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Investigate effects of different step lengths at a preferred walking speed on forefoot and hindfoot motion

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By

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Master's Thesis

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A Thesis
Presented to
The Faculty of
Western Washington University

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Abstract

Hindfoot and forefoot motion during the stance phase of walking provide insights into the forward progression of the body over the feet via the rocker mechanisms. These segmental motions are affected by walking speed. Increases in walking speed are accomplished by increasing step length, cadence, or both. It is unknown if taking short, medium, and long steps at the same speed would also increase hindfoot and forefoot motion similarly to walking speed. We examined effects of different step lengths at the same preferred walking speed on peak forefoot and hindfoot motions related to the foot rocker mechanisms. Twelve young healthy adults completed walking trials under three step length conditions as marker position and force platform data were captured synchronously. Feet and lower extremity motion were measured via marker positions for the combined Oxford foot and conventional gait models. Peak hindfoot and forefoot joint angles associated with the heel, ankle, and forefoot rockers were identified. When walking at the same preferred speed with increase in step length, the peak hindfoot-tibia plantarflexion angle ($p<0.001$) in early stance and dorsiflexion angle ($p=0.016$) in midstance associated with heel and ankle rockers, respectively, increased with step length. The peak forefoot-hallux dorsiflexion angle in late stance indicating forefoot rocker motion also increased with step length ($p=0.004$). When foot kinematics are compared across different individuals or the same individual across different sessions, researchers and clinicians should consider the influence of step length as a contributor to differences in foot kinematics observed.

Keywords: Foot rockers, Oxford foot model, gait

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Manuscript

Introduction

Walking, or ambulation, is a vital activity of daily living required for functional independence. And therefore, it is important to maintain walking abilities in all individuals, especially those who are aging or have disabilities. Older adults and people with disabilities adjust their walking patterns to reduce the demands on their lower extremity muscles and joints. They tend to take shorter steps and walk at a slower speed in comparison to healthy young individuals (Mckeon et al., 2008; Shih et al., 2014; Watelain et al., 2000; Winter et al., 1990). These spatiotemporal gait adjustments remain evident, even when walking at the same speed in older individuals and those with disabilities compared to healthy controls (Buddhadev and Martin, 2016; Cofré et al., 2011; DeVita and Hortobagyi, 2000; Silder et al., 2008). Data from several studies indicate that manipulating walking speed affects foot kinematics (Dubbeldam et al., 2010; Grant and Chester, 2015; Sun et al., 2018; Teixeira-Salmela et al., 2008; van Hoeve et al., 2017). However, no research has examined the effect of step length on foot kinematics when walking speed is controlled.

Understanding the foot kinematics associated with the rocker mechanisms could provide insights into how people adapt foot motion when walking at different step lengths at the same speed. The three rocker mechanisms in the foot that help propel the body forward and maintain forward progression during walking are heel, ankle, and forefoot rockers. From initial contact through the full foot contact on the ground, the foot pivots forward with heel acting as a fulcrum (i.e., heel rocker mechanism). Once the foot is flat on the ground and it is stationary, the tibia pivots forward about the ankle (i.e., ankle rocker mechanism). As the tibia progresses forward the body weight vector is shifted over the forefoot. The foot then pivots forward about the forefoot or the metatarsophalangeal joint as the heel rises (i.e., forefoot rocker mechanism). During the forefoot rocker phase the foot continues pivoting forward from heel rise until toe-off. Among these three rocker mechanisms, the forefoot rocker mechanism associated with the push-off effort of the foot on the ground, makes the largest contribution to forward propulsion (Adams and Cerny, 2018; Perry and Burnfield, 2010; Winter, 1987). Recently, Buddhadev et al. (2020) demonstrated that increasing step length at fixed speeds increased ankle moments associated with the push-off effort in young and older adults. These findings suggest that foot kinematics associated with foot rocker mechanisms, especially the forefoot rocker mechanism, could also be affected when walking at different step lengths at a controlled speed.

The conventional gait model is used often in gait analysis to determine lower extremity and foot kinematics. The major limitation of this model is that the foot is defined as a single rigid segment. This single segment modeling of the foot is not anatomically accurate. From an anatomical perspective, the foot can be separated into three segments: hindfoot, midfoot, and forefoot. From the rocker mechanisms described above, it is apparent that hindfoot and forefoot play distinct roles in heel, ankle, and forefoot rocker mechanisms, respectively. Using the conventional gait model, one can determine the foot kinematics associated with the heel and ankle rockers, but not the forefoot rocker. For example, motion for the forefoot rocker mechanism occurs between the hallux and the forefoot- unfortunately these segments are not a part of the conventional gait model (Baker, 2013; Kadaba et al., 1989; Myers et al., 2004).

One way to determine how the motion associated with the foot rocker mechanisms are affected by different step lengths is to examine foot kinematics using the Oxford Foot Model. The Oxford Foot Model was developed in 2001 as a 4-segment foot-tibia model (Carson et al., 2001). In this model, the foot is separated into the hindfoot, forefoot, and hallux segments. The key difference between the conventional gait model and the Oxford foot model is the separate forefoot and hallux segments defined in the Oxford foot model. Thus, Oxford foot model is a more anatomically accurate model to measure foot kinematics associated with the foot rocker mechanisms (Carson et al., 2001; Levinger et al., 2010; Stebbins et al., 2006).

In summary, it is unknown how different step lengths at identical speed affects motions associated with the foot rocker mechanisms, specifically the forefoot rocker mechanism. This study examined foot kinematics using the multisegmented Oxford foot model to determine these motions. The purpose of this study is to investigate effects of different step lengths at the same walking speed on peak forefoot and hindfoot motions related to the foot rocker mechanisms. We hypothesize peak hindfoot and forefoot joint angles associated with the heel, ankle, and forefoot rockers are larger when walking at longer compared to shorter step lengths.

Methods

Participants

Twelve young healthy adults (6 Females and 6 Males, 22.9 ± 2.5 years, 168.2 ± 9.4 cm, 70.6 ± 17.1 kg) participated in this study. The sample size for this project was determined using GPower 3.1 software. A sample size of 12 participants was needed to achieve a statistical power of 0.8 to detect a large effect size main effect at an alpha level of 0.05. Participants completed a health history questionnaire, Foot & Ankle Disability Index (FADI), and navicular drop test to screen them for conditions (e.g., flat feet), injuries, surgeries, or pain that affect walking or foot motion. The participants total FADI scores were 103.8 ± 0.4 out of 104 points, with scores above 94 indicating no chronic ankle instability or dysfunction associated with foot and ankle function at rest and during various activities (Mckeon et al., 2008; McKeon and Hertel, 2008). Their right and left navicular drop test results were 0.54 ± 0.27 cm and 0.58 ± 0.27 cm, respectively. These data were below the cut off value 1 cm which indicate flat feet (Magee, 2013). Overall, the health history questionnaire results, the FADI scores and the navicular drop test results indicate that participants did not have any conditions, injuries, surgeries, or pain that affect walking or foot motion. The procedures for this study were approved by Western Washington University Institutional Review Board. All participants provided an informed consent prior to participating in the study.

Data collection

All participants completed a single, 90-minutes testing session. The study objectives and procedures were explained to the participants. The participants then completed a health history questionnaire, FADI, and the navicular drop test on the left and right sides. For the navicular drop test, the height of navicular bone from the ground was measured in sitting and standing positions and the differences in these heights was compared against the cut-off of 1 cm which indicate flat feet (Magee, 2013). The participants were instructed to change into spandex tank tops and shorts provided by the researchers. The participants' anthropometric data were measured and recorded next. These measurements were done to determine the inertial properties needed to scale the model and predict joint center positions, segment masses, and segment center of mass locations. Body height and mass were measured using a standard stadiometer. The leg lengths for the right and left sides were measured using a soft tape measure from anterior-superior iliac spine to the medial malleolus on the same side. And a digital caliper was used to measure the horizonal inter-anterior superior iliac spine distance, knee width, and ankle width according to Plug-in Gait Guidelines (Vicon, 2002).

Then 42 reflective markers were attached on both lower extremities and feet of the participants using double-sided tape based on the combined Oxford Foot Model and Conventional Plug-in Gait Model guidelines (Figure 1; (Vicon, 2002)). A static calibration trial was performed with the participant standing in the "motorcycle pose" as per the Vicon Plug-in Gait guidelines. During the static and walking trials, the marker position data were captured at 100 Hz using a 10-camera Vicon motion capture system (Vicon, Centennial, CO).

