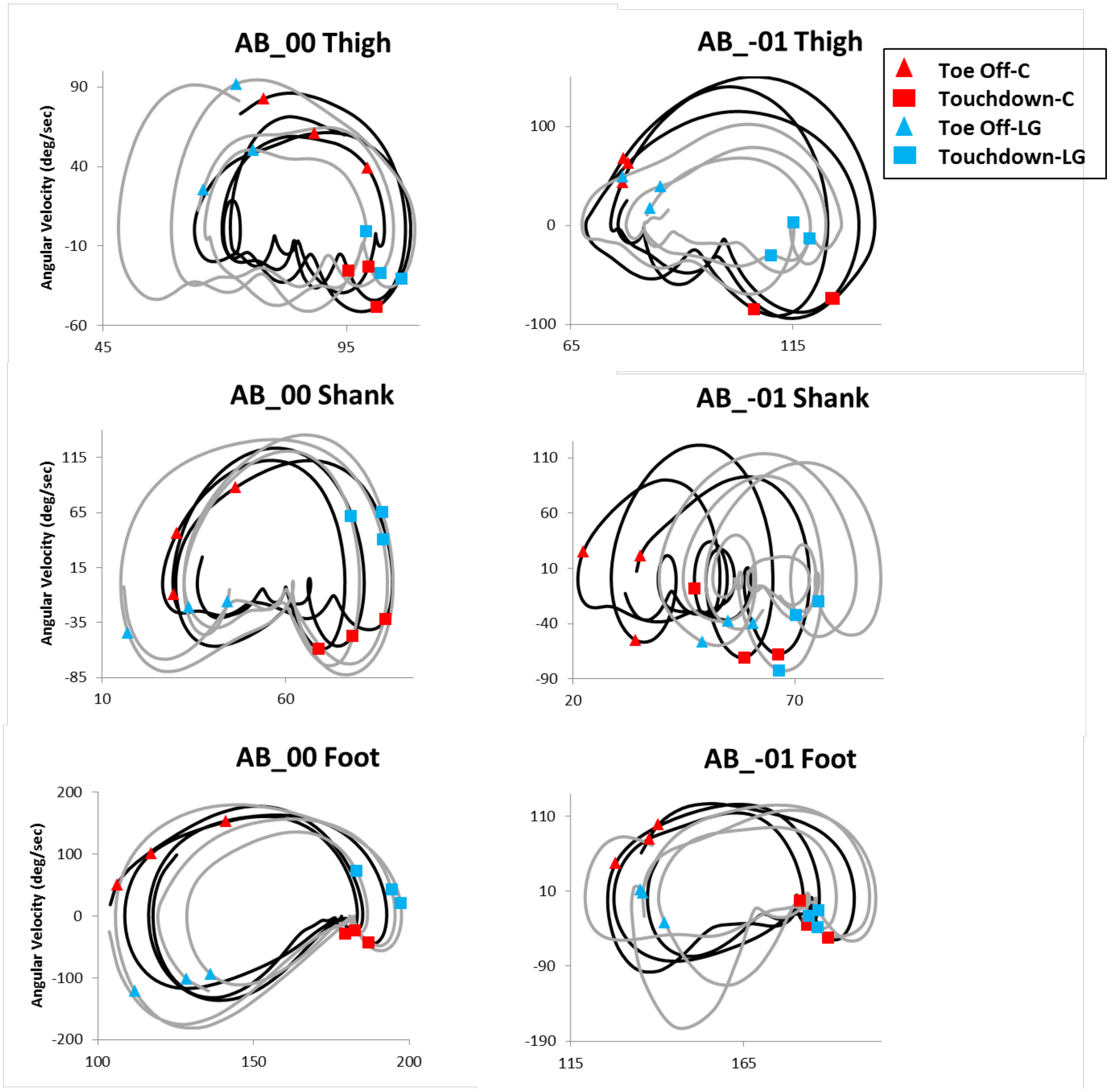
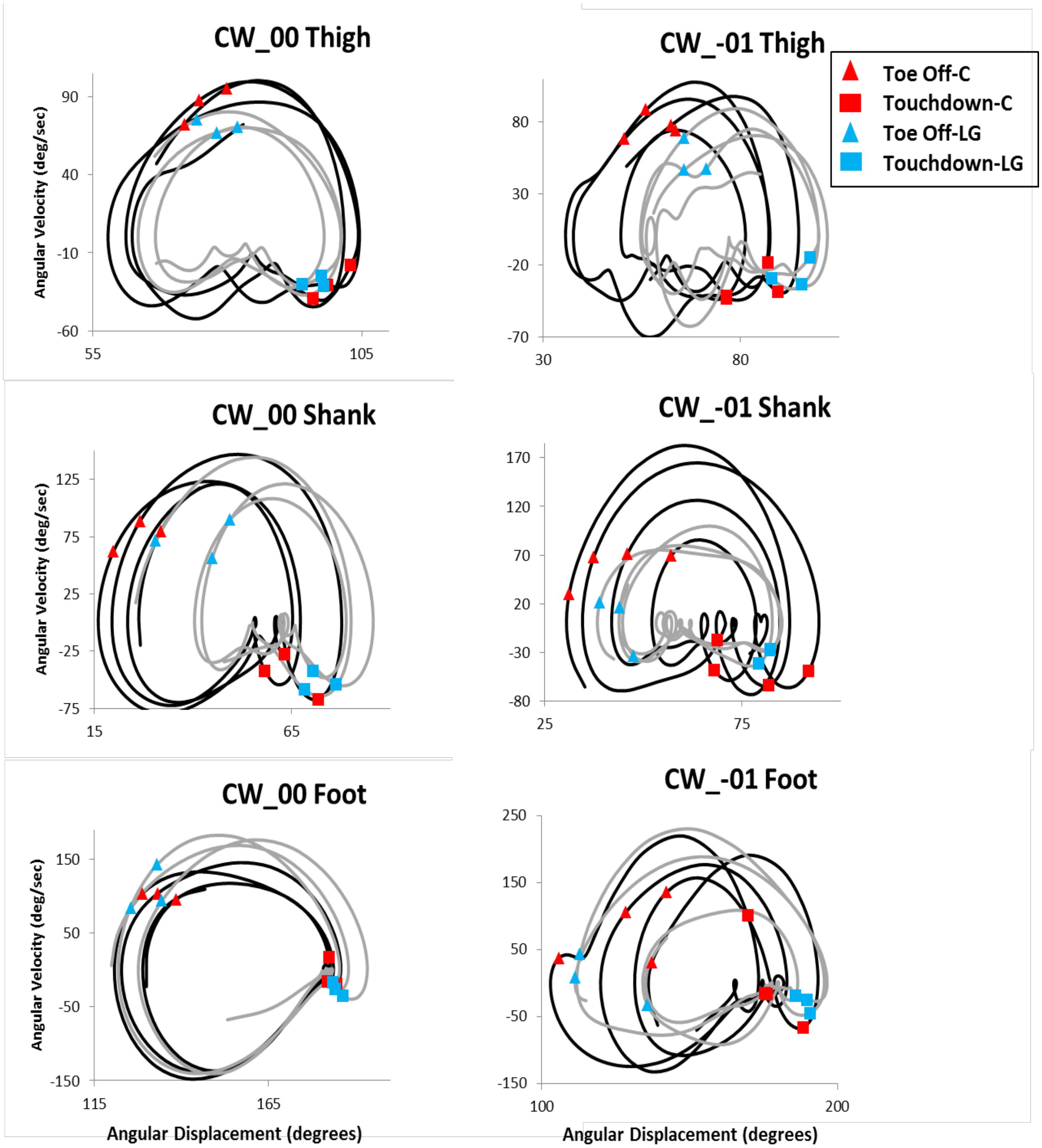


Figure F.1. Phase portrait plots for 3 consecutive cruising strides per condition for infants across visits for thigh, shank, and foot segments. Trajectories unfold in clockwise direction over the cycle duration. — Control (diaper-only); — Lycra Garment.



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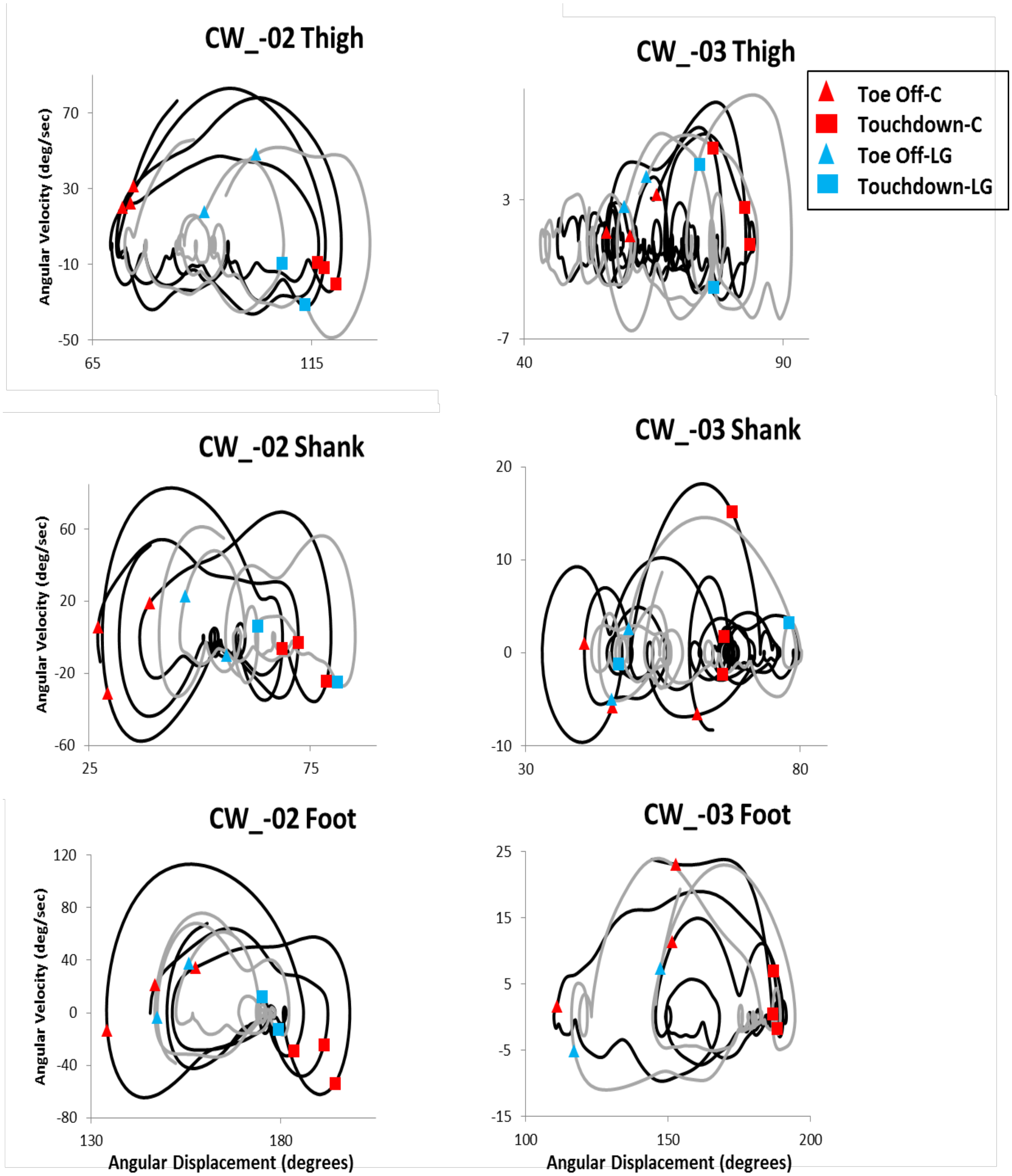
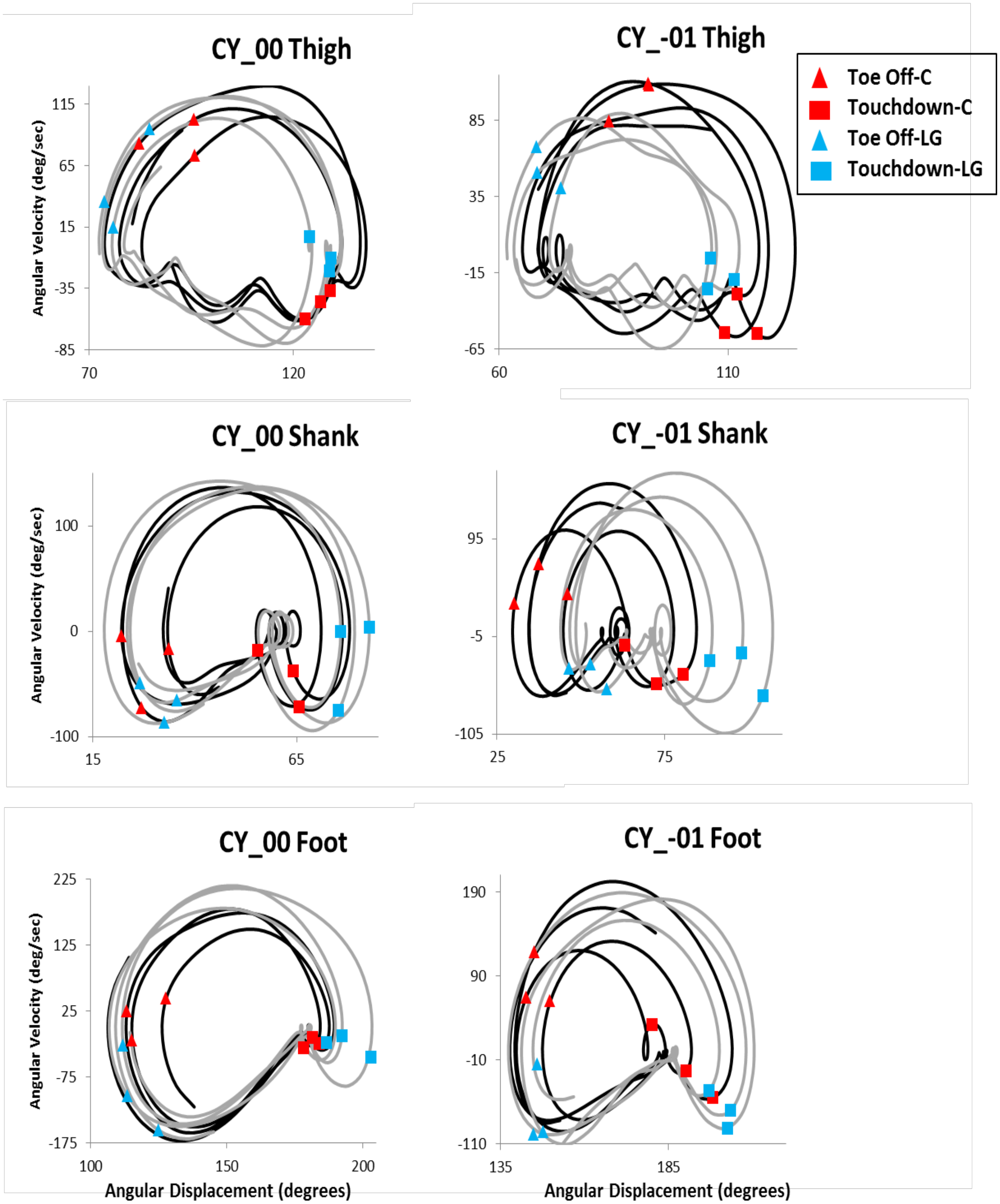


Figure F.1. (continued)