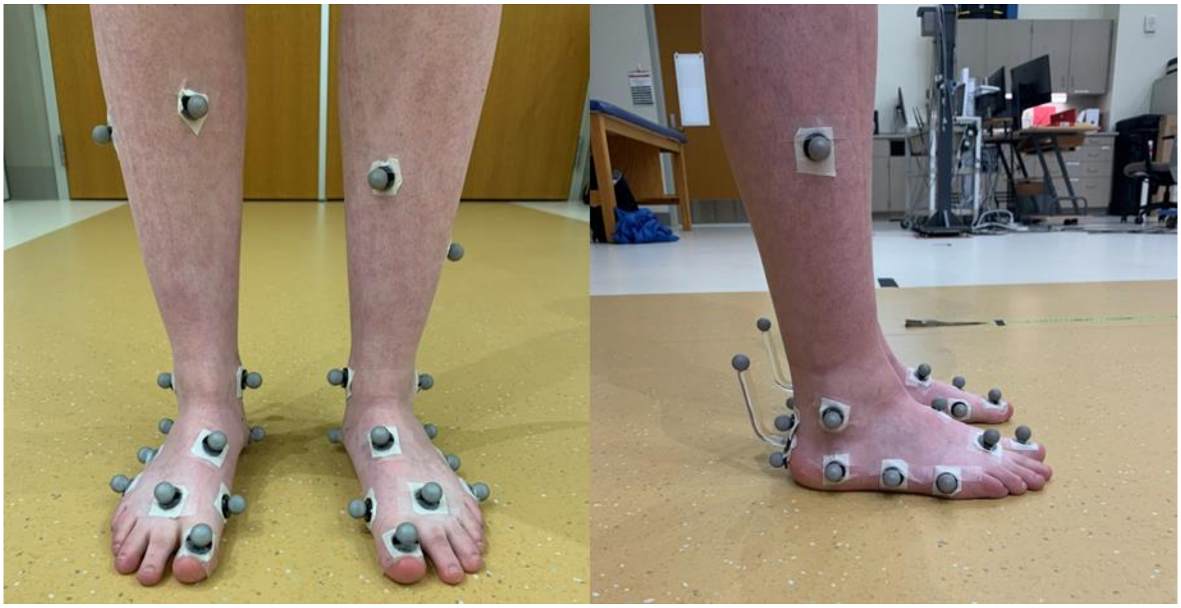

Figure 1. Oxford Foot Model marker placement

The participants’ preferred speed was determined next as they walked overground along a 12-m walkway for five trials. Participants’ preferred walking speed was defined as the speed they would feel comfortable with if they had to walk for a prolonged period. After each trial, the walking speed of the participants was calculated by tracking the position of the virtual sacral marker (located midway between right and left posterior superior iliac spine markers) over a gait cycle. The Vicon Nexus Software (Vicon, Centennial, CO) was used to calculate the step length and step rate they preferred while walking at their preferred speed. Based on the average preferred walking speed, step length, and step rate data, the step rate and step lengths were calculated for the three experimental conditions.

After the preferred walking speed was determined, the participants completed overground walking trials for the three step length conditions in a random order. The three step length conditions were 1) preferred step length (PSL), 2) PSL – 10% of leg length (LL), 3) PSL + 10% of LL. These step length manipulations are nearly identical to those done in previous studies (Allet et al., 2011; Buddhadev et al., 2020; Martin and Marsh, 1992; Umberger and Martin, 2007). To implement the step length conditions, floor tapes were placed at distances corresponding to the calculated step length along the 12-m walkway. Similarly, a metronome was used corresponding to the calculated step rate to help participants maintain the appropriate step rate (Allet et al., 2011). The participants were familiarized with the experimental conditions by performing several practice trials prior to collecting data. The participants completed five walking trials for each step length condition while stepping on the tape marks placed on the floor at the beat of a metronome. After each trial, participants' walking speed was determined to ensure that it was within ±3% of the preferred speed (Buddhadev & Martin, 2016; Buddhadev, Smiley, & Martin, 2020; Barbee et al., 2020; Buddhadev & Barbee, 2020). The walking speed of the participants was calculated by tracking the position of the virtual sacral marker over a gait cycle.

Data analysis

The marker position data for the walking trials were low pass filtered at 6 Hz using a fourth order Butterworth filter (Buddhadev et al., 2020). These filtered data were then processed using the Dynamic Oxford Foot Model and Dynamic Plug-in gait pipelines for a stride (i.e., gait cycle). A stride was defined as two consecutive heel strikes of the right foot. Using this pipeline/program sagittal, frontal, and transverse joint kinematics between the tibia-hindfoot, hindfoot-forefoot, and forefoot-hallux for a gait cycle were computed. Further analysis of the kinematic data was done using a custom MATLAB (Mathworks, Natick, MA, USA) program as follows.

The peak sagittal plane flexion and extension angles were identified for all three joints of the Oxford Foot Model. These angles in the sagittal plane were chosen to better understand how the different step lengths are affecting the foot rocker mechanisms. For the sagittal plane hindfoot-tibia angle, the plantarflexion motion occurring from heel strike to loading response indicates the motion related to the heel rocker mechanism. The extent of this motion was determined by examining peak plantarflexion angle from heel strike to the end of loading response (i.e., from 0-12% of the gait cycle; Perry & Burnfield, 2010). If the magnitude of the heel rocker was larger (as with a longer step length), then the peak plantarflexion angle would be greater. In addition, for the sagittal plane hindfoot-tibia angle, the dorsiflexion motion occurring from the beginning of midstance to early part of terminal stance indicates the motion related to the ankle rocker mechanism. The extent to which step length affects this ankle rocker mechanism was determined by examining peak dorsiflexion angle from beginning of midstance to early part of terminal stance (i.e., from 12-35% of the gait cycle; Perry & Burnfield, 2010). If the magnitude of the ankle rocker is larger (as with a longer step length), then the peak dorsiflexion angle would be greater during this time.

For the forefoot rocker mechanism in the sagittal plane, the dorsiflexion-plantarflexion angle between hallux-forefoot was examined. The time period of interest was from the beginning of terminal stance to end of the stance phase (31-60% of the gait cycle). Peak dorsiflexion angle during this time indicated the magnitude of motion associated with the forefoot rocker mechanism. With increase in step length, it is expected that the peak dorsiflexion angle will increase (Perry and Burnfield, 2010).

Statistical analysis

The primary independent variable of the study was step length (PSL, PSL-10%LL, and PSL+10% LL). The primary dependent variables were i) peak plantarflexion hindfoot-tibia angle indicative of heel rocker mechanism (from heel strike to end of the loading response,0-12% of gait cycle), ii) peak dorsiflexion hindfoot-tibia angle indicative of ankle rocker mechanism (from midstance to early part of terminal stance, from 12-35% of the gait cycle), and iii) peak dorsiflexion forefoot-hallux angle indicative of forefoot rocker mechanism (from beginning of terminal stance to end of the stance phase, 31-60% of the gait cycle). One-way repeated measure ANOVAs were used to assess the effects of step length on these joint angles. Prior to conducting ANOVAs, the data were checked for normality and sphericity using the Shapiro-Wilk test and Mauchly's test of sphericity, respectively. When the assumption of sphericity of data was violated a Greenhouse-Geisser correction was applied to the alpha level. The alpha level of < 0.05 was considered statistically significant. For significant main effect of step length, post-hoc analyses were performed with t-tests. Effect sizes were calculated as partial eta squared (η_p^2), where large effect size was $\eta_p^2 > 0.15$, medium effect size was $\eta_p^2 > 0.06$, and small effect size was $\eta_p^2 > 0.01$ (Weir and Vincent, 2020). All statistical procedures were performed using SPSS version 28.

Results

The participants completed the three experimental conditions (PSL-10%LL, PSL, PSL+10%LL) at their preferred speed which was kept consistent (±3%) across the conditions. The participants' measured walking speed, step length, and cadence are provided in Table 1. Although walking speed for PSL+10%LL condition was significantly greater than PSL-10%LL and PSL conditions, this difference was very small (~1.5%, 0.02 m/s, 0.04 mph) and not meaningful. As expected, step length increased and cadence decreased going from PSL-10%LL, PSL, and PSL+10%LL conditions. These results indicate that the experimental conditions were successfully implemented. The female compared to male participants were shorter in height (p=0.002) and had less body mass (p=0.002). Across the experimental conditions, female participants walked at slower speeds (p<0.05) and at shorter step lengths (p<0.05) compared to the male participants. However, there was no difference in the step rate or cadences across the sexes.

The data for the primary dependent variables (peak plantarflexion hindfoot-tibia angle indicative of heel rocker mechanism, peak dorsiflexion hindfoot-tibia angle indicative of ankle rocker mechanism, and peak dorsiflexion forefoot-hallux angle indicative of forefoot rocker mechanism) are also provided in Table 1. The ensemble average sagittal hind-foot tibia angle and sagittal forefoot-hallux angle are presented in Figure 2. The peak hindfoot-tibia plantarflexion angle associated with the heel rocker mechanism significantly decreased with an increase in step length across the conditions (PSL-10%LL > PSL > PSL+10%LL). These differences in the hindfoot-tibia plantarflexion angle were large in effect size (η_p^2=0.76).

With an increase in step length across the conditions, the peak hindfoot-tibia dorsiflexion angle associated with the ankle rocker mechanism significantly increased with a large effect size. These peak hindfoot-tibia dorsiflexion angles were significantly larger for PSL+10%LL condition compared to PSL-10%LL and PSL conditions. There was a trend (p=0.054) for the peak hindfoot-tibia dorsiflexion angle to be greater for PSL compared to PSL-10%LL condition, however this trend was not statistically significant. Finally, the peak forefoot-hallux dorsiflexion angle associated with the forefoot rocker mechanism significantly increased with an increase in step length across the conditions (PSL-10%LL > PSL > PSL+10%LL).

Table 1. Experimental variables and joint angles associated with the rocker mechanisms.

	PSL-10%LL	PSL	PSL+10%LL	Step Length Main Effect F-value; *p*-value; effect size η_p^2	PSL-10%LL vs. PSL *p*-value	PSL vs. PSL+10%LL *p*-value	PSL-10%LL vs. PSL+10%LL *p*-value
Experimental Variables							
Walking Speed (m/s) *	1.29 ± 0.16	1.29 ± 0.16	1.31 ± 0.16	$F_{2, 22}$=4.21; ***p*=0.028**; η_p^2=0.28	*p*=0.818	*p*=0.035	*p*=0.020
Step Length (m) *	0.60 ± 0.07	0.69 ± 0.07	0.78 ± 0.06	$F_{2, 22}$=351.53; ***p*<0.001**; η_p^2=0.97	*p*<0.001	*p*<0.001	*p*<0.001
Cadence (steps/min) *	129.0 ± 9.9	112.8 ± 6.2	100.6 ± 6.2	$F_{2, 22}$=176.39; ***p*<0.001**; η_p^2=0.94	*p*<0.001	*p*<0.001	*p*<0.001
Primary Dependent Variables							
Heel Rocker-Peak Hindfoot-Tibia Plantarflexion Angle Early Stance (deg) *	4.1 ± 4.7	2.3 ± 4.2	1.3 ± 4.0	$F_{2, 22}$=34.34; ***p*<0.001**; η_p^2=0.76	*p*<0.001	*p*=0.004	*p*<0.001
Ankle Rocker-Peak Hindfoot-Tibia Dorsiflexion Angle Mid Stance (deg) *	18.7 ± 4.2	20.1 ± 5.5	21.3 ± 6.1	$F_{1.2, 13.2}$=6.97; ***p*=0.016**; η_p^2=0.39	*p*=0.054	*p*=0.016	*p*=0.016
Forefoot Rocker – Peak Forefoot-Hallux Dorsiflexion Angle Late Stance (deg) *	46.6 ± 9.4	48.9 ± 11.1	51.2 ± 11.9	$F_{1.3, 14.4}$=9.88; ***p*=0.004**; η_p^2=0.47	*p*=0.020	*p*=0.016	*p*=0.006

Data presented as mean ± 1 standard deviation. *statistically significant step length main effect (*p*<0.05).

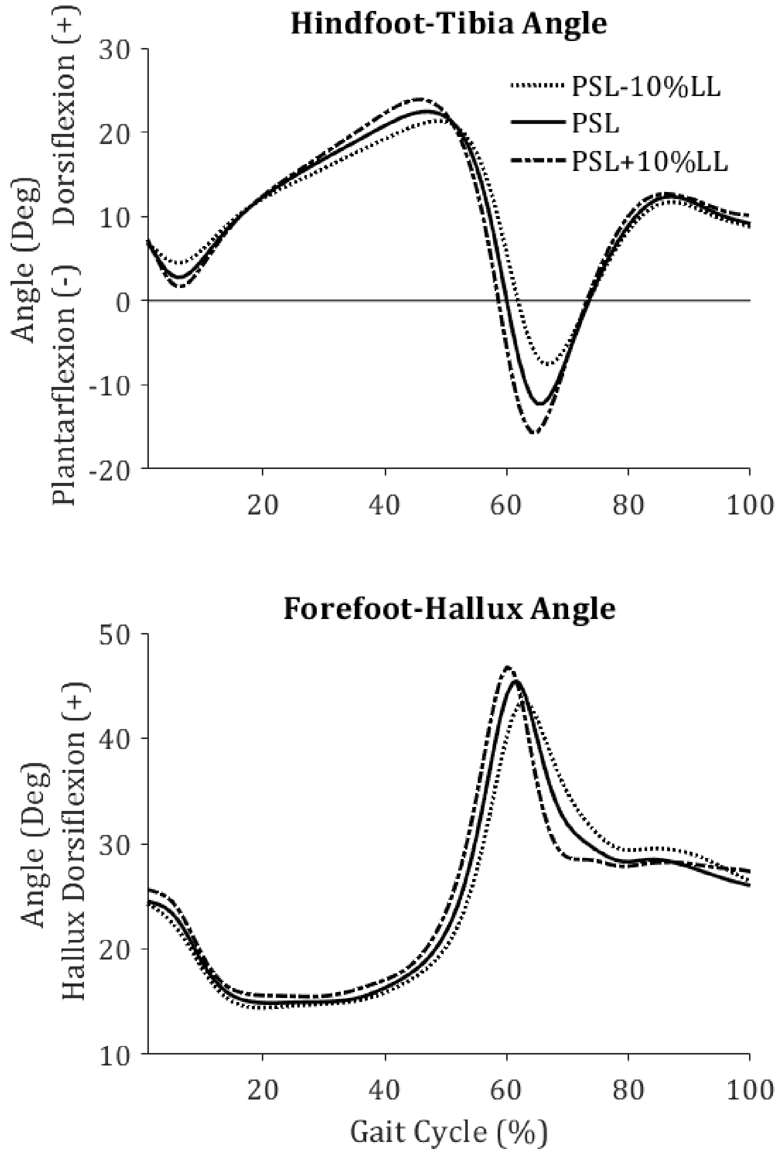

Figure 2. Ensemble average sagittal hindfoot-tibia angle and forefoot-hallux angle across the experimental conditions. The gait cycle begins with initial contact of the right heel and ends with subsequent contact with the right heel (approximately 0-60% is the stance phase and 60-100% is the swing phase).

Discussion

The purpose of this study was to investigate effects of different step lengths at the same preferred walking speed on peak forefoot and hindfoot motions related to the foot rocker mechanisms. We hypothesized these peak foot angles would increase with increase in step length. The data from the current study support these hypotheses. With increase in step length, increases were observed in i) peak plantarflexion hindfoot-tibia angle in early stance for the heel rocker, ii) peak dorsiflexion hindfoot-tibia angle in midstance associated with the ankle rocker, and iii) peak dorsiflexion hallux-forefoot angle in late stance related to the forefoot rocker. Understanding these effects of step length on kinematics associated with the foot rockers are important as these variables provide insight into the forward progression and stability of the body over the feet in the stance phase (Perry and Burnfield, 2010).

To the authors' knowledge, this is the first study to examine the effects of step length at controlled preferred speeds on foot kinematics. The data show that modest manipulations of step length (i.e., ±10%LL or about ±3.5 inches) affect foot kinematics (see Table 1 and Figure 2). Over the range of step length conditions (i.e., from PSL-10%LL to PSL+10%LL) the joint angles associated with the foot rockers changed but these changes were not equal across the rockers. There were larger changes in peak angle for the forefoot rocker compared to heel and ankle rockers. Specifically, the peak hindfoot-tibia plantarflexion angle in early stance related to the heel rocker changed by approximately 2.7° over the range of step length conditions. Similarly, there was a change of 2.7° over the range of step length conditions for peak hindfoot-tibia dorsiflexion angle in midstance for the ankle rocker. However, for the peak forefoot-hallux dorsiflexion angle in late stance associated with the forefoot rocker the change was 4.7°. The larger increment for the forefoot rocker could be due to greater push-off moment and power applied during the late stance phase to produce a longer step length (Allet et al., 2011; Buddhadev et al., 2020; Umberger and Martin, 2007).

Individuals with foot disabilities and older adults walk taking shorter steps at slower speeds compared to healthy adults (Kerrigan et al., 1998; Shih et al., 2014; Watelain et al., 2000; Winter et al., 1990). In the current study we found that step length affects foot kinematics and similarly, several studies have shown walking speed to also affect foot kinematics (Dubbeldam et al., 2010; Sun et al., 2018; Tulchin et al., 2009; van Hoeve et al., 2017). During clinical gait analysis, when foot kinematics are compared across different individuals (e.g., adults with or without disabilities) or the same individual across different sessions (i.e., to monitor disease progression or determine effectiveness of an intervention), researchers and clinicians should consider the influences of speed and step length as contributors to differences in foot kinematics observed in these scenarios.

A merit of the current study includes the unique intervention assigned to the participants which we think is high on internal validity. In the current study, the participants walked at their freely chosen speeds and step lengths. The researchers manipulated the participants' walking by having them walk taking steps which were shorter or longer by 10% of their leg length. The manner of walking observed as the participants walked under these different step length conditions could be similar to how individuals naturally manipulate step lengths during their everyday lives. In contrast to this intervention, had the researchers chosen to have participants walk at fixed speeds with fixed increments or decrements in their step lengths, such walking conditions may or may not be similar to how people walk in daily life, or these walking conditions may not have felt natural for people of different heights.