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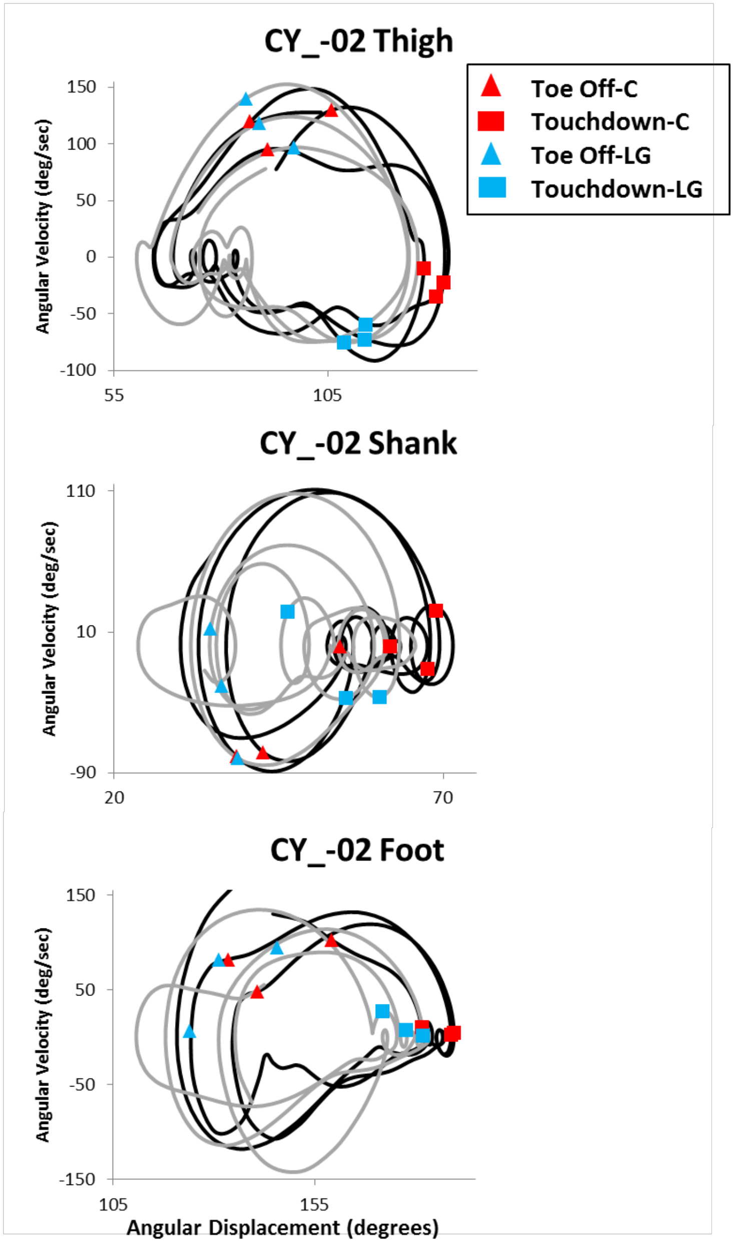

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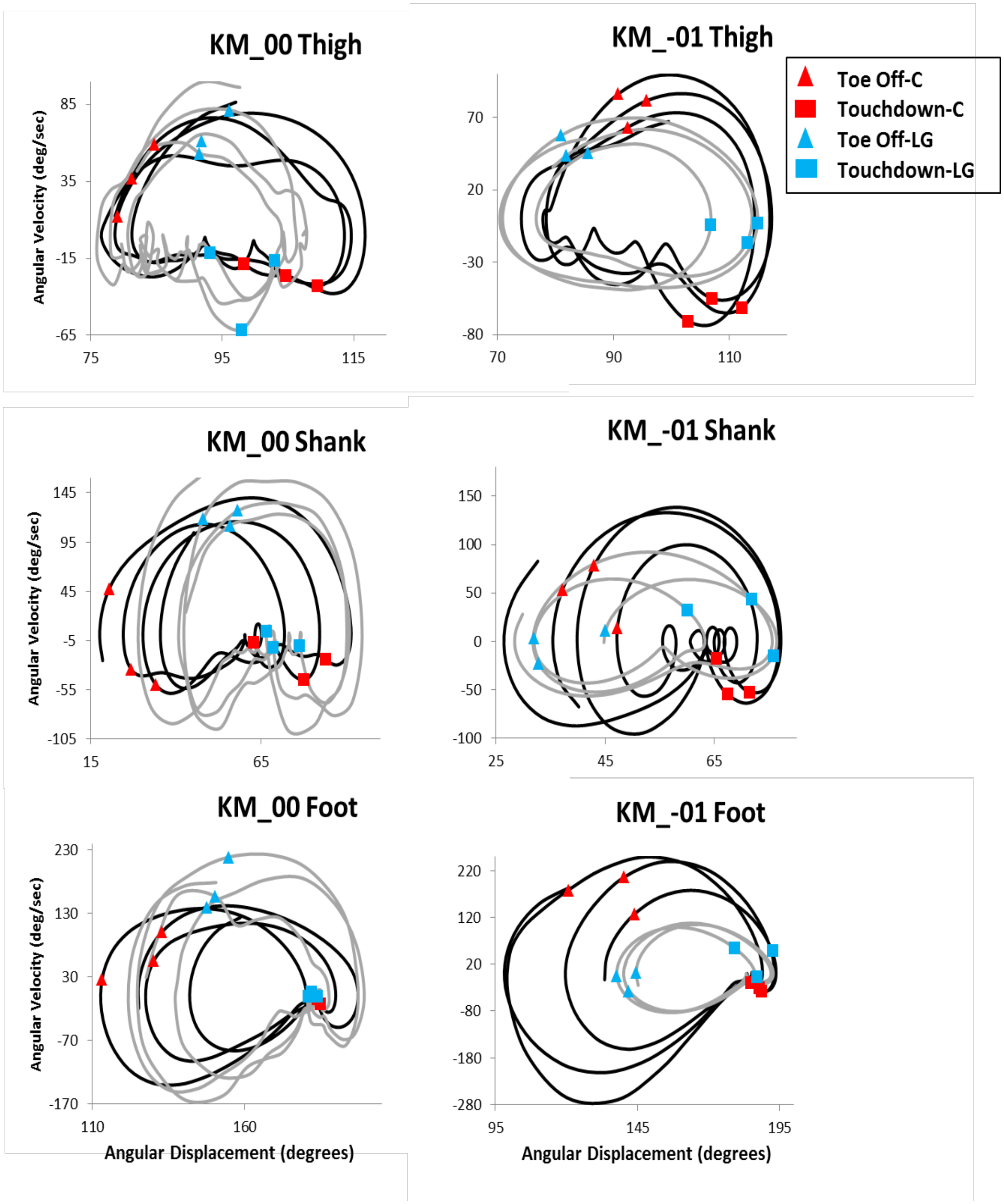
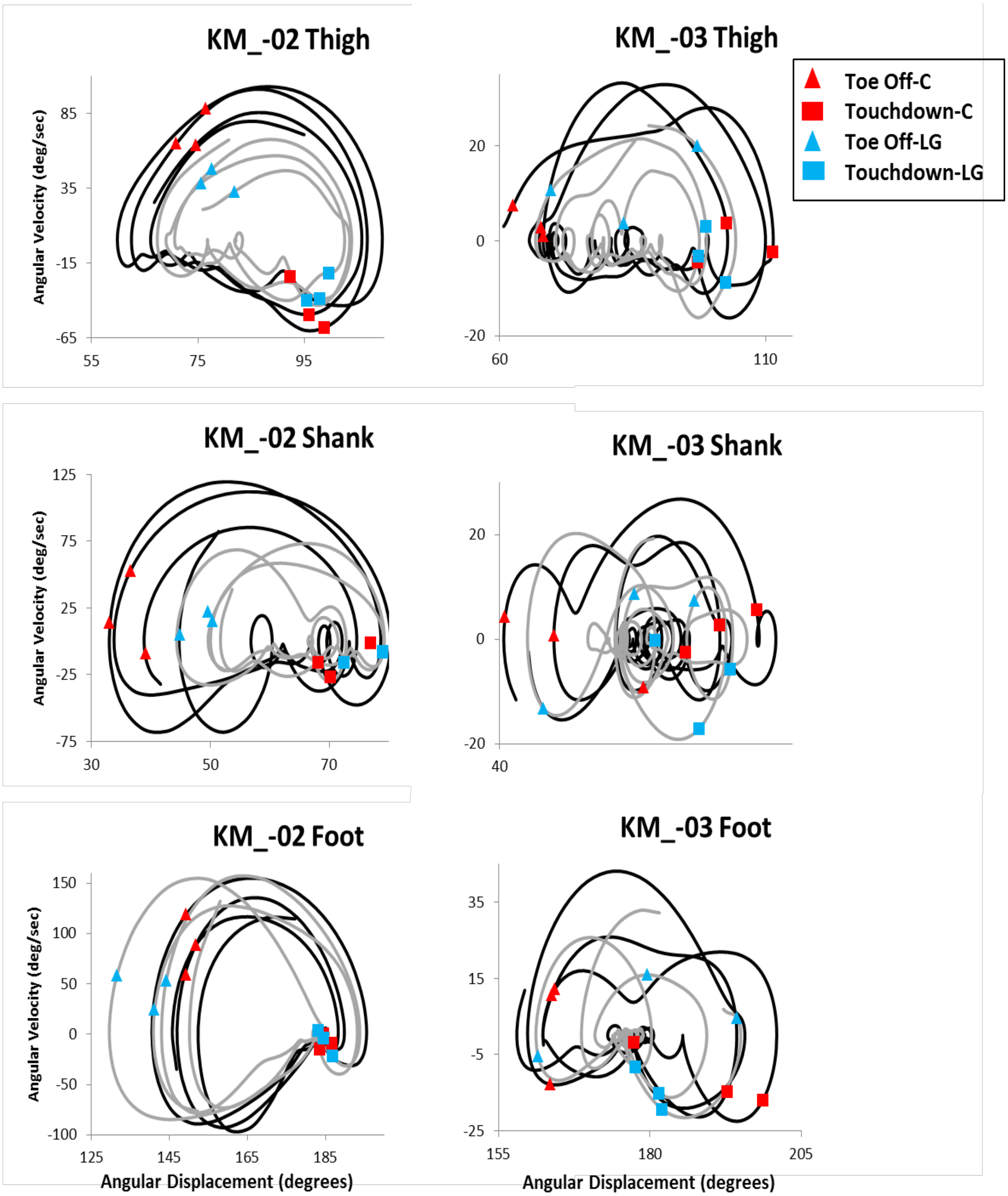


Figure F.1. (continued)



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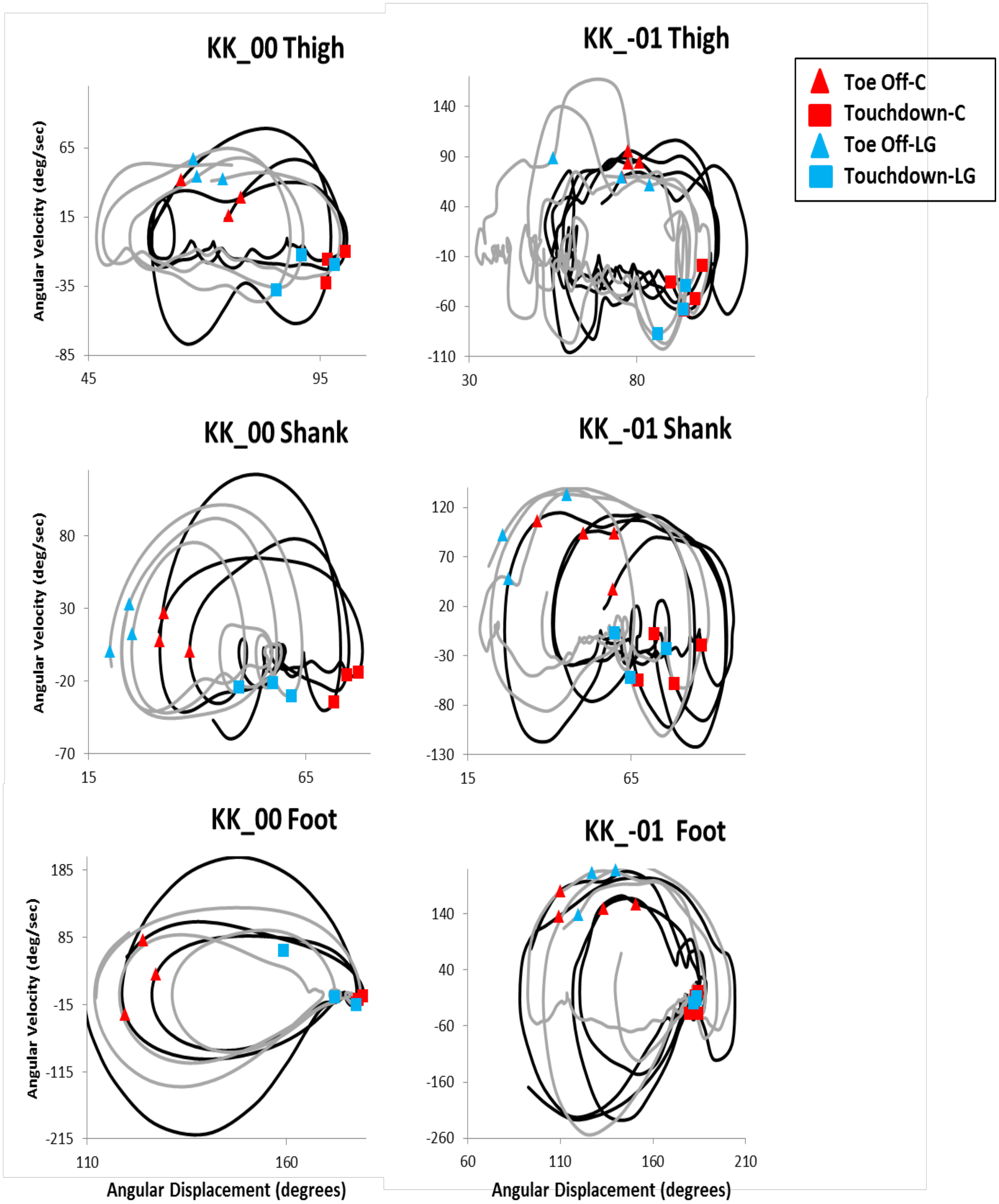


Figure F.1. (continued)

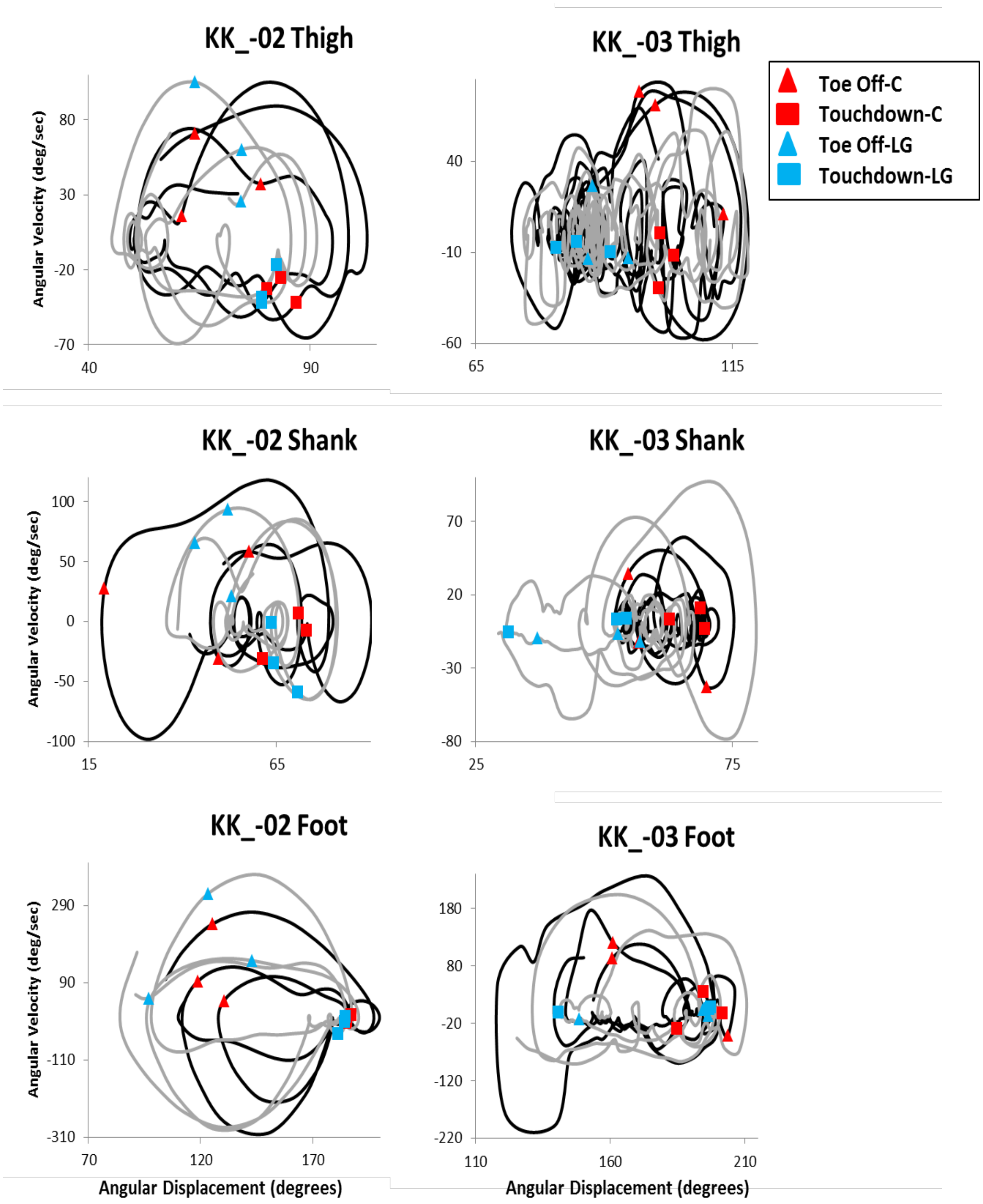
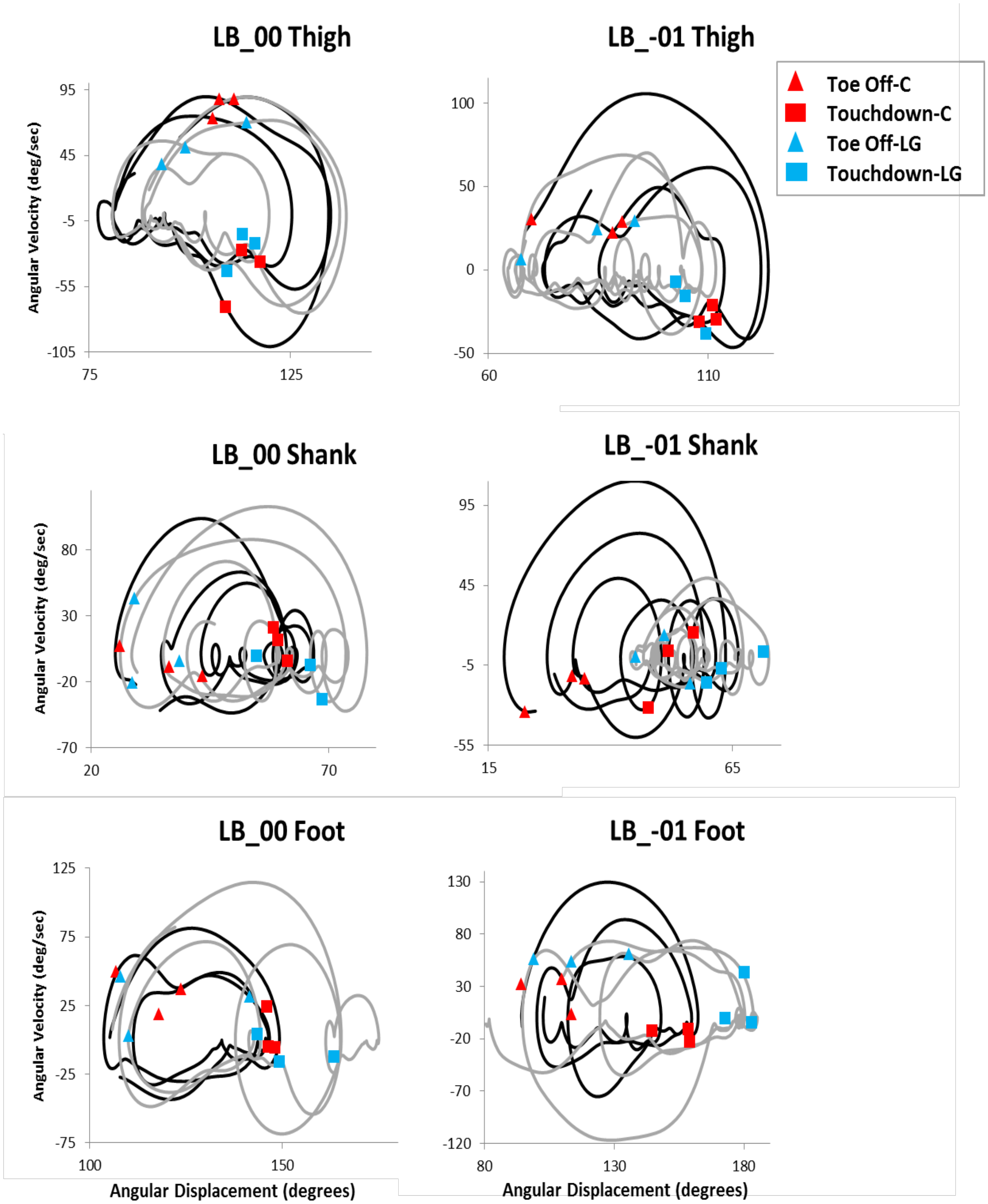