The foot kinematics data (i.e., joint angle values and profile shapes) observed in the current study are comparable to published research (Carson et al., 2001; Dubbeldam et al., 2010; Nicholson et al., 2018; Stebbins et al., 2006; Sun et al., 2018; Tulchin et al., 2009; Wright et al., 2011). For example, peak hindfoot-tibia plantarflexion angle during early stance in the current study ranged from 1.3-4.1° which were approximately 1-4° larger than that reported by others (Stebbins et al., 2006; Sun et al., 2018; Tulchin et al., 2009). Similarly, the peak hindfoot-tibia dorsiflexion angle during midstance in the current study ranged from 18-21° which are in the range of values reported by (Sun et al., 2018)), who reported values of 17-29° for a range of slow to fast walking speeds. Lastly, the peak forefoot-hallux angle in late stance in the current study ranged from 46-51° which was in the range of values reported by (Dubbeldam et al., 2010) who reported values of 42-51° for a range of slow to comfortable speeds.

Since the current study was the first to examine the effect of step length at controlled preferred speeds on foot kinematics, the authors were not able to compare these data with other studies. However, as walking speed increments are accomplished via increases in step length and cadence, the trends (i.e., increments) in foot joint angles with longer step lengths could be similar to increases in these angles with faster walking speeds. For example, (Dubbeldam et al., 2010) reported an increase of 6° in peak hindfoot-tibia plantarflexion angle during early stance when walking from slow to comfortable speeds. Similarly, Sun et al., 2018 reported an increase of 11° for peak hindfoot-tibia dorsiflexion angle in midstance when walking from slow to fast speeds. And for peak forefoot-hallux angle in late stance with increases in walking speeds, Dubbeldam et al. (2010) reported an increase of 9°. In summary, these trends indicate that changes in foot kinematics observed with step length increments in the current study are similar to trends reported in previous studies for increases in walking speed.

A limitation of the current study is that these data are only generalizable to young healthy adults. These data may or may not apply to children, older adults, or individuals with disabilities. Future studies could consider examining the effect of step length on foot kinematics in these populations. Another limitation of the study is that we examined kinematic variables which provide information on the foot motion of the participants during the experimental conditions. These kinematic variables do not provide insights into factors causing changes in kinematics with step length. To determine the factors causing these changes, kinetic variables such as moments and powers at these joints should be examined.

In conclusion, increases in step length at the same preferred walking speed increased peak forefoot and hindfoot motions related to the foot rocker mechanisms. During clinical gait analysis, when foot kinematics are compared across different individuals or the same individual across different sessions, researchers and clinicians should consider the influence of step length as a contributor to differences in foot kinematics observed.

Literature Review

Overview

This review will explore hindfoot and forefoot motion during walking. It is important to know the gait functions, parameters, and phases during walking. As the foot progresses through a gait cycle, there are three foot rocker mechanisms that assist with forward propulsion during walking. This review will thoroughly explain why the conventional gait model is not an appropriate biomechanical model when exploring foot motion associated with these foot rocker mechanisms. Instead, the Oxford Foot Model has been shown to be a reliable and valid model in demonstrating overall foot motion. The Oxford Foot Model is a more anatomically accurate model as it breaks up the foot into three segments. Foot kinematics for these three segments during normal walking will be described in the context of foot rocker functions. This review will also explore the effects of walking speed on kinematics of these foot segments using the Oxford Foot Model. It is currently unknown how walking at different step lengths at the same speed affects foot motion and foot rocker mechanisms during walking in healthy young adults. This review will provide a convincing rationale for examining internal foot motion at different step lengths during walking.

Introduction to gait functions, parameters, and phases

Walking gait has five main motor functions that must be performed during a gait cycle: generation of mechanical energy to achieve forward velocity, shock absorption for stability, maintenance of support of the trunk and upper body, maintenance of upright posture and balance of the total body, and safely control foot trajectory (Perry and Burnfield, 2010). These functions are vital in the walking gait cycle to transport the body safely and efficiently. The primary external propulsive force during walking is the forward fall of body weight. The internal propulsive forces associated with the leg pushing on the ground are produced via ankle plantarflexion, knee extension, and hip extension movements. Hip flexion motion is created by an anterior pulling force that also assists in forward propulsion (Buddhadev et al., 2020; Buddhadev and Martin, 2016; DeVita and Hortobagyi, 2000; Winter et al., 1990).

These internal propulsive forces generated by muscles acting about the lower extremity joints help an individual progress forward from one step to another. During a gait cycle (i.e., a stride), parameters of walking speed, step length, and step rate are interrelated. Step length is the distance measured from a specific point on one foot to the same point on the other foot when both feet are in contact with the ground. Stride length is the distance measured from initial contact of one foot to the next initial contact of the same foot. A stride is also, the sum of the left and right step lengths. Walking speed (meters/second) is a product of step length (in meters) and step rate (steps per second) (Adams and Cerny, 2018; Baker, 2013; Perry and Burnfield, 2010). Owing to this relationship, walking speed is influenced by step length and step rate. Walking speed increases with increase in step length and step rate (Buddhadev et al., 2020; Teixeira-Salmela et al., 2008).

Phases of a gait cycle

A walking gait cycle is represented by a single stride, consisting of both stance and swing phases. The stance period takes place during which the foot is in contact with the ground. The stance period consists of weight acceptance, midstance, late stance, and pre-swing phases. It starts with weight acceptance, which is the period between initial contact and maximal knee flexion of the support limb. Mid stance is the period between weight acceptance and heel rise associated with the push off effort. Late stance starts with heel rise of the support limb and ends with heel strike of the contralateral limb. During this phase, the foot pushes off on the ground and this motion is associated with a rapid ankle plantarflexion movement. Pre-swing is the last phase of the gait cycle during which both feet are in contact with the ground. During this phase the trailing limb transfers body weight to the leading limb to unload the trailing limb and prepare it for the swing phase. The stance period is 58-61% of total stride period (Winter, 1987).

During the stance phase, there are periods of single and double limb support. Single support is the period when one limb is in contact with the ground. It is equal to the swing period of the contralateral limb. Double support occurs when both feet are in contact with the ground, between initial contact of one limb and toe-off of the opposite limb. Double support varies from 16-22% of the total stride period (Winter, 1987). The goal of double support is to transfer body weight from one limb to the other.

The swing period is the period when the foot is not in contact with the ground and swinging forward to position the limb for the next heel strike. The swing period is 39-42% of total stride period and it consists of early, mid, and late swing phases (Winter, 1987). Early swing starts with toe off of the trailing limb, and it continues until swing phase is in line with the contralateral limb. Mid-swing phase is the mid third of the swing phase. It spans from the end of the initial swing phase to the time when swing leg is ahead of the contralateral limb and the shank segment is vertical. Late swing is the period beginning with the end of mid swing phase and it ends with heel strike of the swing leg (Adams and Cerny, 2018; Baker, 2013; Perry and Burnfield, 2010; Winter, 1987).

Tasks accomplished during a gait cycle

During a gait cycle, there are different tasks accomplished at different points of the cycle. The first task is weight acceptance which is accomplished in the initial contact and loading responses parts of the early stance phase. This is the most challenging task because when the heel of one foot contacts the ground, it needs to absorb the shock of impact while simultaneously providing initial limb stability, and the preservation of forward progression (Perry and Burnfield, 2010). Initial contact is the foot making its first contact with ground and is the start of the double support period. It initiates the heel rocker mechanism and decelerates the impact. During initial contact, the hip is flexed, knee is extended, and the ankle is dorsiflexed to the neutral position. Heel contact initiates a five-degree arc of the plantarflexion motion. Initial contact takes place between the zero to two percent of the gait cycle.

Loading response is the next phase during which body weight is transferred from the trailing to the forward limb. During this time, the heel is used as a pivot point or rocker for maintaining forward progression. As the foot is going through the heel rocker mechanisms, the knee is flexed, ankle is plantar-flexed, and subtalar joint is everted. This subtalar eversion, which occurs during loading response, is the second way the body absorbs the shock of heel strike which starts the gait cycle (Perry and Burnfield, 2010). Loading response takes place between the 2-12 % period of the gait cycle (Adams and Cerny, 2018; Perry and Burnfield, 2010; Winter, 1987).