Figure F.1. (continued)


LB_00 Thigh

LB_-01 Thigh

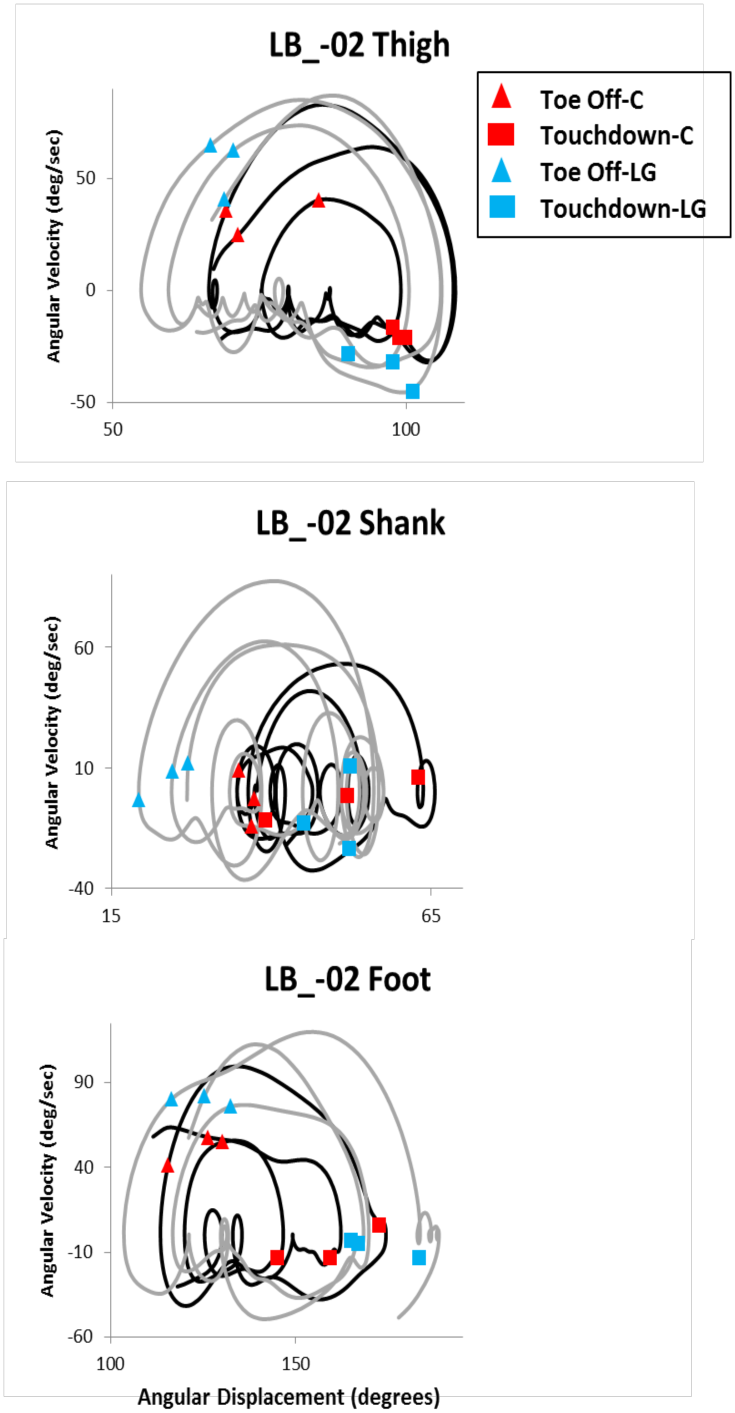
LB_00 Shank

LB_-01 Shank

LB_00 Foot

LB_-01 Foot

Figure F.1. (continued)



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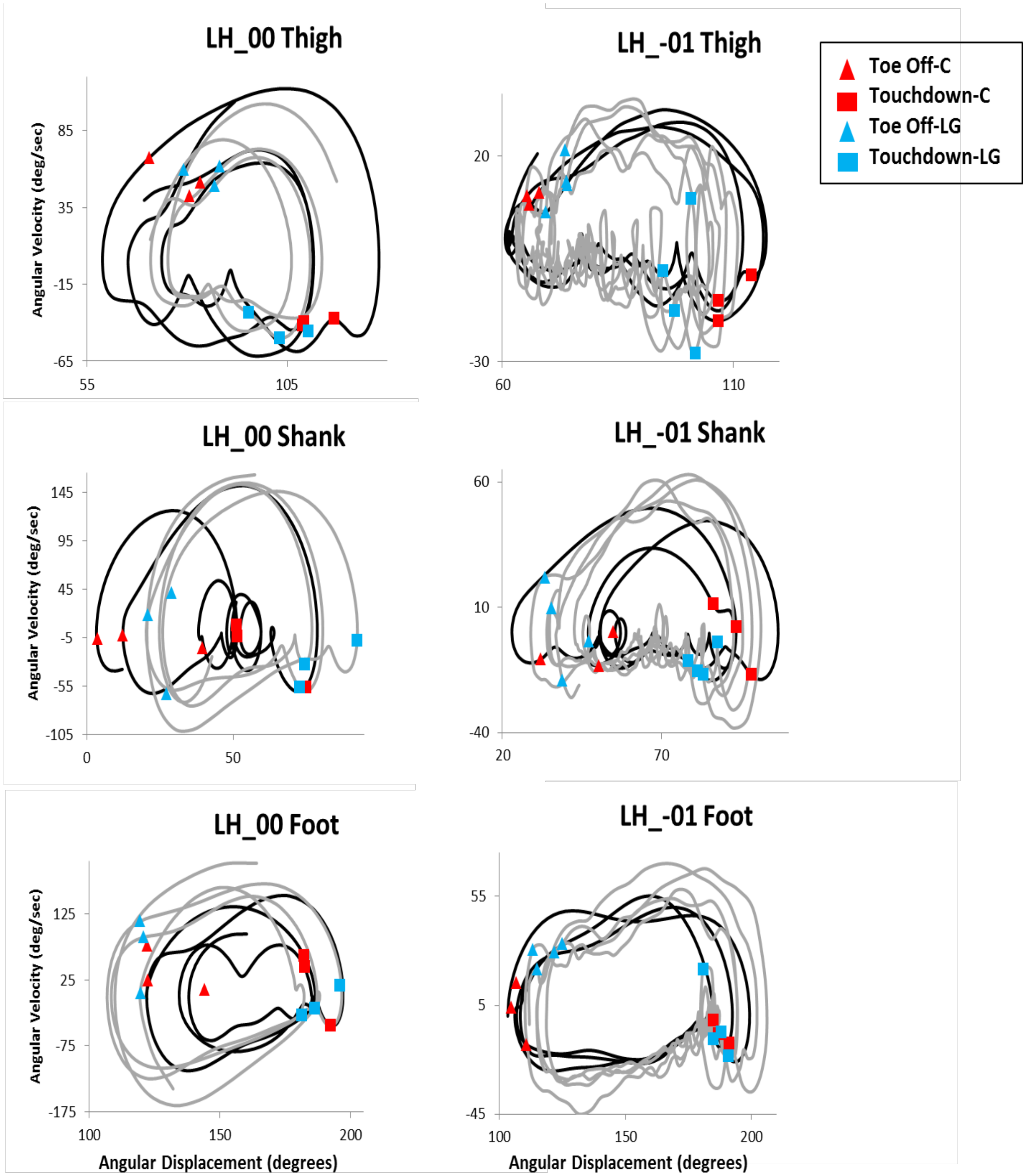
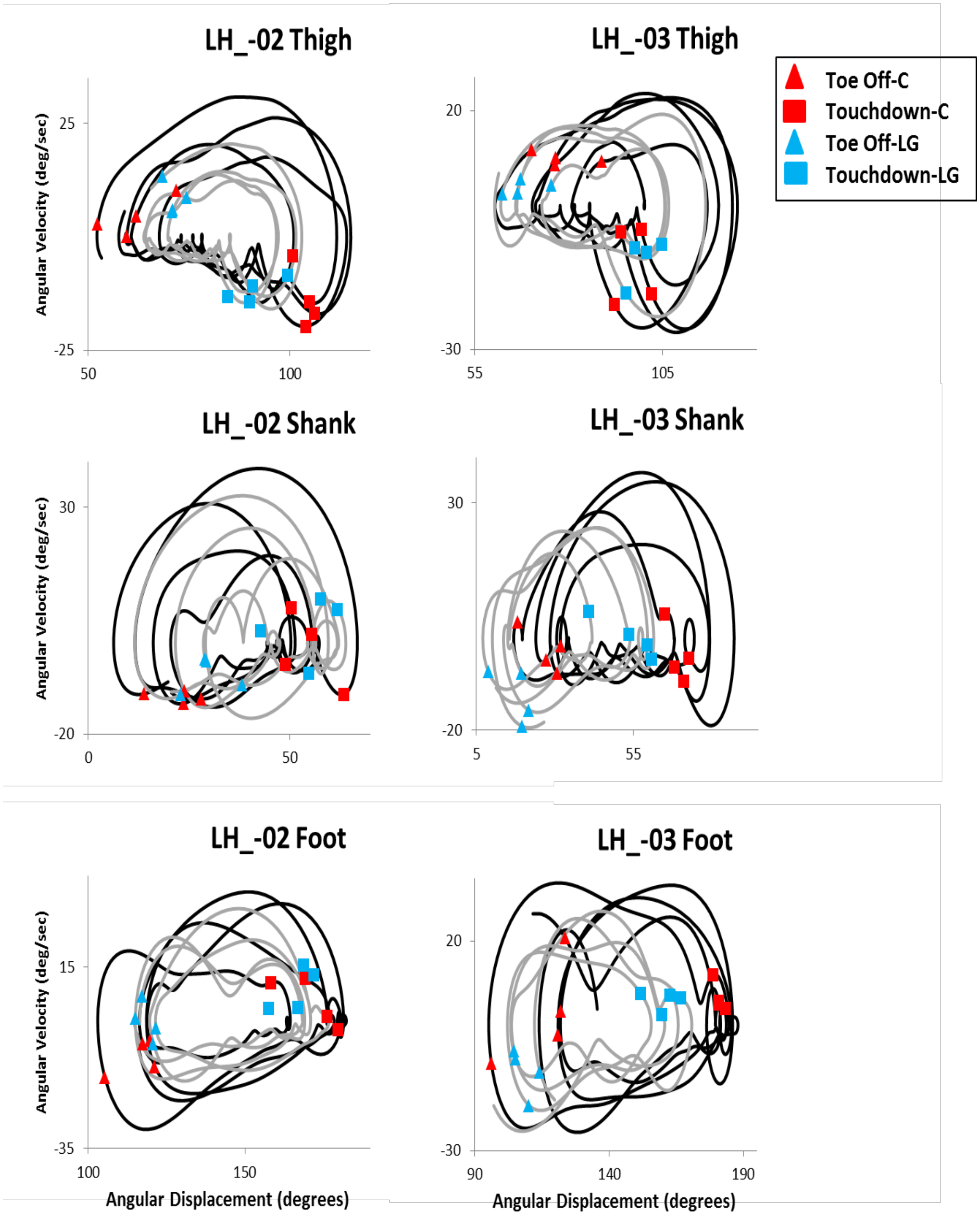


Figure F.1. (continued)



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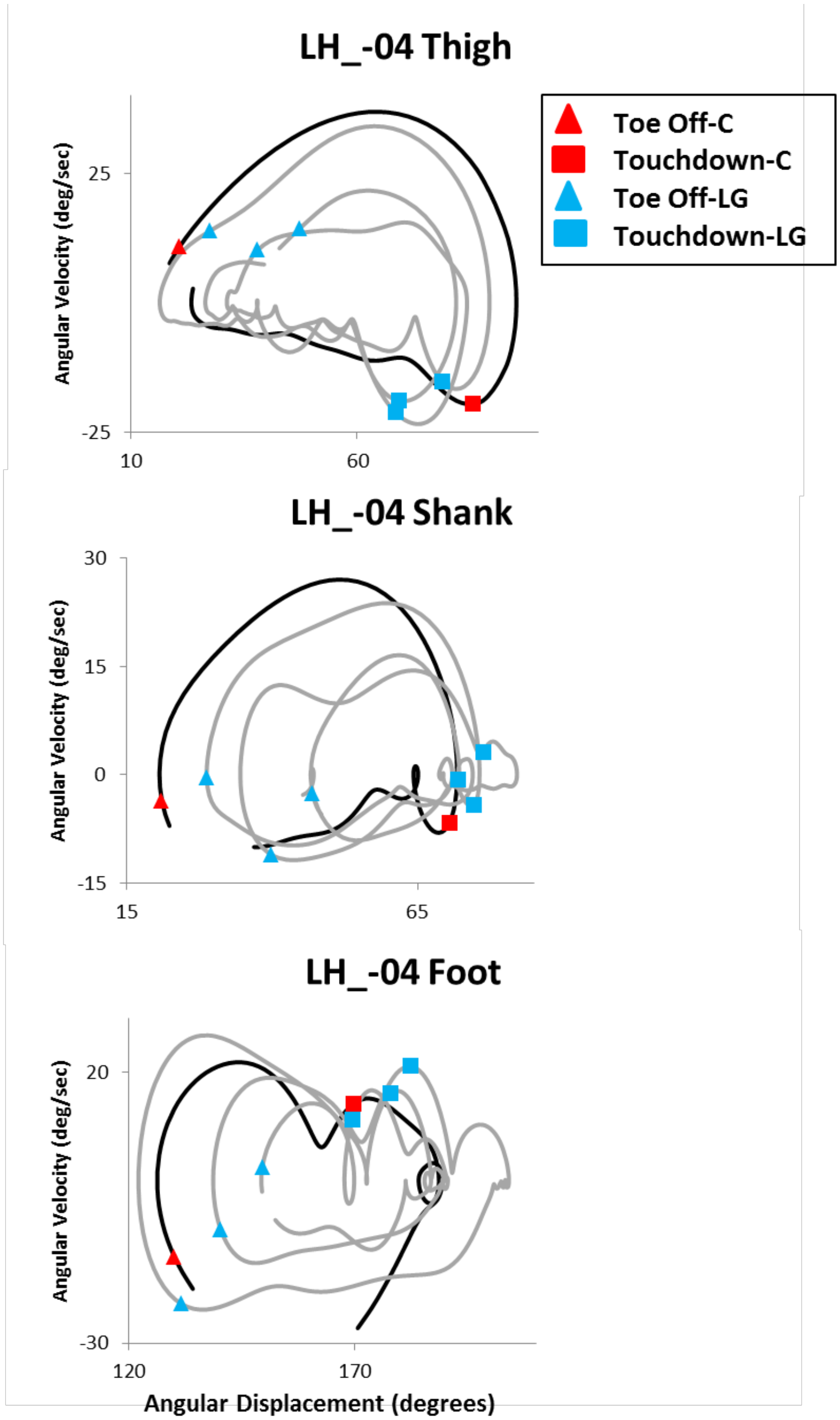