The next task in the gait cycle is single leg support. At beginning of midstance phase, one limb is supporting the body weight while the contralateral limb is in swing phase. Mid-stance is the first half of the single leg support occurring from 12-31% of the gait cycle. The foot is stationary and flat on the ground, while the body pivots forward about the ankle via the ankle rocker mechanism. During this phase the ankle goes through dorsiflexion, and the knee and hip are extending. The midtarsal (MT) joint, formed by the talonavicular and calcaneocuboid articulations, is also dorsiflexed during this time. The foot then goes through ankle plantarflexion motion which manifests as rise of the heel of the support limb. This heel rise marks the end of the midstance and start of the terminal stance phase. Terminal stance is the second half of the single leg support occurring from 31-50% of the gait cycle. It begins with heel rise and ends with the contralateral limb's initial contact. During terminal stance, the knee completes extension and begins flexion, and the hip continues extension. The ankle continues plantarflexion as the subtalar joint has reduced eversion and the MT joint locks. This locking of the MT joint during terminal stance and pre-swing phase is important to make the foot segment function as a rigid lever for push off against the ground. From the midportion of terminal stance through the whole pre-swing phase the foot pivots forward about the forefoot or hallux segment. This pivoting is also known as the forefoot rocker mechanism and it is crucial for push-off required for maintaining forward progression (Adams and Cerny, 2018; Perry and Burnfield, 2010; Winter, 1987).

The last task of the gait cycle is swing limb advancement. This starts with the preparation of swing and ends with initial contact. Pre-swing begins with initial contact of the contralateral limb and ends with ipsilateral toe-off. Weight transfer from stance limb to contralateral limb increases dorsiflexion at the metatarsophalangeal joint, ankle plantarflexion and knee flexion and decreases hip extension of the ipsilateral limb. Pre-swing takes place during the 50-62% of the gait cycle. Initial swing starts with toe-off and ends with ipsilateral limb advancing forward to be in line with the contralateral limb. During this phase the knee flexion is increased for foot clearance. The hip also flexes to advance limb forward. Initial swing occurs at 62-75% of the gait cycle. Mid-swing starts at the end of initial swing and ends with the limb forward of the contralateral limb and the tibia vertical. Hip flexion and knee extension continue to advance limb forward and clear foot from the ground. Ankle is dorsiflexed to neutral position. Mid-swing occurs at 75-87% of the gait cycle. Terminal swing starts with the tibia in a vertical position and ends with initial contact. Limb advancement is completed during this phase and the limb is prepared for stance. Knee continues to extend and hip flexes to twenty degrees. Terminal swing occurs from 87-100% of the gait cycle, making it the last phase of the cycle (Adams & Cerny, 2018; Perry & Burnfield, 2010; Winter, 1987).

Foot Rocker Mechanisms

There are three foot rocker mechanisms that help propel the body forward during walking (Perry & Burnfield, 2010). These foot rocker mechanisms are separated into three segments: consisting of heel, ankle, and forefoot rocker.

During the heel rocker, the heel functions as a fulcrum as the foot pivots forward about the heel to lay flat on the ground after heel strike. The heel rocker mechanism starts with initial contact of the calcaneal tuberosities. As the heel rocks forward, the force of falling is redirected to forward motion. The quadriceps draw the femur forward at a slower rate than the advancement of the tibia, while restraining the rate of knee flexion. The heel rocker ends at the contact of the entire foot on the ground, the tibia being vertical, the knee flexed at twenty degrees, and the ankle is plantarflexed at five degrees. The dorsiflexor muscles eccentrically control the forward motion of the tibia during the heel rocker mechanisms (Perry and Burnfield, 2010).

The ankle rocker is next in the gait cycle. The ankle is the fulcrum due to the foot being stationary, flat on the ground, and the tibia pivots and progressing forward via eccentric ankle dorsiflexion. The soleus contracts to stabilize and slow down the advancement of the tibia. The gastrocnemius assists the soleus in allowing tibia advancement (Perry and Burnfield, 2010).

The forefoot rocker mechanism is last in the gait cycle. The metatarsophalangeal joint act as the fulcrum as the heel rises. The gastrocnemius and soleus muscles stabilize the ankle, and their contraction produces a vigorous push-off on the ground. The propelling forces are strongest during this mechanism as the leg propels forward (Perry and Burnfield, 2010). During gait, the foot segment also acts as a dynamic base of support. The source of support changes during the gait cycle. The heel is the first point of contact with the ground and is the only source of support for the first six to twelve percent of the gait cycle. As the gait cycle advances, the fifth metatarsal head is the next point of contact with the ground and progresses to include all five metatarsal heads going from lateral to medial. This is called the foot-flat support and is about twenty percent of the total gait cycle. The forefoot support starts at the onset of the heel rise that leads to the metatarsal heads and toes to be in contact with the ground. It commonly ends with the first metatarsal and big toe being the last source of support before toe-off (Perry & Burnfield, 2010). The foot rocker mechanisms are a key aspect to the foot motion during walking.

Motion capture for walking relevant to foot motion

Kinematic data of walking gait is typically captured using a 3D motion capture system. Reflective markers are placed on the body and motion of these markers are tracked using a motion capture system. Using the track position of these markers, a biomechanical model of the human body is created. The most accepted biomechanical model is the conventional gait model. It models the lower limb segments as rigid levers and is the most widely used model for clinical gait analysis. Angular motion occurring at the hip, knee, and ankle joints throughout a walking gait cycle can be computed from the position of the reflective markers. In the conventional gait model using the marker position data, body segments are defined as rigid bodies. These rigid bodies include the left/right foot, left/right tibia, left/right femur, and pelvis. For each segment there are three axes: mediolateral axis (X), anteroposterior axis (Y), and longitudinal axis (Z). The segments are connected by joints that are assumed to be ball and socket joints, giving each joint three degrees of freedom. The kinematic outputs of the conventional gait model are the joint angles that describe the orientation of the distal segment in relation to the proximal segment for each rigid body.

The conventional gait model was first created in the 1980s and has significant limitations (Baker, 2013). The most significant limitation is in in relation to the foot. In this model the foot is defined as a rigid segment and has a single axis at the ankle. This simplistic model of the foot allows calculation of ankle motion, but it is unable to measure motion in the hindfoot, midfoot, and forefoot segments. As previously explained for the foot rockers mechanisms, there are distinct motions that occurs in the forefoot, midfoot, and hindfoot segments. Because these motions cannot be measured via the conventional gait model, it may not give an accurate representation of foot motion during walking. This is more true in the presence of a foot deformity due to the model not giving accurate measures of the alignment of the hindfoot (Baker, 2013; Kadaba et al., 1989; Myers et al., 2004).

Assessing the foot as a single segment model does not allow for determination of forefoot, midfoot, and hindfoot motion. The foot has three distinct segments which go through different amounts of motion during a walking gait cycle. The Oxford foot model was created in 2001 as a multi-segment foot model (Carson et al., 2001). In this model, the foot is separated into three segments (hindfoot, forefoot, and hallux) plus a tibial segment. The tibial segment of the Oxford foot model is made up of the tibia and fibula and is assumed to move as a single rigid body. This segment is defined as the line from the knee joint center and the ankle joint center. The knee joint center is the center point between the femoral condyle, tibial tuberosity, and the head of the fibula. The ankle joint center is located between the lateral and medial malleolus markers. The placement of the five markers on this segment are the head of the fibula, tibial tuberosity, anterior aspect of the tibia, and lateral and medial malleoli (Carson et al., 2001; Levinger et al., 2010; Stebbins et al., 2006).

The hindfoot segment is made up of the calcaneus. As the hindfoot is defined as a rigid segment, the motion at the talocrural and subtalar joints are considered to equally contribute to the motion of the hindfoot relative to the tibia. The segment is defined as the line from the posterior surface of the calcaneus and the midway point between the medial and lateral calcaneus. To define to hindfoot segment, markers are placed on the posterior distal and proximal aspects of the calcaneus, the medial and lateral aspects of the calcaneus, and a wand marker on the posterior calcaneus midway between the two markers on the posterior aspects of the calcaneus (Carson et al., 2001; Levinger et al., 2010; Stebbins et al., 2006).

The forefoot segment is made up of the five metatarsals that are assumed to move as a single rigid body. The segment is defined as the midpoint of proximal heads of the first and fifth metatarsals to the midpoint of the distal heads and the second and third metatarsals. Markers are placed at the bases and heads of the first and fifth metatarsals, and between the second and third metatarsal heads (Carson et al., 2001; Levinger et al., 2010; Stebbins et al., 2006).

Finally, the hallux segment includes the proximal phalanx of the hallux and is defined as the longitudinal line along the proximal phalanx. A marker is placed on the medial side of the hallux proximal phalanx. The Oxford foot model can also be combined with the Plug-in Gait model to obtain markers on the lower extremities and feet. Additional markers are placed on the sacrum, left and right thigh, and left and right anterior iliac spine (van Houve et al., 2015). This model classifies the foot segments slightly different than the anatomical classifications of the foot segments (Carson et al., 2001; Stebbins et al., 2006).