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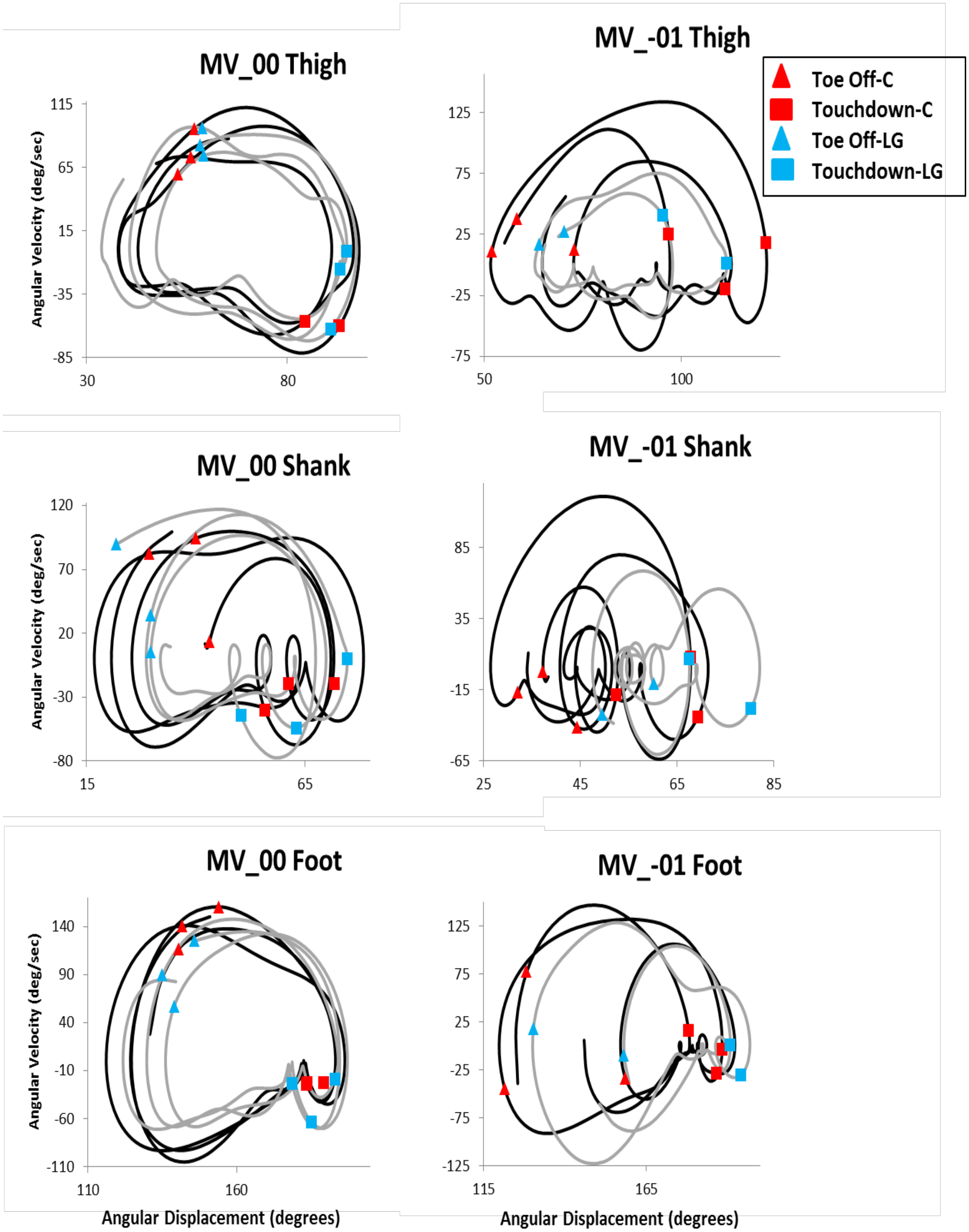
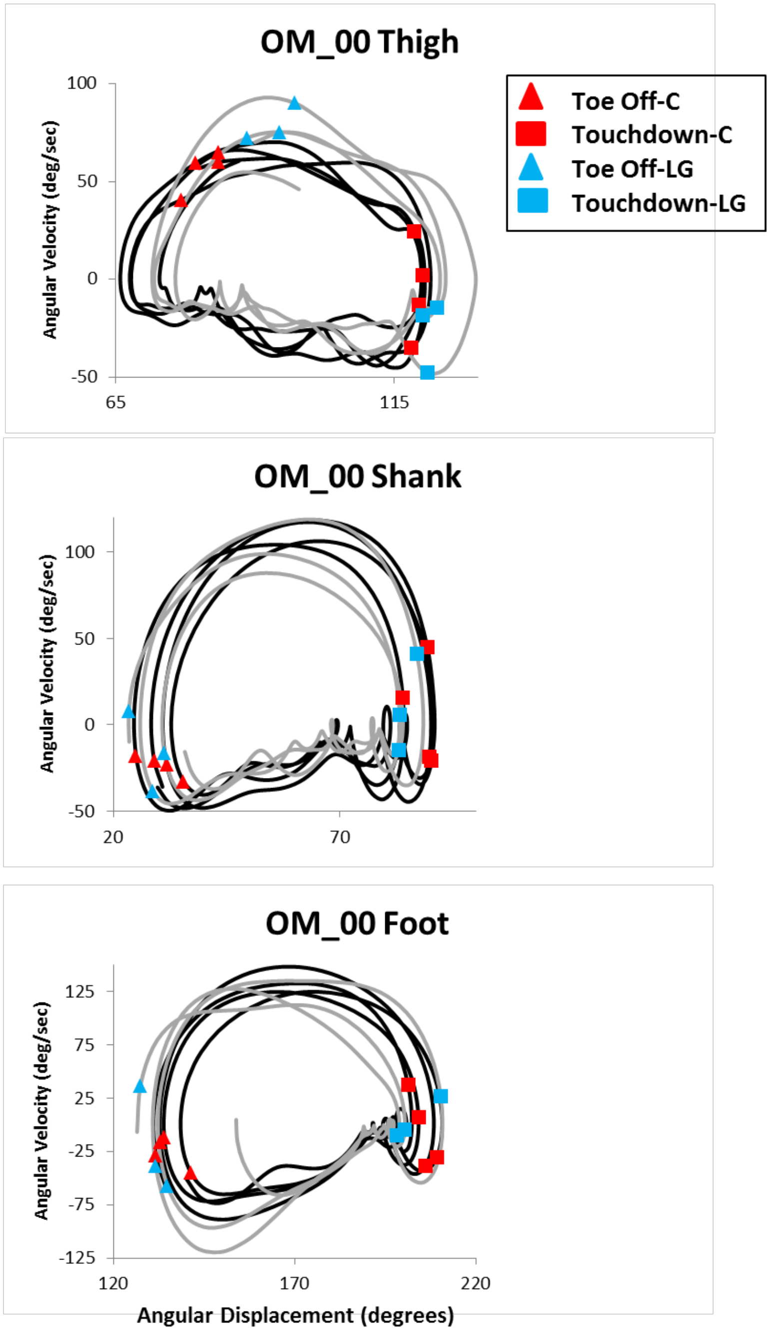


Figure F.1. (continued)



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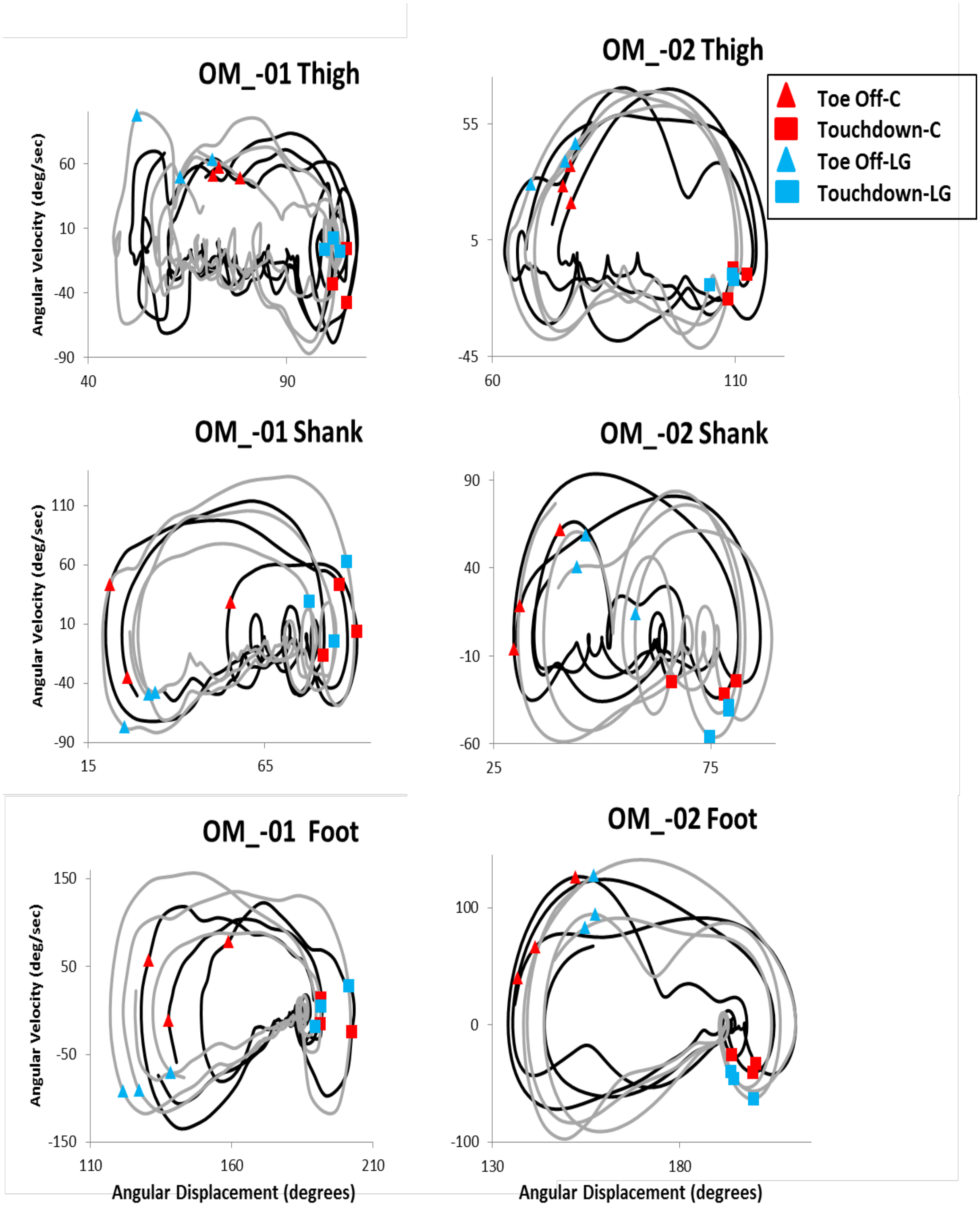

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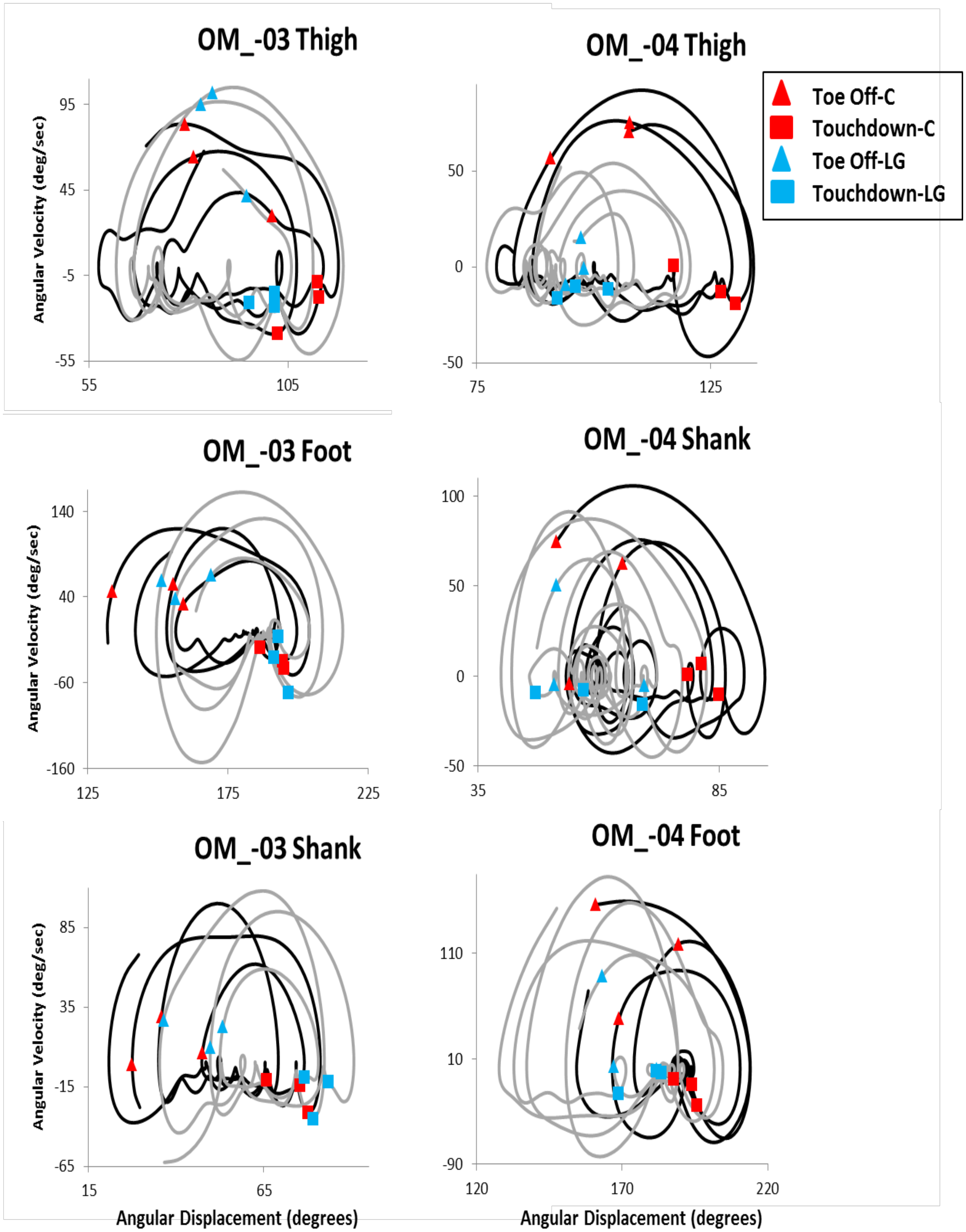


Figure F.1. (continued)

Appendix G.