The Oxford foot model has been shown to be reliable in measuring both barefoot and shod walking (Balsdon and Dombroski, 2018; Curtis et al., 2009; Stebbins et al., 2006; Wright et al., 2011; Yoo et al., 2022) There are other multi-segment foot models, such as duPont, Heidelberg and Rizzoli (Caravaggi et al., 2011; Leardini et al., 2007; Nicholson et al., 2018; Simon et al., 2006). When comparing the gait kinematic results of the different models, duPont and the Oxford foot model were shown to have the most similarities (Nicholson et al., 2018). In a systematic review assessing the quality of different single- and multi-segment foot models to propose, the Oxford foot model was one of two models that have demonstrated external validity (Bishop et al., 2012). Overall, the Oxford foot model has proven to be a useful and versatile tool for the study of foot mechanics and gait.

Anatomy of the Foot and Internal Foot Motion during a Gait Cycle

Segments and Joints

The Oxford foot model is a more anatomically accurate model in capturing foot motions compared to the single axis foot segment in the conventional gait model. The following section will explain the complex motions that occurs within the foot segment in all three planes. This explanation is provided to reinforce the position, that a single axis foot segment used in the conventional gait model does not adequately capture the internal foot motion occurring during walking gait.

The foot is broken into three functional subdivisions: hindfoot, midfoot, and forefoot. The hindfoot is made up of the calcaneus and talus. The joints in the hindfoot are the talocrural and subtalar joints. The talocrural joint is a uniaxial hinge joint, made up of the tibia and fibula to form the tibiofibular joint and the tibia and talus to make up the tibiotalar joint. Plantarflexion and dorsiflexion are the primary movements at this joint. The subtalar joint consists of the articulation between the talus and calcaneus. This joint can move in all three planes. The primary movements at this joint are pronation and supination, which includes calcaneal eversion/inversion, abduction/adduction, and dorsiflexion/plantarflexion (Fraser et al., 2016; Hamill et al., 2015; Schuenke et al., 2014). At initial contact, the calcaneus is 2 degrees inverted and the subtalar joint is allowing for calcaneal supination. During the loading response and early mid stance, the calcaneus everts, and the subtalar joint motion is pronation. Continuing on to late mid stance, terminal stance, and pre-swing, the calcaneus inverts and the subtalar joint produces supination (Adams and Cerny, 2018).

The midtarsal joint is the joint between the midfoot and hindfoot. It consists of the calcaneocuboid and talonavicular joints. The movement of the midtarsal joints depends on the position of the subtalar joint. When the subtalar joint is in pronation, the calcaneocuboid and talonavicular axes of rotation are parallel. This leads to the midfoot having the ability to move more freely and allows for shock absorption from heel strike to foot flat. When the subtalar joint is in supination, the two midtarsal axes intersect, locking the midtarsal joint. The locking of the midtarsal joint creates a rigid lever that is used to propel the foot forward from foot flat to toe-off (Hamill et al., 2015; Perry and Burnfield, 2010; Schuenke et al., 2014). At initial contact, the midtarsal joint is locked due to the subtalar joint in supination. During loading response, the midtarsal joint unlocks, increases flexibility in the forefoot, and absorbs sock. Unlocking of the midtarsal joint continues through early mid stance. The midtarsal joint starts to lock during the late mid stance phase and becomes locked at terminal stance, suppressing midfoot motion. Pre-swing allows for the midfoot to unload (Adams and Cerny, 2018).

The midfoot is made up of the cuboid, navicular, and cuneiforms. The joints in the midfoot include the cuneonavicular, intercuneiform, and cuneocuboid joints. The primary movements of these joints are gliding and slight rotation to alter the shape of the transverse arch (Hamill et al., 2015; Perry and Burnfield, 2010; Schuenke et al., 2014).

The tarsometatarsal joint separates the forefoot and midfoot. This joint is the articulation between the cuneiforms and metatarsals as well as between the cuboid and metatarsals. Flexion and extension are the primary movements at this joint and contribute to inversion and eversion of the foot. These movements change the shape of the arch (Hamill et al., 2015; Perry and Burnfield, 2010; Schuenke et al., 2014).

The forefoot is made up of the metatarsals and phalanges. The joints in the forefoot are the metatarsophalangeal and interphalangeal joints. The metatarsophalangeal joints allow for flexion/extension and abduction/adduction of the phalanges. The interphalangeal joints allow for flexion and extension in the phalanges. At initial contact, the forefoot is adducted. During loading response and early mid stance, the forefoot abducts. Once the foot is in late mid stance, the forefoot starts to adduct as the foot rocks forward and the forefoot rocker mechanism comes into play. The forefoot stays adducted through terminal stance and pre-swing (Adams and Cerny, 2018; Hamill et al., 2015; Perry and Burnfield, 2010; Schuenke et al., 2014).

Muscles

The triceps surae is made up of the soleus and the medial and lateral heads of the gastrocnemius. The soleus originates at the posterior surface of the head and neck of the fibula. The medial and lateral heads of the gastrocnemius originate at the medial and lateral epicondyles of the femur, respectively. The triceps surae inserts at the calcaneal tuberosity via the Achilles' tendon. The main action of the triceps surae is plantarflexion at the talocrural joint. The triceps surae produces the majority of the plantarflexion force, and the tibialis posterior, flexor digitorum longus, flexor hallucis longus, and the peroneus longus and brevis assist with plantarflexion (Schuenke et al., 2014).

The tibialis posterior originates on the upper posterior tibia, fibula, and interosseous membrane and inserts at the inferior navicular. Its primary actions are inversion and ankle plantarflexion. The flexor digitorum longus originates at the posterior tibia and inserts at the distal phalanx of toes 2-5. Its primary action is flexion of the toes 2-5 and assists with ankle plantarflexion and inversion. The flexor hallucis longus originates at the distal end of the posterior surface of the fibula and the interosseous membrane.

It inserts at the base of the distal phalanx of the hallux. Its primary action is flexion of the big toe and assists with ankle plantar flexion. The peroneus longus originates at the lateral condyle of the tibia and upper lateral fibula. It insets at the medial cuneiform and lateral first metatarsal. The peroneus brevis originates at the lower lateral fibular and inserts at the lateral fifth metatarsal. Both the peroneus longus and brevis primarily everts and assists with ankle plantarflexion (Hamill et al., 2015; Schuenke et al., 2014).

The three dorsiflexor muscles in lower extremity and foot are the tibialis anterior, extensor digitorum longus, and extensor hallucis longus (Hamill et al., 2015; Schuenke et al., 2014). Dorsiflexion at the ankle is crucial during the swing phase of gait for foot clearance and plantarflexion during the stance phase for controlled lowering of the foot after heel strike. The tibialis anterior originates on the upper lateral tibia and interosseous membrane and inserts on the medial plantar surface of the first cuneiform. Along with ankle dorsiflexion, the tibialis anterior assists with inversion. This is the strongest dorsiflexor muscle. The extensor hallucis longus originates at the anterior fibula and interosseous membrane and inserts at the distal phalanx of the big toe. It dorsiflexes the hallux, adducts the forefoot, and assists with ankle dorsiflexion. The extensor digitorum longus originates at the lateral condyle of the tibia, fibula, and interosseous membrane. It inserts at the dorsal expansions of toes 2-5. The primary action of the extensor digitorum longus is dorsiflexion of toes 2-5 and eversion. It assists with ankle dorsiflexion (Hamill et al., 2015; Schuenke et al., 2014).

Foot kinematics during walking

This next section will describe foot motion during walking as defined by the Oxford foot model. The inter-segment joints of the Oxford foot model measure the angles of rotation for each segment. The hindfoot/tibia joint shows movement of the hindfoot with respect to the tibia. The movements that can occur at the hindfoot/tibia joint are the plantar/dorsiflexion movement about the transverse axis of the tibia, inversion/eversion about the posterior/anterior axis of the hindfoot, and the internal/external rotation about the common perpendicular axis. At initial contact, the hindfoot is neutral or dorsiflexed, then after a brief period of plantarflexion associated with heel rocker, it then dorsiflexes again through midstance. During the push-off phase the hindfoot plantarflexes, inverts and internally rotates (Carson et al., 2001; Leardini et al., 2007; Levinger et al., 2010; Stebbins et al., 2006).

The forefoot/hindfoot joint shows the movement of the forefoot with respect to the hindfoot. The movements that can occur at this joint are the plantar/dorsiflexion about the mediolateral axis of the hindfoot, supination/pronation about the posterior/anterior axis of the forefoot, and abduction/adduction about the common perpendicular axis. The forefoot slightly dorsiflexes in midstance as the longitudinal arch flattens. From foot flat to heel off, the ankle becomes the rocker. The foot is stationary, and the tibia advances due to passive ankle dorsiflexion. In terminal stance, the longitudinal arch restores its shape as the forefoot plantarflexes, supinates, and adducts. The forefoot motion changes from pronation to supination around 30% of the stance phase (Carson et al., 2001; Leardini et al., 2007; Lee et al., 2022; Levinger et al., 2010; Perry and Burnfield, 2010; Stebbins et al., 2006).