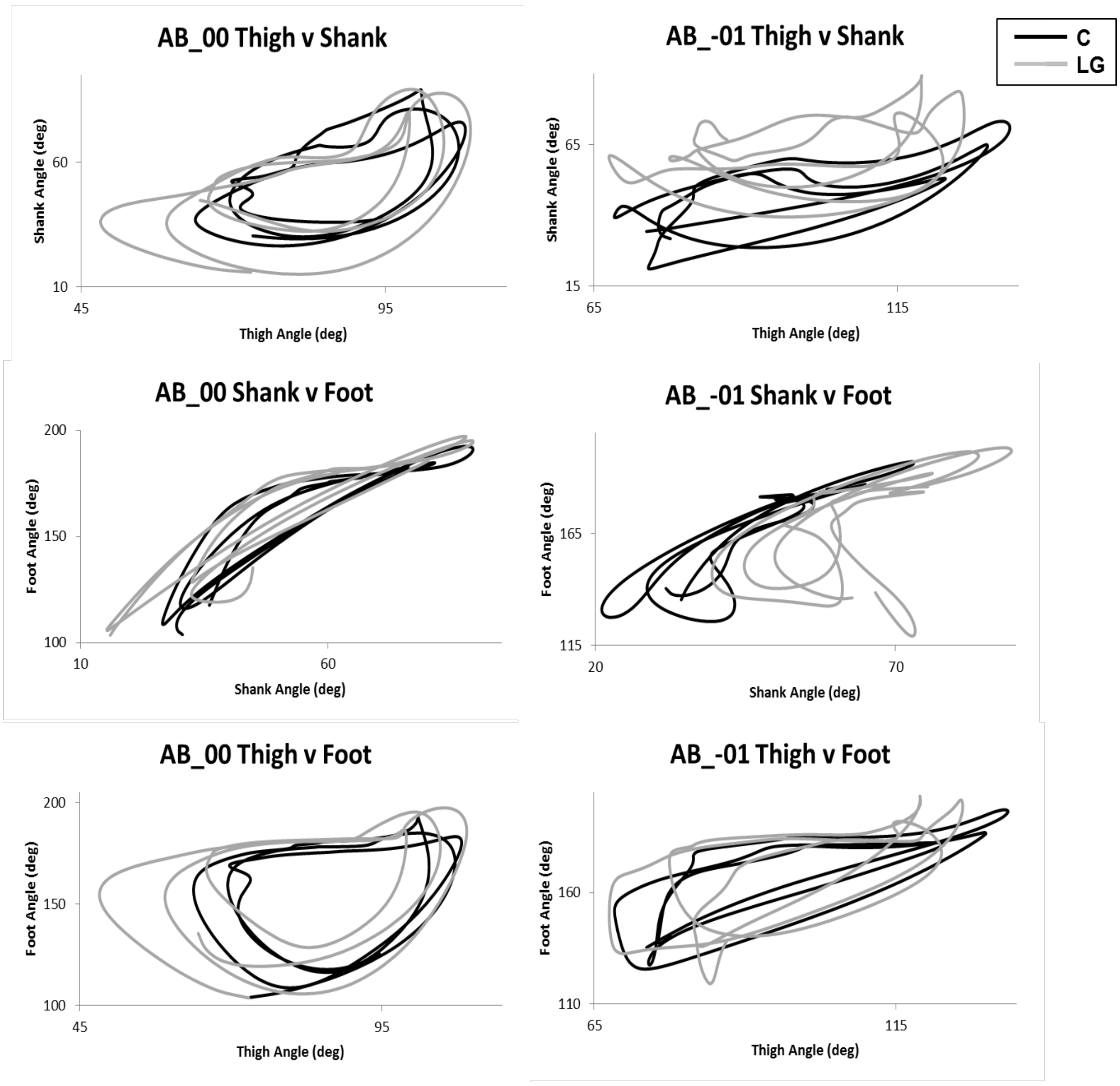
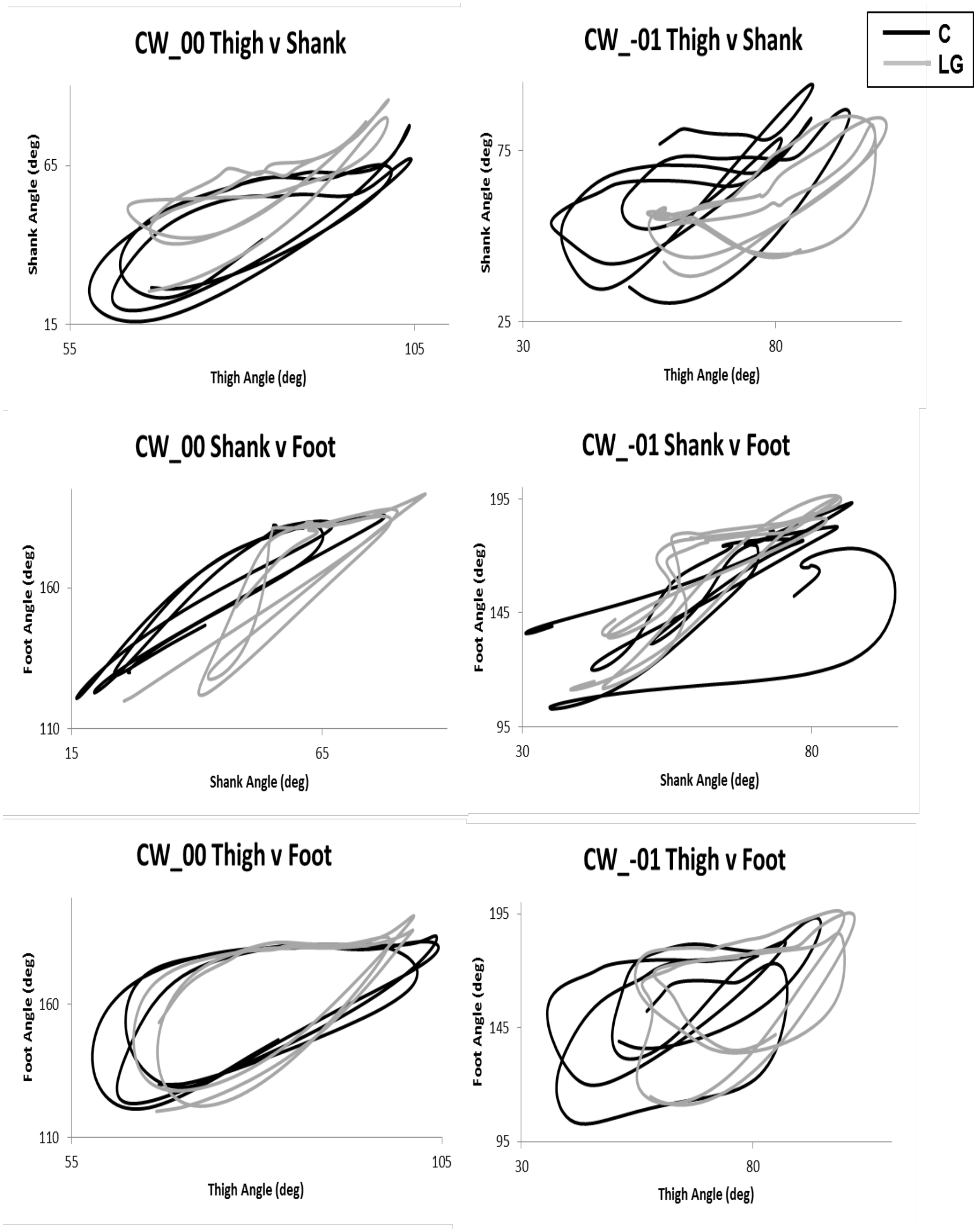


Figure G.1. Segmental angle-angle plots for 3 consecutive cruising strides per condition for infants across visits. Trajectories unfold in clockwise direction over the cycle duration.
— Control (diaper-only); — Lycra Garment.


CW_00 Thigh v Shank

CW_-01 Thigh v Shank

CW_00 Shank v Foot

CW_-01 Shank v Foot

CW_00 Thigh v Foot

CW_-01 Thigh v Foot

Figure G.1. (continued)

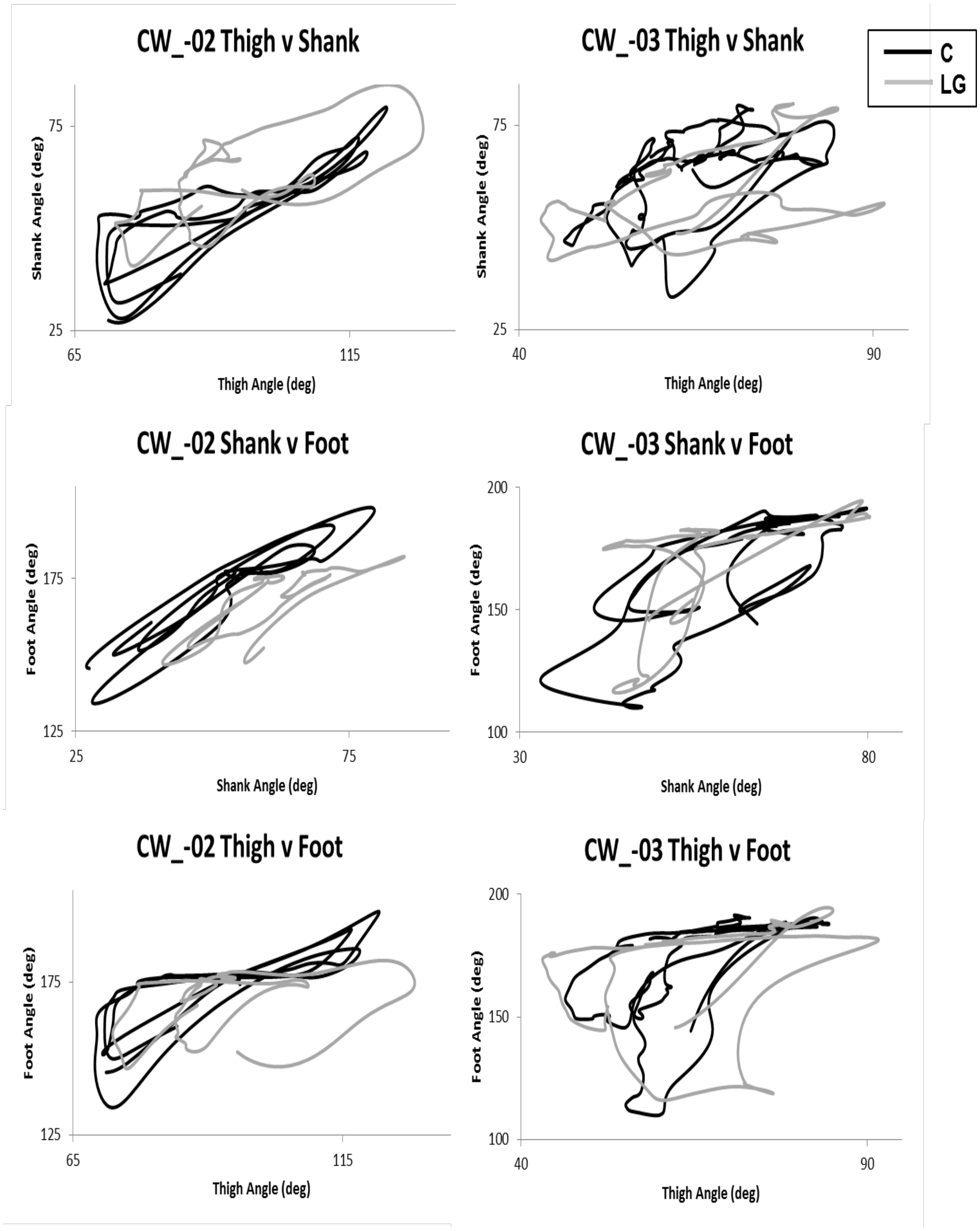


Figure G.1. (continued)

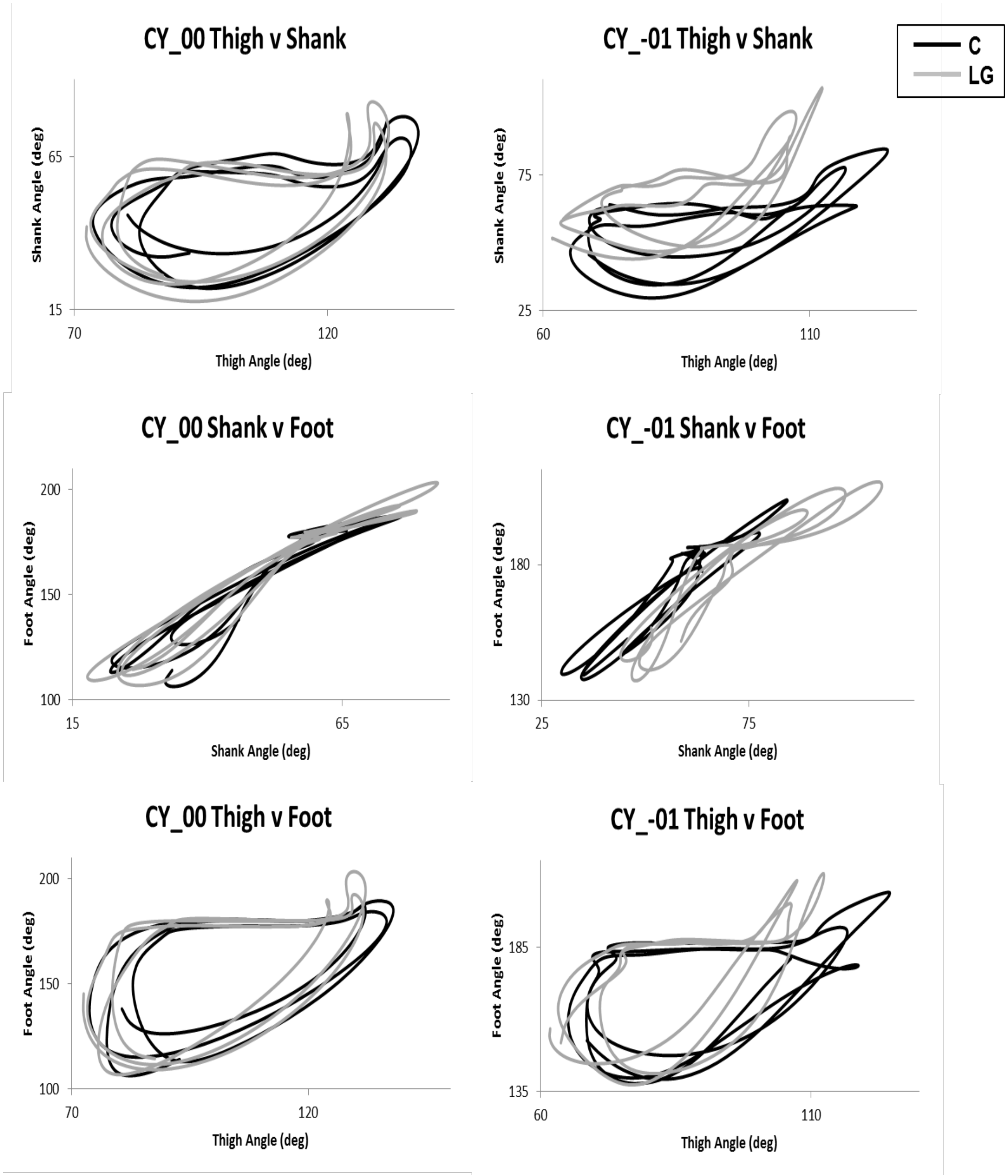


Figure G.1. (continued)

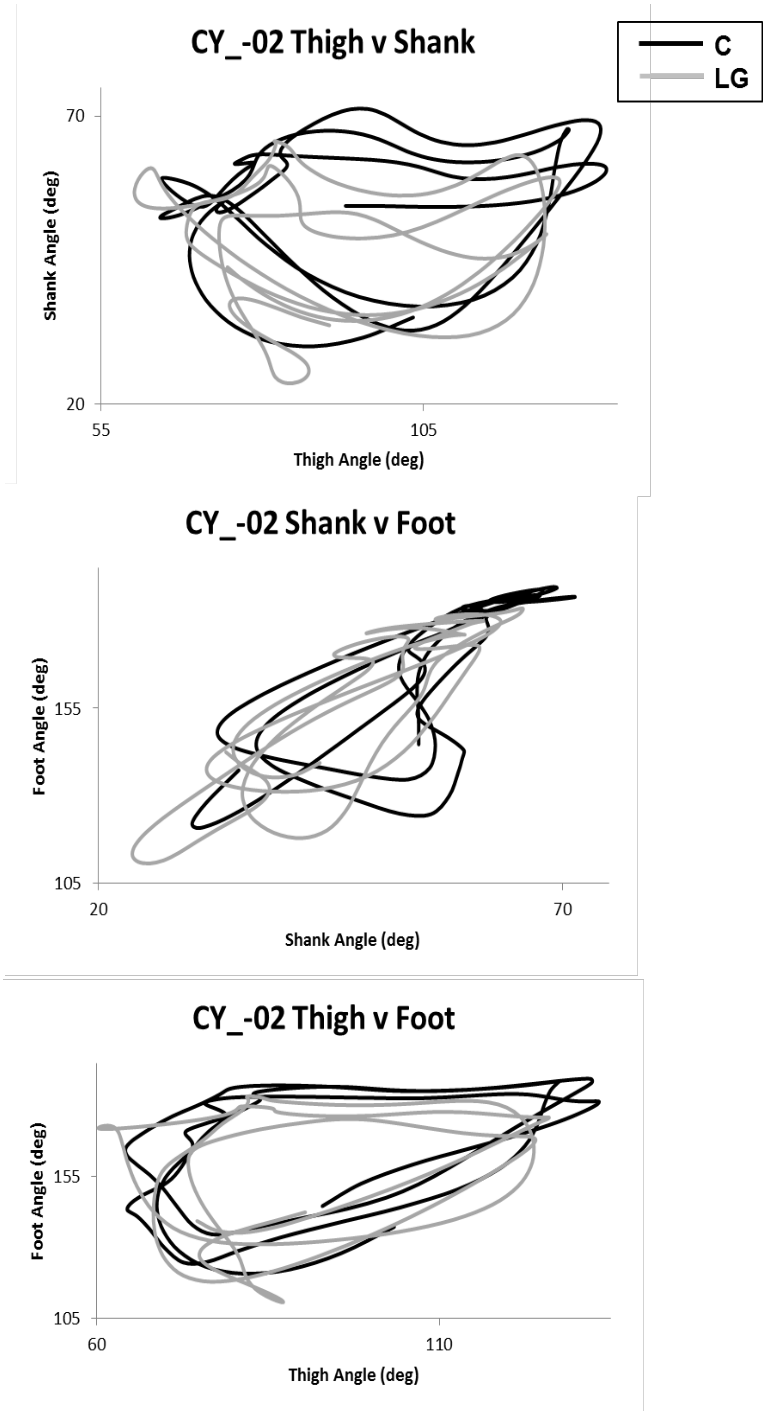


Figure G.1. (continued)