The hallux/forefoot joint shows the movement of the hallux with respect to the forefoot. The movements that can occur at this joint are plantar/dorsiflexion about the mediolateral axis of the forefoot. As the heel comes off the ground at the end of midstance and the foot rocker moves to the hallux, the hallux dorsiflexes. During pre-swing, the hallux is the base of support for accelerated limb advancement (Carson et al., 2001; Levinger et al., 2010; Perry and Burnfield, 2010; Stebbins et al., 2006).

Effect of walking speed on foot kinematics

Walking is an important aspect to all activities of daily living, and it is important to maintain walking abilities in all individuals, especially those who are aging or have disabilities (Perry and Burnfield, 2010; Simonsick et al., 2005). Walking speed can be measured as a key indicator of function and is a good indicator of disability outcomes (Guralnik et al., 2000; Perry and Burnfield, 2010; Simonsick et al., 2005).

As walking speed increases, the foot motion also changes to support the increased demand on the body. Increased walking speed increases step length, cadence, and duration of swing phase (Buddhadev and Martin, 2016; Teixeira-Salmela et al., 2008). Walking speed, stride length, and cadence are important gait parameters that influence kinematics of the lower extremity and foot segments during walking. A self-selected walking speed is commonly used to determine normal kinematic measures during walking. However, self-selected walking speed may differ between groups.

Normal self-selected walking speed for men ages 20 to 59 years old is 1.34-1.43 m/s while for women ages 20 to 59 years old it is 1.27-1.40 m/s (Bohannon, 1997; Kadaba et al., 1990; Perry & Burnfield, 2010; Waters et al., 1988). Older adults tend to have a decreased walking speed compared to young adults (Buddhadev and Martin, 2016; Kang and Dingwell, 2008). For older women ages 60 to 80 years old, the normal self-selected walking speed is 1.20 m/s. For older men ages 60 to 80 years, the normal self-selected walking speed is 1.23 m/s (Perry and Burnfield, 2010; Waters et al., 1988).

The effect of walking speed on specific changes in kinematics of hindfoot, forefoot, and hallux motion – defined using the Oxford Foot Model – will now be explained. During fast walking, the hindfoot has increased peak dorsiflexion and dorsiflexion/plantarflexion range of motion. In the stance phase, there was a significant increase in eversion at heel contact and decrease in peak inversion. There was greater peak external rotation and range of motion during the gait cycle. (Sun et al., 2018). In the loading phase of the gait cycle, hindfoot abduction/adduction, dorsiflexion/plantarflexion, and inversion/eversion ranges of motion are higher with increased speed. In the hindfoot during propulsion, range of motion in the frontal and transverse planes were higher with increased speed (van Hoeve et al., 2017). If the hindfoot has greater range of motion in the sagittal plane, the heel rocker mechanism may have a greater role in the progression of the body during walking.

The forefoot had a greater peak dorsiflexion during fast walking compared to slow walking (Dubbeldam et al., 2010; Tulchin et al., 2009). Dorsiflexion/plantarflexion range of motion was increased with fast walking (Dubbeldam et al., 2010). There was an increase in supination with decreased speed. Decreased speed increases adduction angles at initial contact and adduction/abduction range of motion throughout the entire gait cycle (Sun et al., 2018; van Hoeve et al., 2017). During fast walking the hallux had decreased dorsiflexion/plantarflexion range of motion and dorsiflexion peak compared to normal and slow walking (Sun et al., 2018). The timing of maximal hallux dorsiflexion happened earlier during the gait cycle with an increased walking speed (Tulchin et al., 2009). This indicates the forefoot rocker mechanism has an important role in foot motion with an increase in walking speed.

The effect of walking speed on specific changes in kinematics of hindfoot, forefoot, and hallux motion has been defined using other multi-segment foot models. Dubbeldam et al. (2010) used a 19-marker model to assess how walking speed affected foot motion. They found that in the sagittal plane a lower walking speed correlated with a reduction of the peak plantarflexion at the talocrural joint, a flatter medial arch, and decreased dorsiflexion of the hallux at toe-off. Their findings with respect to peak hallux dorsiflexion angle at different walking speeds contrast with the finding of Sun et al. (2018). While Sun et al. (2018) found a decrease in hallux dorsiflexion/plantarflexion ROM and dorsiflexion peak hallux angle with an increase in walking speed, Dubbledam et al. (2010) found an increase in peak hallux dorsiflexion angle with an increase in walking speed. In the frontal plane, there was an increase in pronation of the midfoot during midstance during a lower walking speed. ROM of the talocrural joint, the medial arch and hallux joint in the sagittal plane was less during lower walking speeds at toe-off. In the transverse plane, there was less abduction of the hallux during a lower walking speed (Dubbeldam et al., 2010).

A 34-marker model was used by Grant and Chester (2015) to assess foot motion during five walking gait speeds ranging from very slow to very fast. As walking speed increased, the mean maximum dorsiflexion angle between the foot and the tibia decreased. The mean maximum midfoot-calcaneus plantar flexion and the forefoot with respect to the calcaneus angles increased with increased speed. The maximum absolute sagittal plane angle for the hallux increased with increased gait speed. The temporal-spatial variables significantly changed with walking speed changes. As speed increased, time to toe-off decreased, and single leg stance time and stride length increased (Grant and Chester, 2015).

Grant and Chester (2015) and Tulchin et al. (2009) reported an increase in maximum forefoot-calcaneus plantar flexion angle as walking speed increased, although the values of the increase differed. The difference in the angles reported may have been contributed to a different in the anatomical landmarks used to define the forefoot segment between models. Tulchin et al. also used a treadmill to control speed while Grant & Chester used overground walking at different speeds to assess change in foot motion (Grant and Chester, 2015; Tulchin et al., 2009).

Overall, the foot mechanics during walking at faster speeds involve a combination of joint angle change and quicker movements to support the body and maintain balance.

Effect of Step Length on Foot Kinematics

This section will explain the importance of manipulating step length for gait analysis. With increased walking speed there is an increase in step length (Buddhadev et al., 2020; Buddhadev and Martin, 2016; Teixeira-Salmela et al., 2008). Normal self-selected stride length and cadence for men ages 20 to 59 years old is 1.41-1.51m and 108-112 steps/min, respectively. Normal self-selected stride length and cadence for women ages 20 to 59 years old is 1.30-1.32m and 115-118 steps/min, respectively (Kadaba et al., 1990; Perry and Burnfield, 2010; Waters et al., 1988). Stride length and cadence tend to be decreased in older adults (Buddhadev and Martin, 2016; Kang and Dingwell, 2008). Normal stride length and cadence for older men are 1.45m and 105.85 steps/min, respectively. Normal stride length and cadence for older women are 1.27m and 112.85 steps/min, respectively (Kadaba et al., 1990; Waters et al., 1988).

When speed is held constant and step length is increased, there is an increase in net metabolic rate (Umberger and Martin, 2007), anteroposterior braking and propulsive forces, contact time (Martin and Marsh, 1992), and net joint moment impulses at the ankle (Allet et al., 2011). It is known how the foot motion changes with increased speed. However, there is no research on how the manipulation of step length, when controlling for speed, effects changes in kinematics of forefoot, midfoot, and hindfoot motion during walking. Using to the knowledge of foot motion with manipulation of speed using the Oxford Foot Model, the changes in motion of the forefoot, midfoot, and hindfoot segments at different step lengths could be hypothesized. It is expected that with an increase of step length there is an increase in range of motion of the hindfoot and forefoot segments, but a decrease in range of motion of the hallux segment. There may be increased dorsiflexion and plantarflexion in the hindfoot and forefoot segments, but a decrease in the hallux segment.

Summary

From the literature, it can be concluded the conventional foot model is not appropriate to investigate the internal foot motion during walking. Instead, the Oxford Foot Model provides a more anatomically accurate model as it breaks the foot into three segments: hindfoot, forefoot, and hallux. Past research has examined the effects of walking speed on specific kinematics of hindfoot, forefoot, and hallux motion using the Oxford Foot Model. However, there is no research examining the effects of step length on specific changes in kinematics of the hindfoot, forefoot, and hallux. From this review, it is apparent that there is a need for research examining internal foot motion associated with foot rocker mechanisms when step length is manipulated while walking speed is held constant.

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Appendix A: Journal of Biomechanics Author Guidelines

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3. All contributors who do not meet the criteria for authorship as defined above should be listed in an acknowledgements section. Examples of those who might be acknowledged include a person who provided purely technical help, writing assistance, or a department chair who provided only general support. Authors should disclose whether they had any writing assistance and identify the entity that paid for this assistance.
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7. Authors are encouraged to suggest referees although the choice is left to the editors. If you do, please supply the address and the email address, if known to you. Please do not include those with whom you have had active collaboration within the past 3 years.
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Work on human beings that is submitted to the Journal of Biomechanics should comply with the principles laid down in the Declaration of Helsinki; Recommendations guiding physicians in biomedical research involving human subjects. Adopted by the 18th World Medical Assembly, Helsinki, Finland, June 1964, amended by the 29th World Medical Assembly, Tokyo, Japan, October 1975, the 35th World Medical Assembly, Venice, Italy, October 1983, and the 41st World Medical Assembly, Hong Kong, September 1989. The manuscript should contain a statement that the work has been approved by the appropriate ethical committees related to the institution(s) in which it was performed and that subjects gave informed consent to the work. Studies involving experiments with animals must state that their care was in accordance with institution guidelines. Patients' and volunteers' names, initials, and hospital numbers should not be used.

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van Werff, K., 1977. Kinematic and dynamic analysis of mechanisms. A finite element approach. PhD. thesis, Delft University Press, Delft.
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van Soest, A. J., van den Bogert, A. J., 1991. Criteria for the comparison of direct dynamics software systems to be used in the field of biomechanics. In Proceedings of the 3rd International Symposium on Computer Simulation in Biomechanics. University of Western Australia, Perth.
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Appendix B: WWU Informed Consent

Informed Consent Form

Introduction

My name is Harsh Buddhadev, and I am a faculty member at Western Washington University (WWU). I also hold a role as a professor.

My name is Emily Lovekin, and I am a student at WWU. I also hold a role as a Graduate Teaching Assistant.

We are conducting a research study to determine effect of step length on 3D foot motion during walking. The name of this research study is "Foot motion during walking". I am seeking your consent to participate in this study.

Please read this document to learn more about this study and determine if you would like to participate. Your participation is completely voluntary, and I will address your questions or concerns at any point before or during the study.

Eligibility

You may participate in this research if you meet all of the following criteria:

1. You are a healthy person between 18 and 30 years who can walk for more than 10 minutes without needing to rest.
2. You do not have flat feet, or any other conditions, injuries, or surgeries that affect your foot or walking.

I hope to include 100 people in this research.

Activities

If you decide to participate in this study, you will be asked to do the following activities:

1. Complete a 1.5-hour testing session in the Biomechanics lab located at Carver 305

2. Review and sign this informed consent document. Complete two short questionnaires and then your arch height will be measured in sitting and standing for both feet. This will take about 10-minutes.

3. Change into spandex tank tops and shorts provided by the researchers. Then your height, weight, leg length, and hip, knee and ankle width will be measured over 10 minutes.

4. 43 markers will be applied on your skin and clothes using a double-sided tape on your right and left feet, lower leg, thigh, and hips. Then you will stand barefoot with hands

across your chest for 5 seconds as we collect data about the positions of the markers using a motion capture system. This process will last 10 minutes.

5. You will then walk barefoot at your preferred speed or a fixed speed of 2.9 miles per hour (mph) five times across a 12-meter distance. If you are assigned walking speed of 2.9 mph, then your speed will be checked after each trial to ensure it is within a range of 2.8-3.0 mph. If the speed is not in this range, then you will repeat the trial until 5 trials within this range are collected. This process will take about 10 minutes.

6. You will then walk barefoot over the 12-meter distance 5 times under these 3 conditions in a random order: 1) walking at the assigned speed at your average preferred step length, 2) walking at the assigned speed with the average step length shortened by 10% of your leg length, and 3) walking at the assigned speed with the average step length lengthened by 10% of your leg length. As you do these walking trials, we will measure the position of the markers using a motion capture system and forces applied by your foot on the ground using force plates located midway along the 12-meter distance.

7. For each condition, you will walk barefoot at the beat of a metronome and step on tape marks placed on the floor. You will be given a few practice trials to get used to walking at the beat of a metronome while contacting the tape marks using the natural heel to toe walking pattern. After each trial we will check if you were within 3% range of the assigned speed, if not then trials will be repeated until 5 trials are within this speed range. This process will take 45 minutes, about 15 minutes per condition.
8. The researchers will then remove the markers placed on your body and then you will change back into your own clothes and return the spandex clothes to the researchers. This will take 5 minutes.

During these activities, you will be asked questions about:

- Your age and gender

- Whether you have or had any injury, surgery, or condition that affects your foot or walking.

- Using a survey, we will ask questions about foot and ankle related difficulty you experienced in the past week doing daily activities such as standing, walking, going up stairs, etc. We will also ask you to share your level of foot and ankle pain in general, at rest, during normal activity, and first thing in the morning.

All activities and questions are optional: you may skip any part of this study that you do not wish to complete and may stop at any time.

Risks

There are no foreseeable risks or discomforts associated with this study. You can still skip any question you do not wish to answer, skip any activity, or stop participation at any time.

Benefits

If you participate, there are no direct benefits to you. This research may increase the body of knowledge in the subject area of this study.

Privacy and Data Protection

We will take reasonable measures to protect the security of all your personal information, but we cannot guarantee confidentiality of your research data. In addition to us, the following people and offices will have access to your data:

- The WWU Institutional Review Board

We will securely store your data for 3 years.

How the Results Will Be Used

We will publish the results in Emily Lovekin's thesis, and in Harsh Buddhadev's and Emily Lovekin's research presentations and publications. Participants will not be identified in the results.

Contact Information

If you have questions, you can contact me at: lovekie@wwu.edu.

My Faculty Advisor's name is Dr. Harsh Buddhadev. He works at Western Washington University and are supervising me on the research. You can contact him at: buddhah@wwu.edu.

If you have questions about your rights in the research or if a problem or injury has occurred during your participation, please contact the WWU Institutional Review Board at compliance@wwu.edu or 360-650-3220.

Voluntary Participation

If you decide not to participate, or if you stop participation after you start, there will be no penalty to you: you will not lose any benefit to which you are otherwise entitled.

Appendix C: The Foot & Ankle Disability Index (FADI) Score

The Foot & Ankle Disability Index (FADI) Score

Clinician's name (or ref) ------------------------------ **Patient's name (or ref** ------------------------------

Please answer every question with one response that most closely describes your condition within the past week. If the activity in question is limited by something other than your foot or ankle, mark N/A

	No difficulty at all	Slight difficulty	Moderate difficulty	Extreme difficulty	Unable to do
1. Standing	○	○	○	○	○
2. Walking on even ground	○	○	○	○	○
3. Walking on even ground without shoes	○	○	○	○	○
4. Walking up hills	○	○	○	○	○
5. Walking down hills	○	○	○	○	○
6. Going up stairs	○	○	○	○	○
7. Going down stairs	○	○	○	○	○
8. Walking on uneven ground	○	○	○	○	○
9. Stepping up and down curves	○	○	○	○	○
10. Squatting	○	○	○	○	○
11. Sleeping	○	○	○	○	○
12. Coming up to your toes	○	○	○	○	○
13. Walking initially	○	○	○	○	○
14. Walking 5 minutes or less	○	○	○	○	○
15. Walking approximately 10 minutes	○	○	○	○	○
16. Walking 15 minutes or greater	○	○	○	○	○
17. Home responsibilities	○	○	○	○	○
18. Activities of daily living	○	○	○	○	○
19. Personal care	○	○	○	○	○
20. Light to moderate work (standing, walking)	○	○	○	○	○
21. Heavy work (push/pulling, climbing, carrying)	○	○	○	○	○
22. Recreational activities	○	○	○	○	○

	NO PAIN	MILD	MODERATE	SEVERE	UNBEARABLE
23. General level of pain	○	○	○	○	○
24. Pain at rest	○	○	○	○	○
25. Pain during your normal activity	○	○	○	○	○
26. Pain first thing in the morning	○	○	○	○	○

Thank you very much for completing all the questions in this questionnaire.

The Foot & Ankle Disability Index (FADI) Score is 0

Appendix: D: Personal Health Questionnaire

PERSONAL HEALTH HISTORY

Please complete this as accurately and completely as possible.

CIRCLE ONE

Are you able to walk continuously at a comfortable walking speed for
10 minutes without major discomfort or need to rest? ……………………………. YES / NO
*If **NO**, do not proceed further with the questionnaire.*

Do you have flat feet?. …………………… ……………………………………...YES / NO
*If **YES**, do not proceed further with the questionnaire.*

Have you ever had any surgery on your feet?. …………………… ………………..YES / NO
*If **YES**, do not proceed further with the questionnaire.*

Have you recently (i.e. in the past year) had any injury to your feet?.. ……..………YES / NO
*If **YES**, do not proceed further with the questionnaire.*

Do you have any other health issues (e.g., pain, numbness, etc.) that
restrict or prevent normal walking ability?..YES / NO
*If **YES**, do not proceed further with the questionnaire.*

Appendix E: Participant Recruitment Flyer

Participants Needed!

Research: To Determine Foot Motion During Walking

What will you do?

- ❖ 1 Testing session lasting 1.5 hours
- ❖ You will stand and walk barefoot during the session

Who can participate?

- ❖ Adults ages 18-30 years
- ❖ Able to walk continuously for +10 minutes
- ❖ No foot/ankle surgeries or injuries
- ❖ No difficulty walking
- ❖ No flat feet

Please email lovekie@wwu.edu to find out if you are eligible!