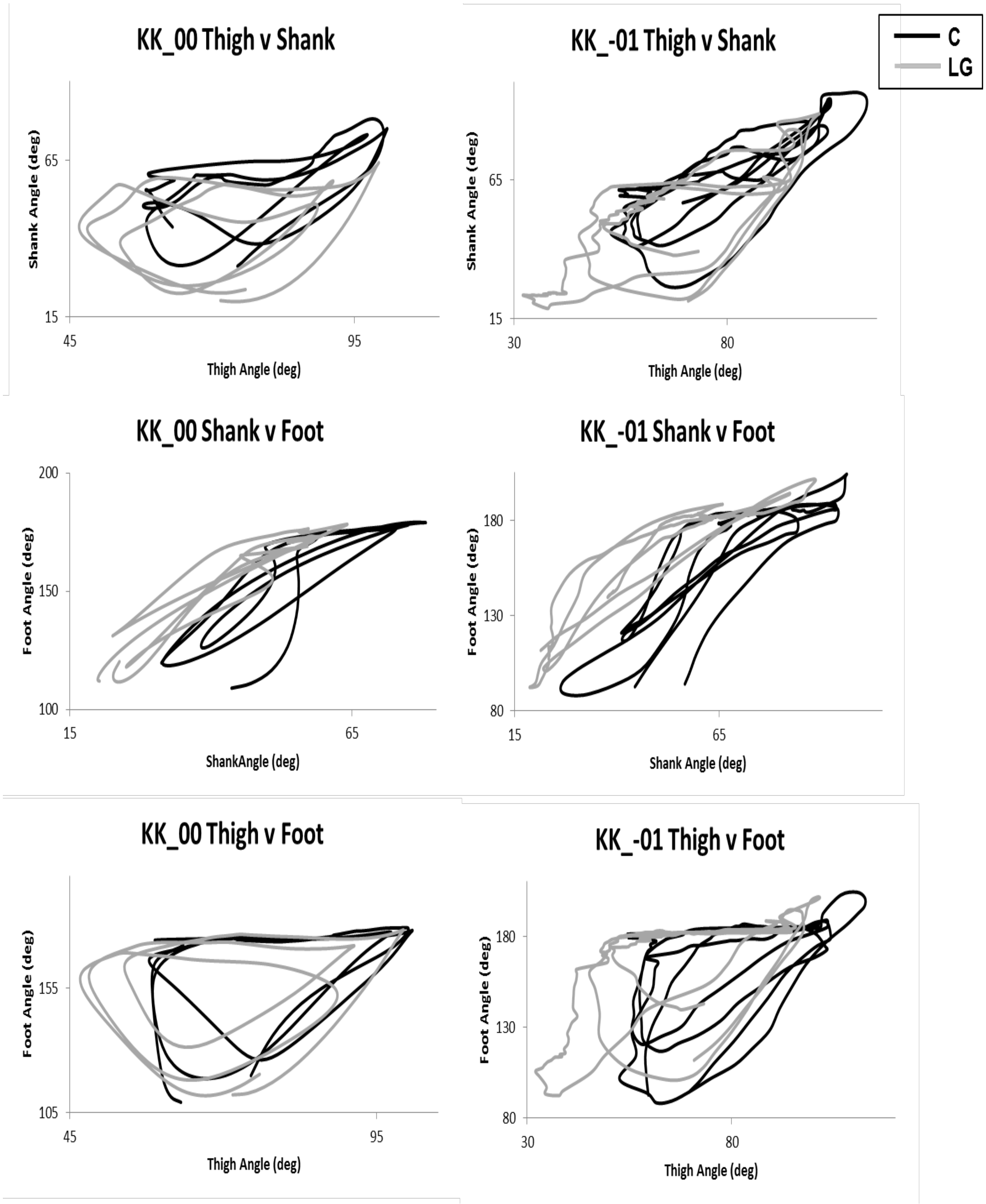
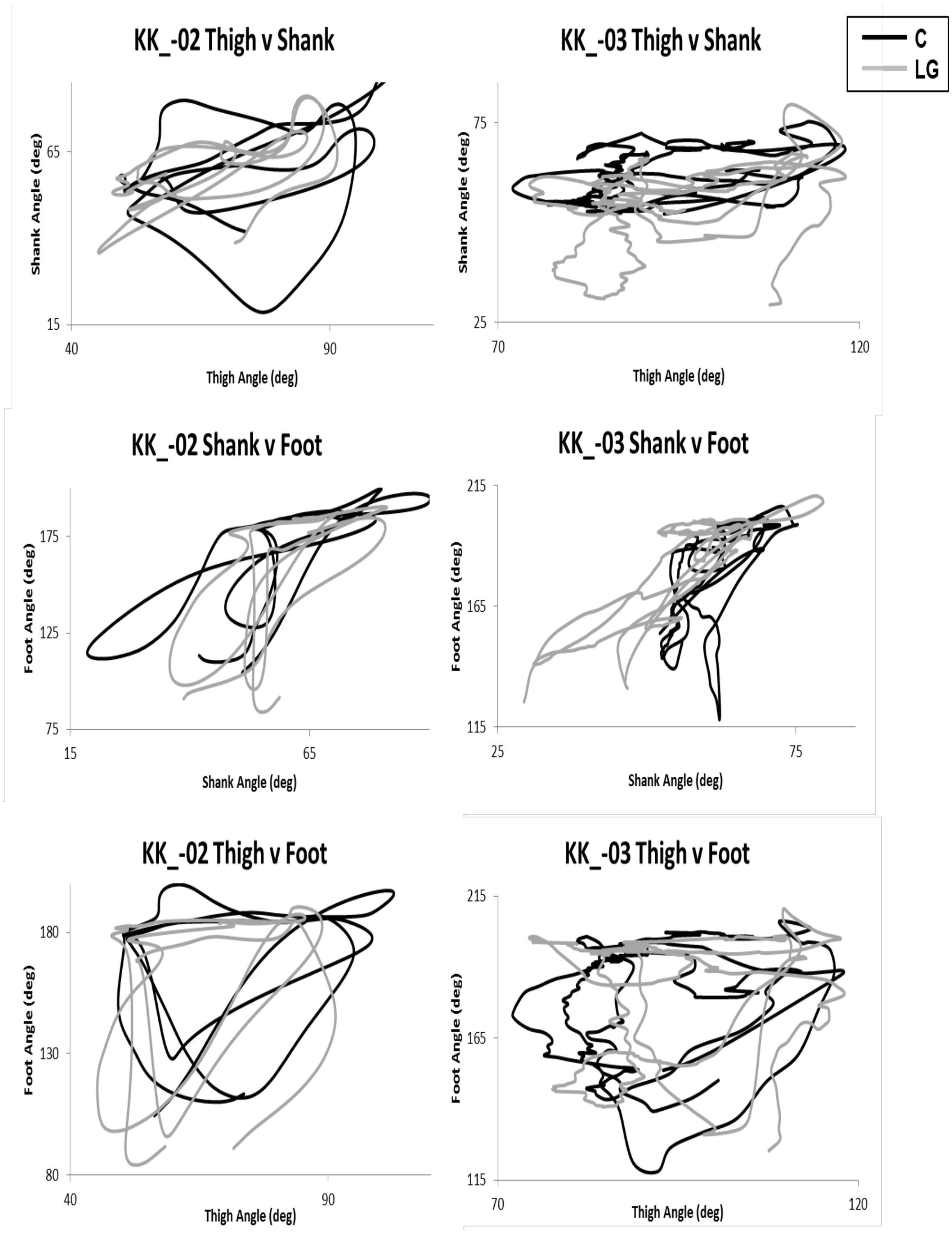


Figure G.1. (continued)



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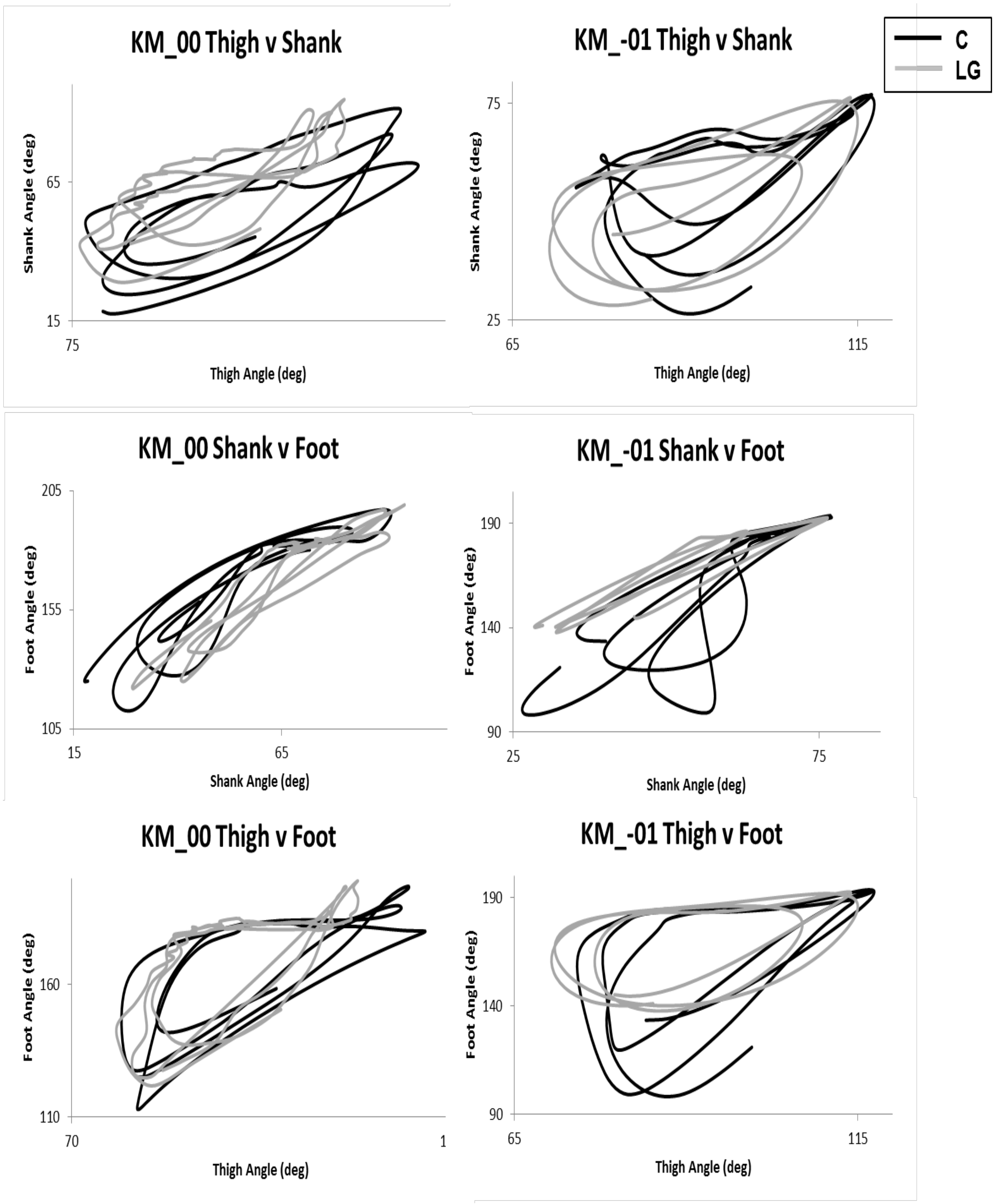


Figure G.1. (continued)

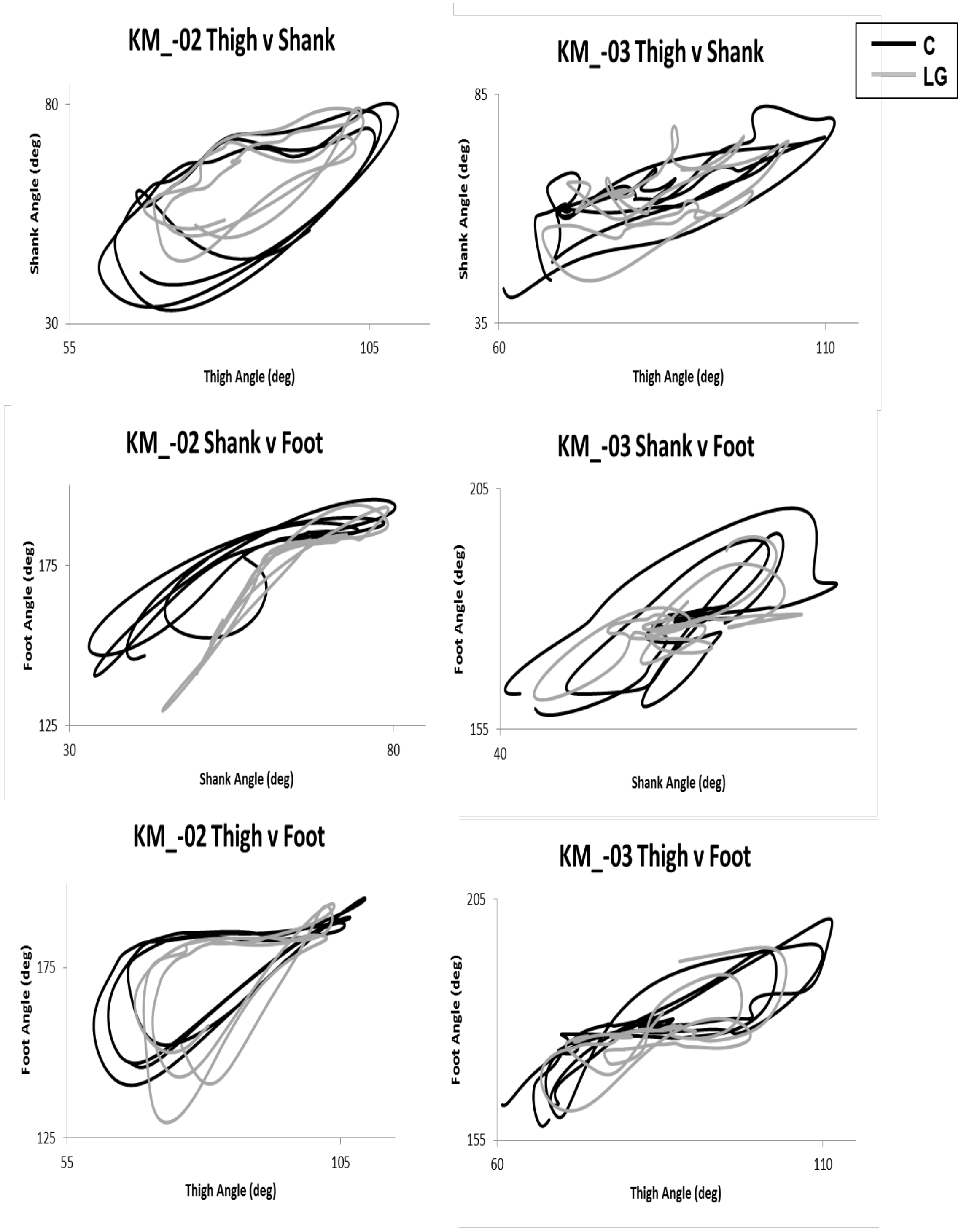


Figure G.1. (continued)

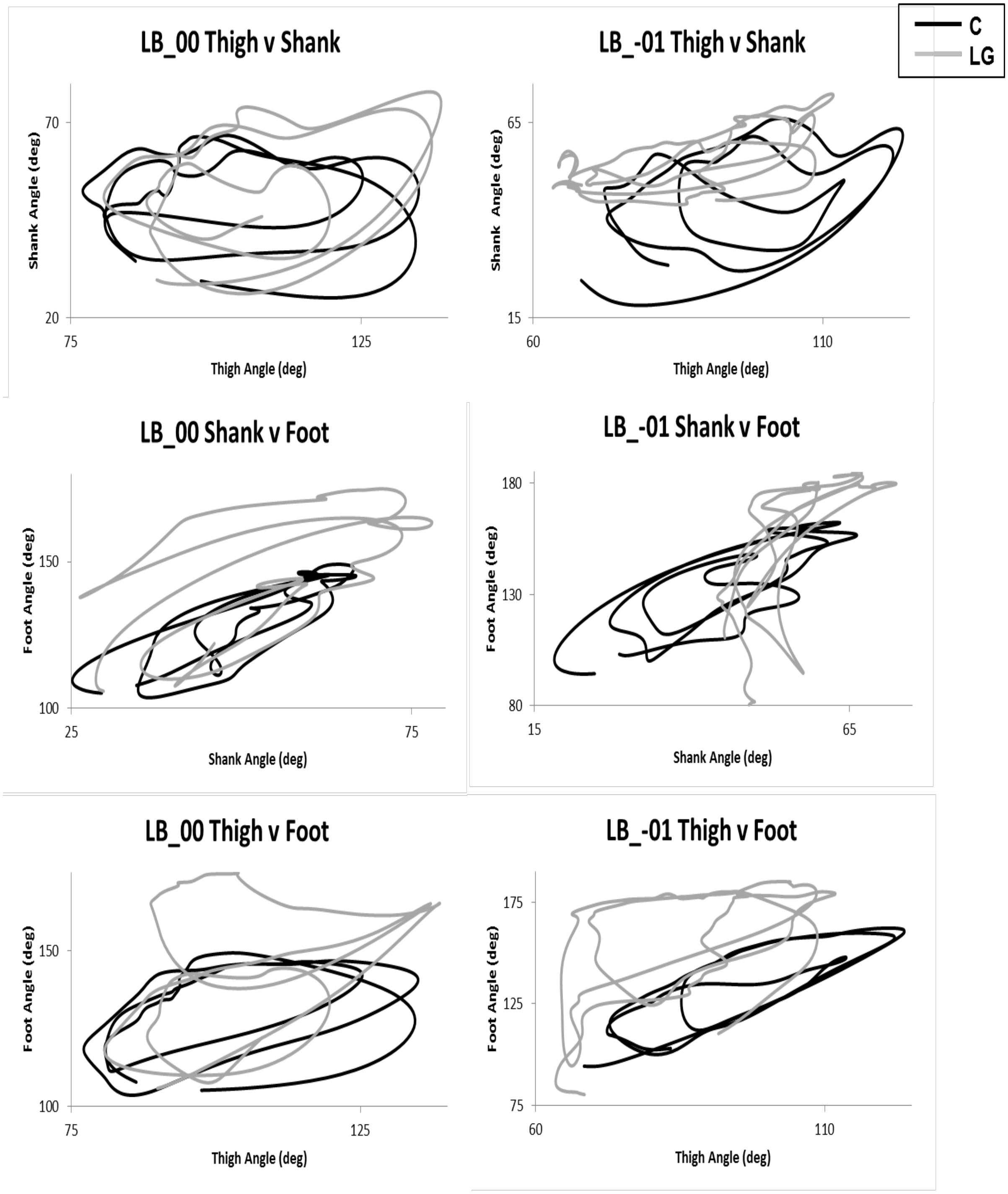


Figure G.1. (continued)

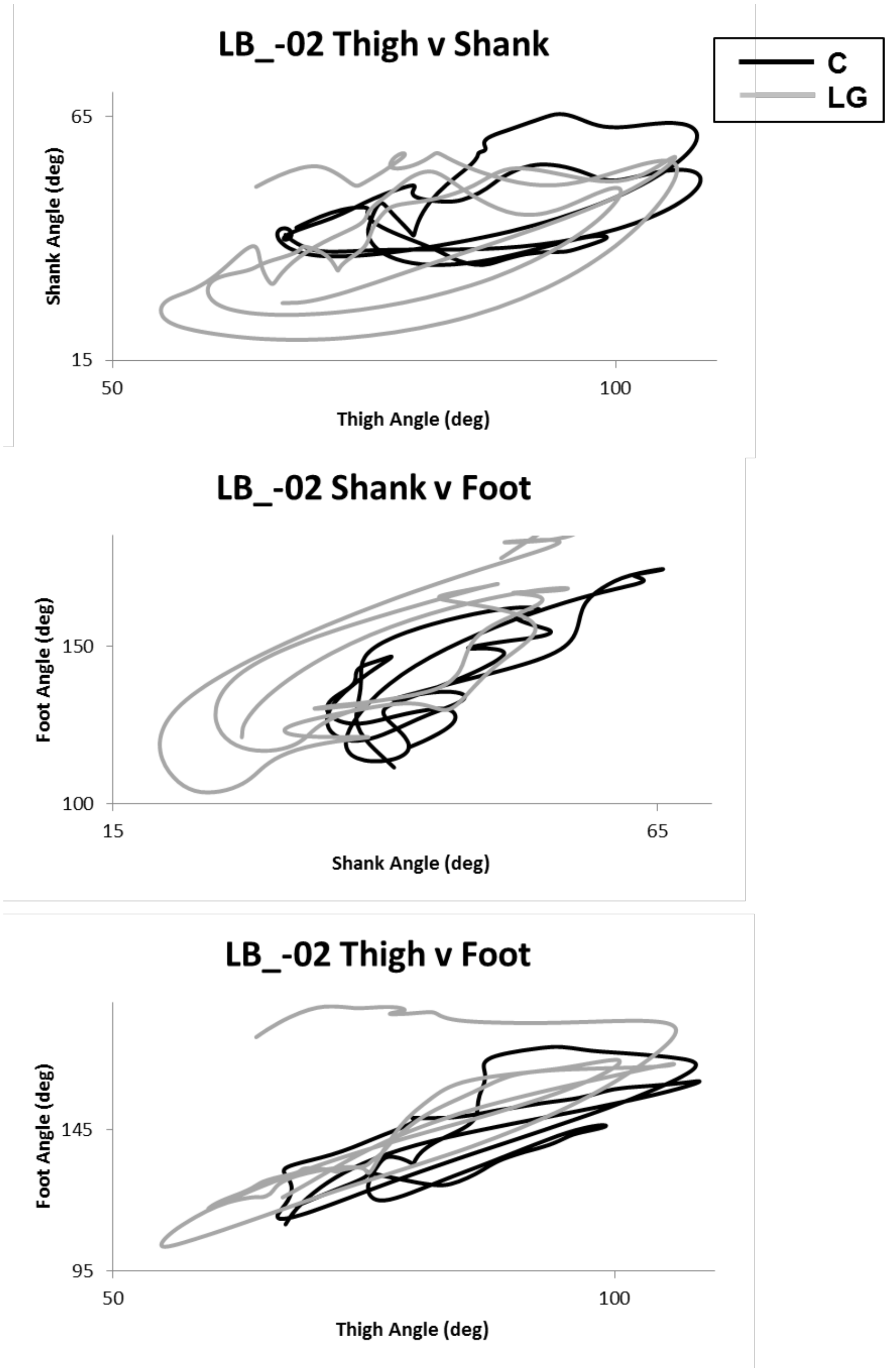


Figure G.1. (continued)

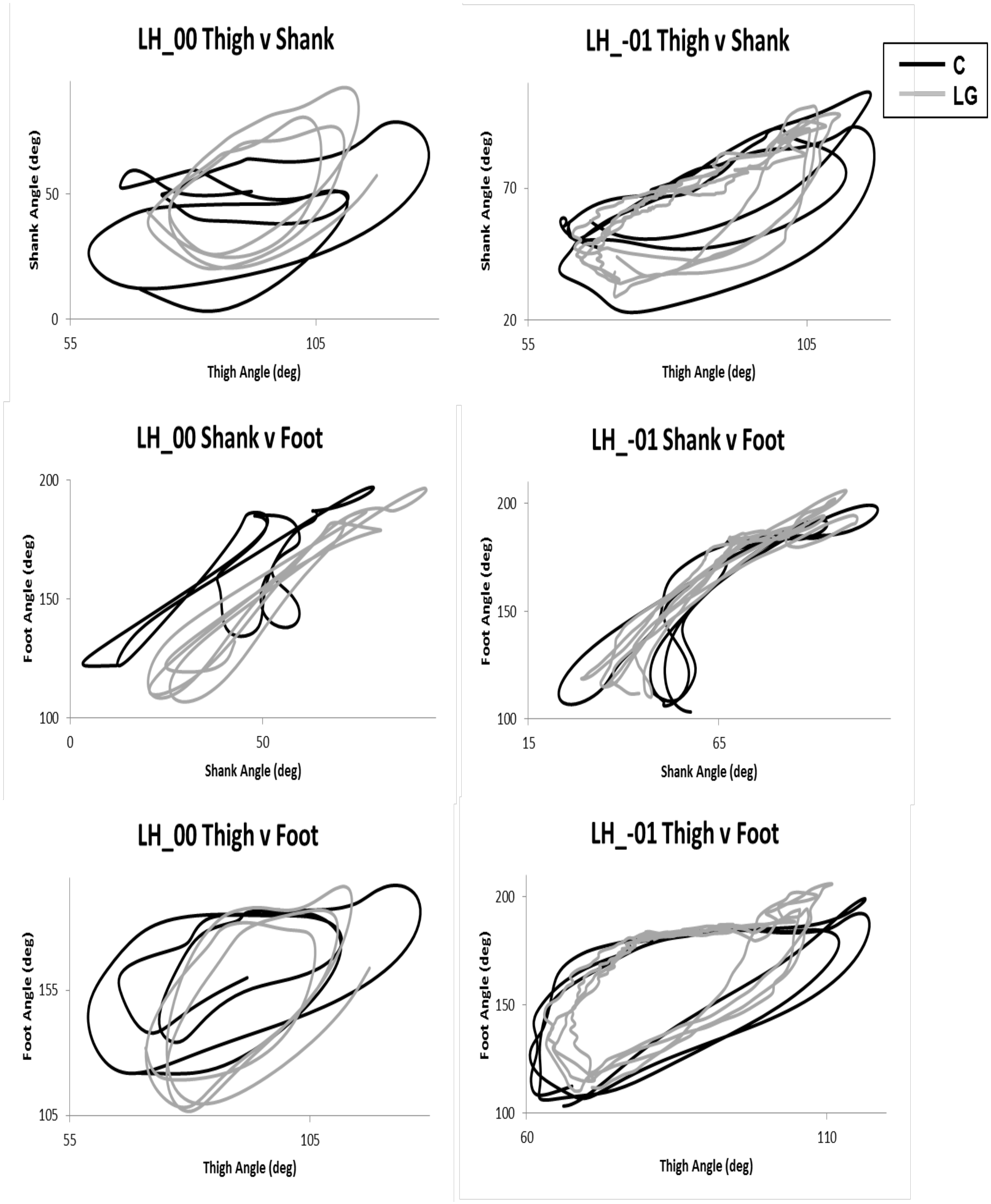


Figure G.1. (continued)

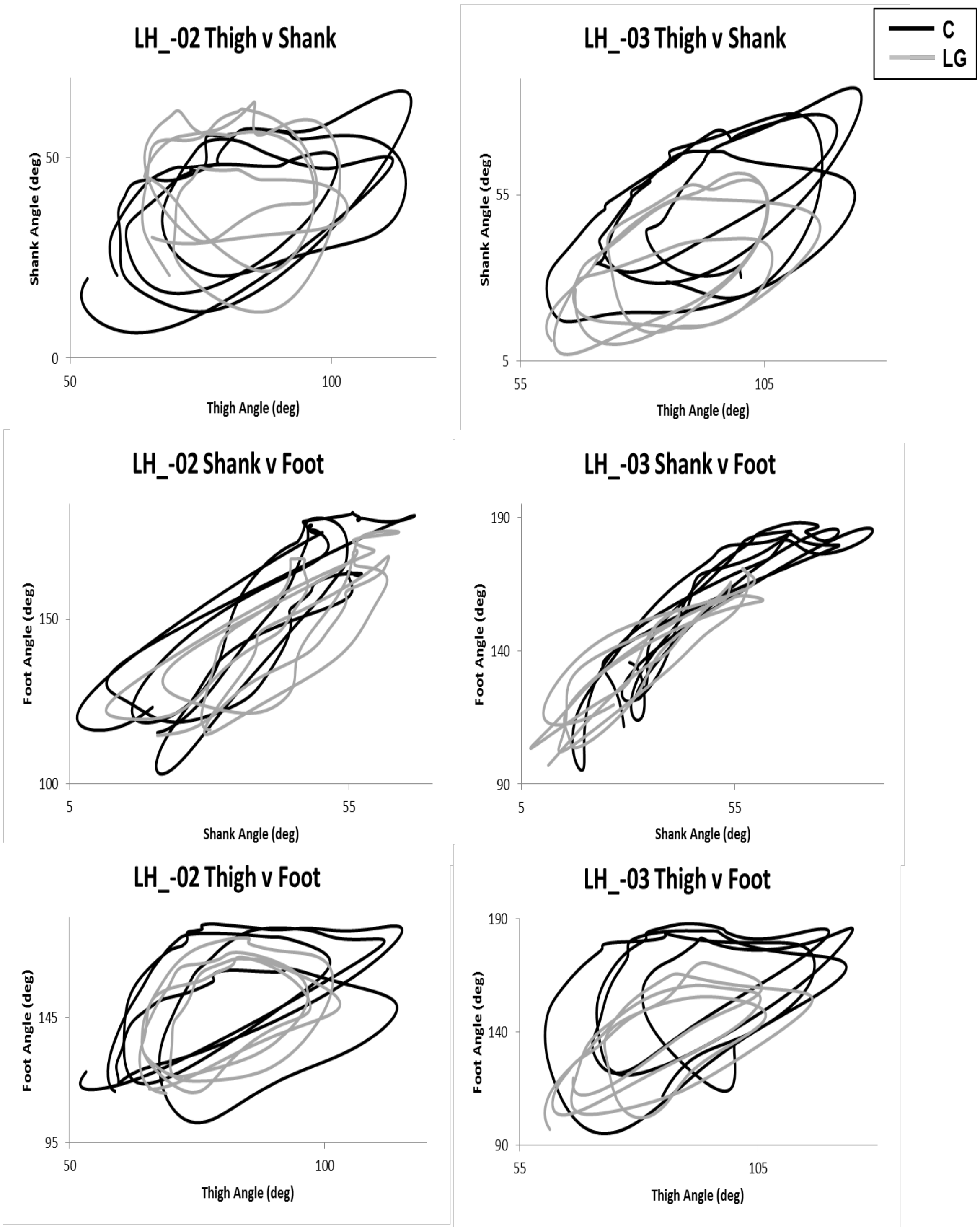
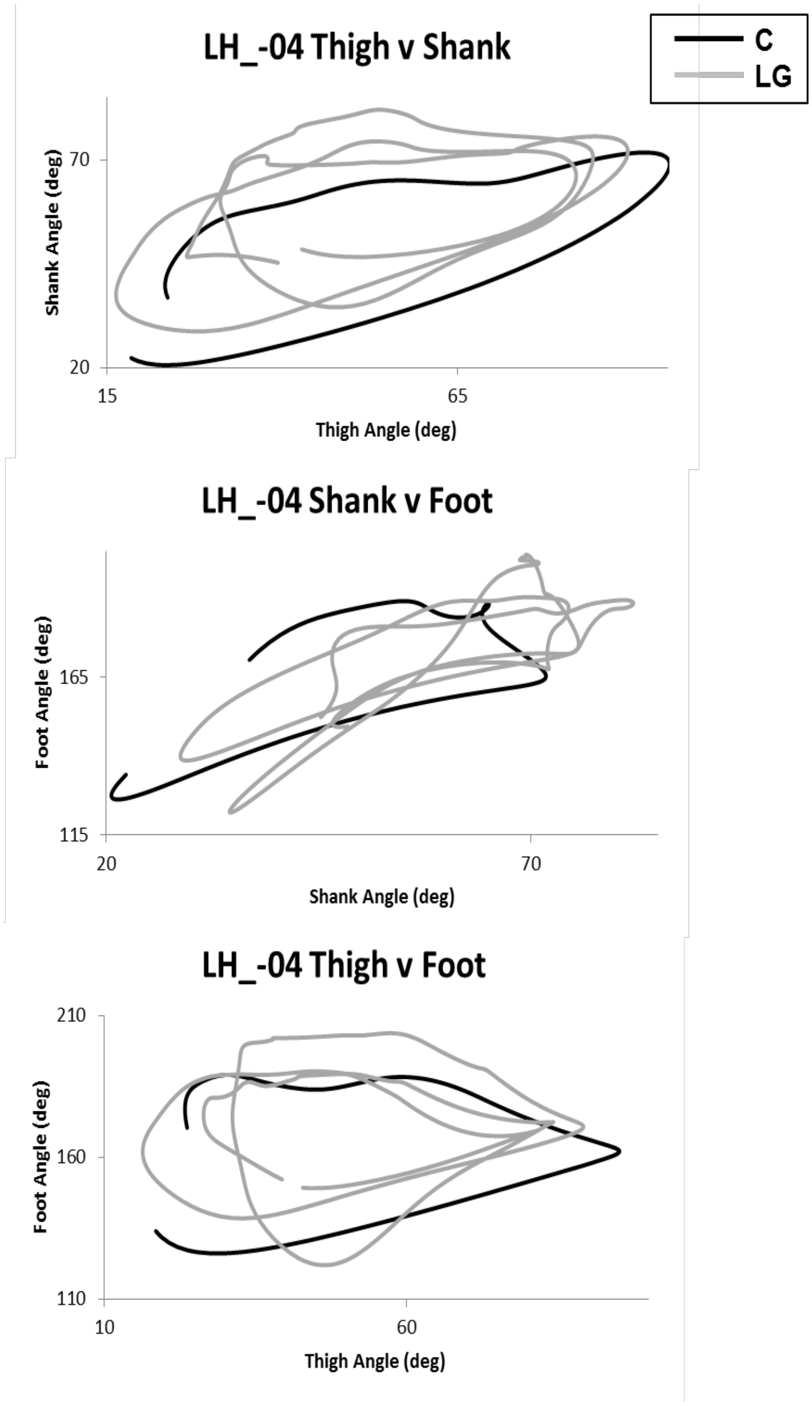


Figure G.1. (continued)


LH_-04 Thigh v Shank

LH_-04 Shank v Foot

LH_-04 Thigh v Foot

Figure G.1. (continued)

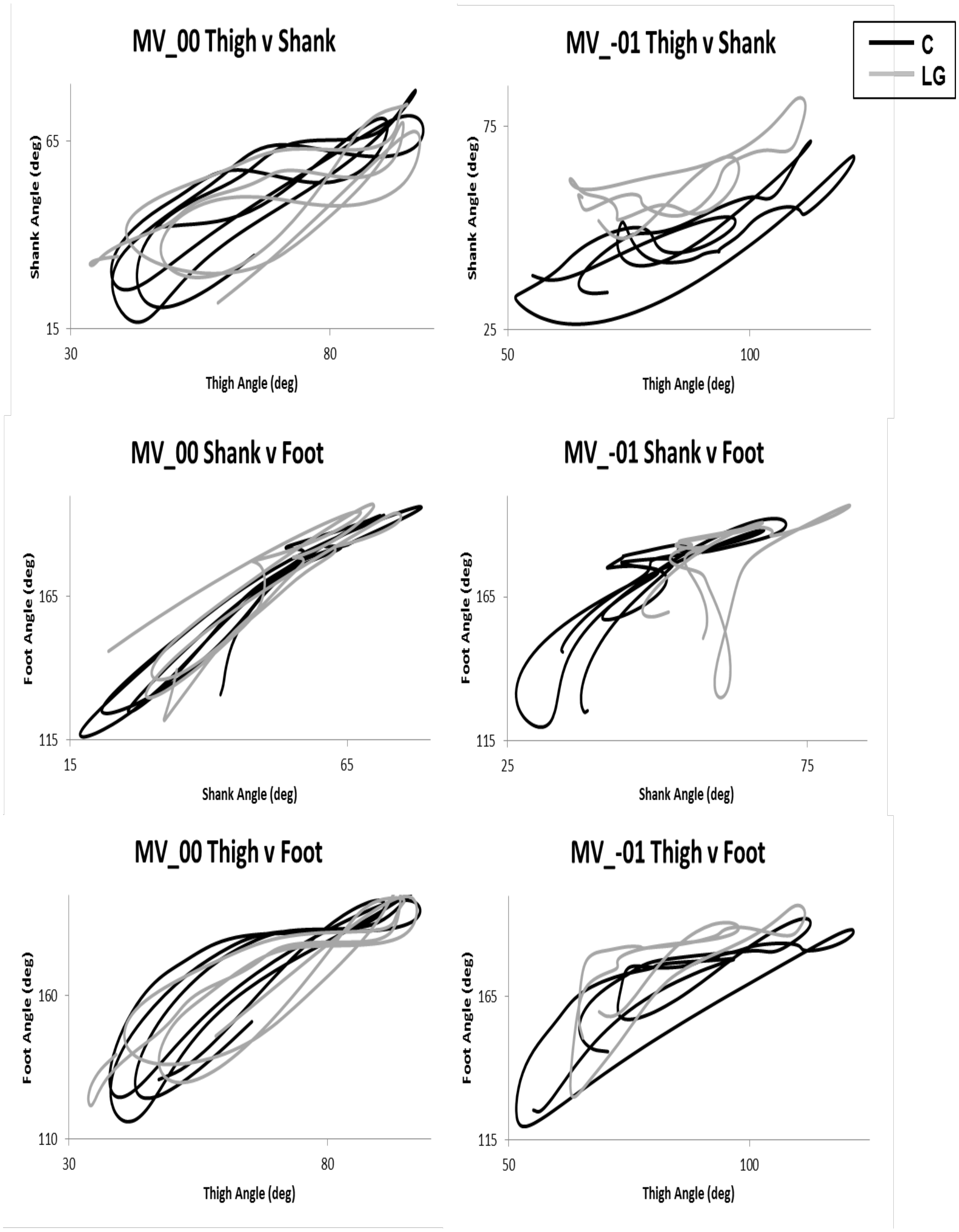


Figure G.1. (continued)

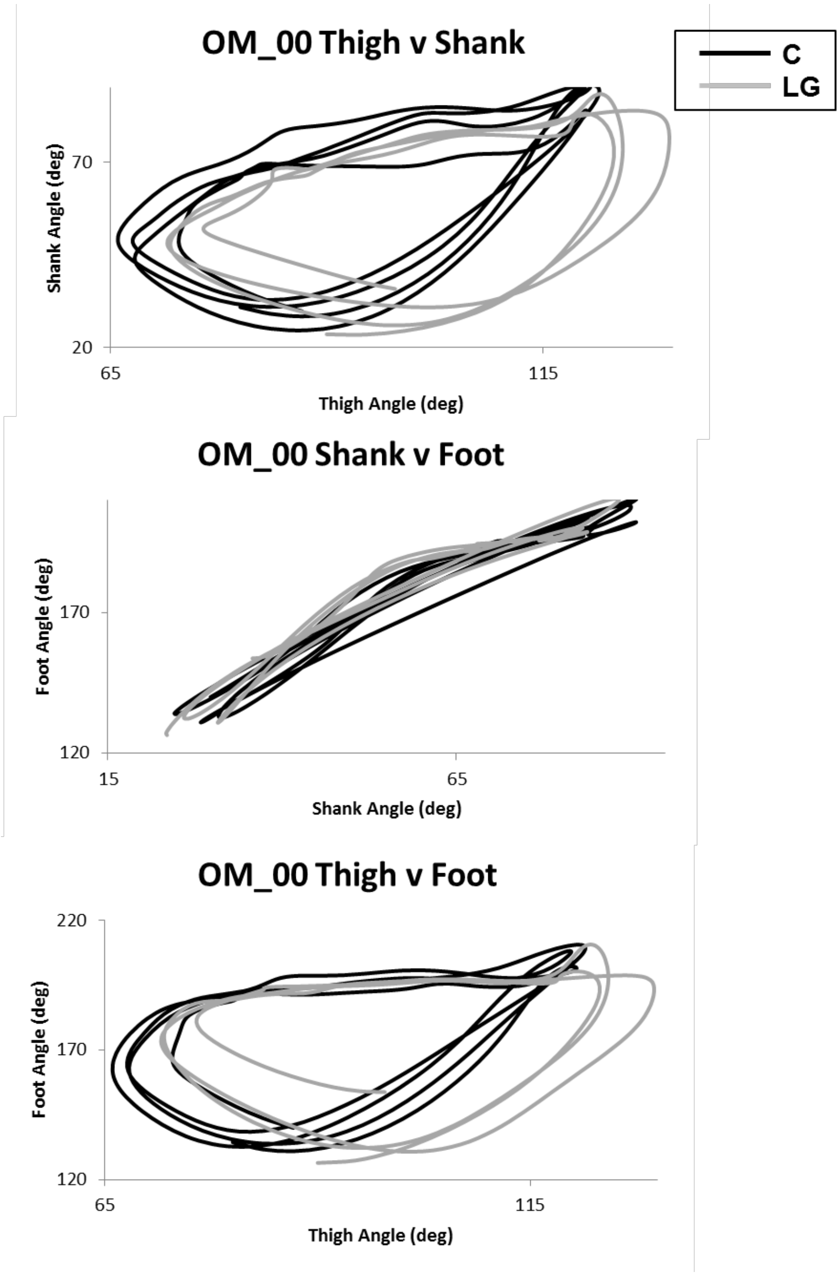
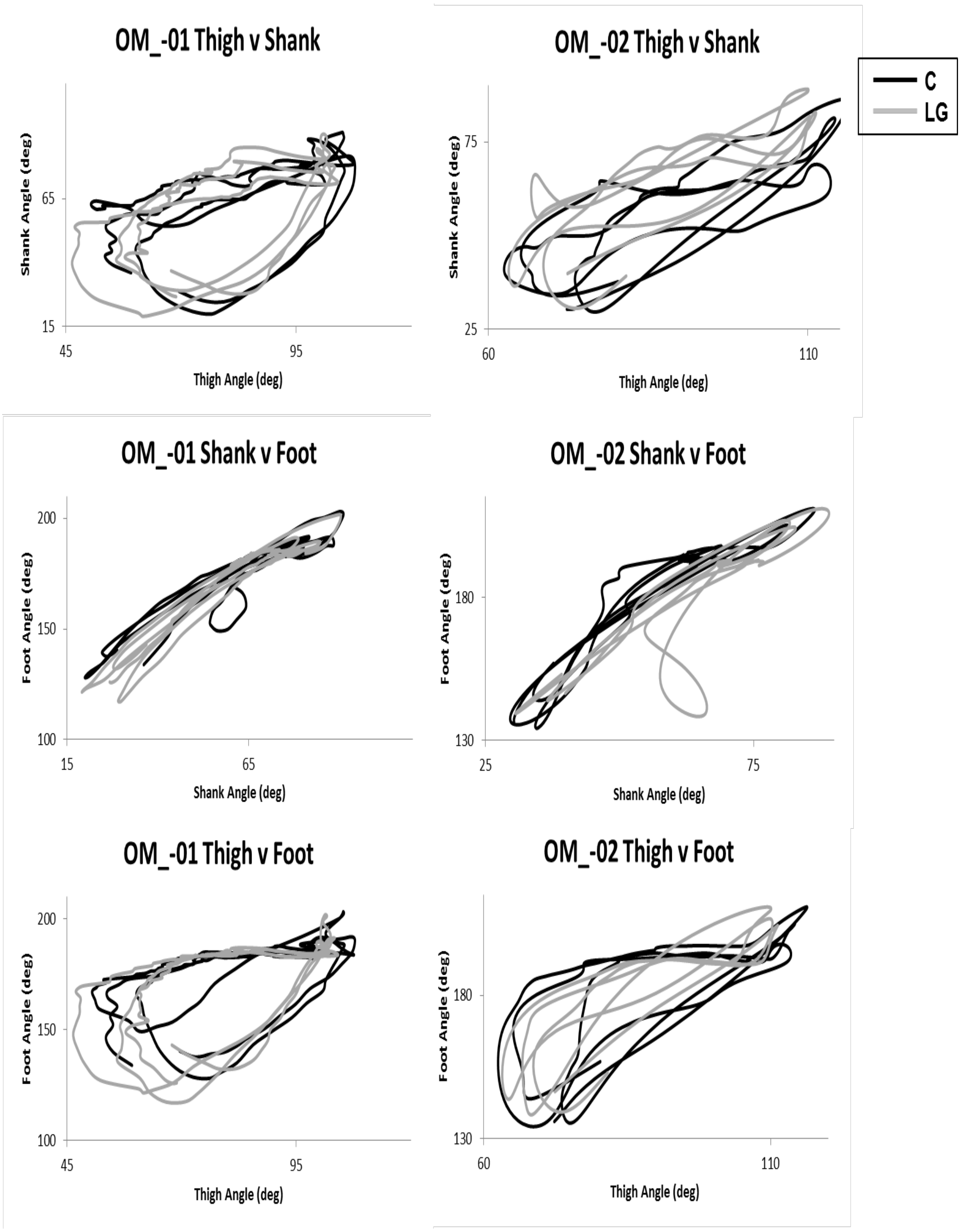


Figure G.1. (continued)



OM_-01 Thigh v Shank

OM_-02 Thigh v Shank

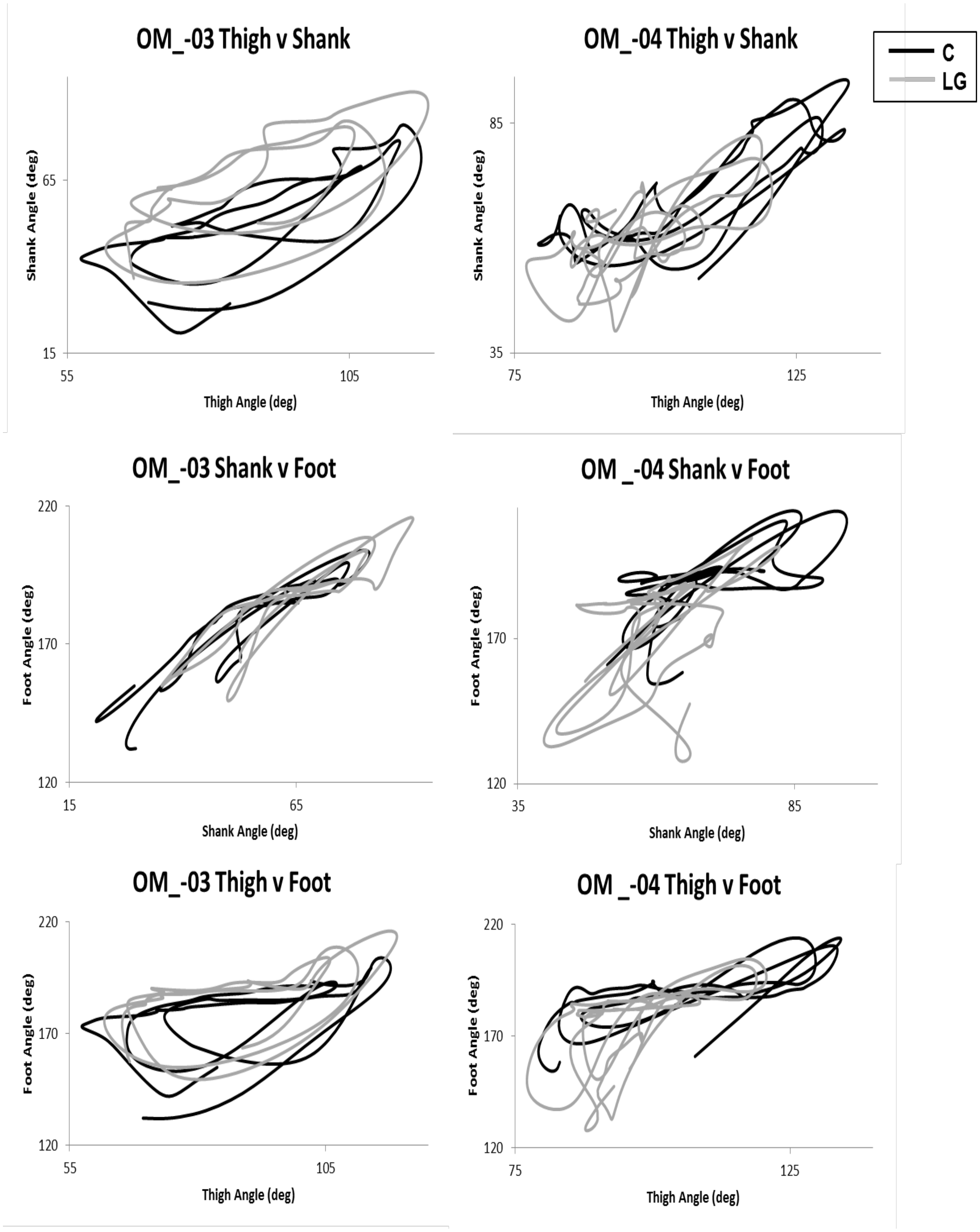
OM_-01 Shank v Foot

OM_-02 Shank v Foot

OM_-01 Thigh v Foot

OM_-02 Thigh v Foot

Figure G.1. (continued)



OM_-03 Thigh v Shank

OM_-04 Thigh v Shank

OM_-03 Shank v Foot

OM _-04 Shank v Foot

OM_-03 Thigh v Foot

OM _-04 Thigh v Foot

Figure G.1. (continued)