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Multi-joint foot kinetics during walking in people with Diabetes Mellitus and peripheral neuropathy

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ABSTRACT

Neuropathic tissue changes can alter muscle function and are a primary reason for foot pathologies in people with Diabetes Mellitus and peripheral neuropathy (DMPN). Understanding of foot kinetics in people with DMPN is derived from single-segment foot modeling approaches. This approach, however, does not provide insight into midfoot power and work. Gaining an understanding of midfoot kinetics in people with DMPN prior to deformity or ulceration may help link foot biomechanics to anticipated pathologies in the midfoot and forefoot. The purpose of this study was to evaluate midfoot (MF) and rearfoot (RF) power and work in people with DMPN and a healthy matched control group. Thirty people participated (15 DMPN and 15 Controls). An electro-magnetic tracking system and force plate were used to record multi-segment foot kinematics and ground reaction forces during walking. MF and RF power, work, and negative work ratios were calculated and compared between groups. Findings demonstrated that the DMPN group had greater negative peak power and reduced positive peak power at the MF and RF (all $p \leq 0.05$). DMPN group negative work ratios were also greater at the MF and RF [Mean difference MF: 9.9%; $p = 0.24$ and RF: 18.8%; $p < 0.01$]. In people with DMPN, the greater proportion of negative work may negatively affect foot structures during forward propulsion, when positive work and foot stability should predominate. Further study is recommended to determine how both MF and RF kinetics influence the development of deformity and ulceration in people with DMPN.

1. Introduction

As many as 50% of people with Diabetes Mellitus (DM) will develop peripheral neuropathy (PN) (Gordois et al., 2003). In the foot, the hallmark signs of DMPN are loss of protective sensation, decreased non-contractile tissue extensibility, and intrinsic muscle atrophy and fatty infiltration (Brownlee, 1992; Cheuy et al., 2013; Pham et al., 2000). The foot-specific effects of PN are principle factors in the development of deformity, elevated plantar pressures and the increased risk for plantar ulceration (Cheuy et al., 2013; Crawford et al., 2007; Mueller et al., 2003).

The loss of intrinsic foot muscle function in people with DMPN is of particular concern because muscle atrophy is one of the earliest detectable precursors of abnormal foot function leading to pathology (Greenman et al., 2005). In healthy adults, contraction of intrinsic foot muscles attenuates, and can reverse, longitudinal arch deformation under increasingly loaded conditions (Kelly et al., 2014). Degradation of intrinsic muscle function in people with DMPN may impair the ability of the midfoot to produce positive work and attain a rigid foot posture during the push off phase of gait. Interestingly, an investigation of people with DMPN and medial column deformity demonstrated decreased forefoot plantarflexion (relative to rearfoot) (i.e. arch deformation) during single-leg heel rise tasks in comparison to healthy controls (Hastings et al., 2014). This kinematic finding suggests midfoot power produced by the interaction of the muscles and ligaments supporting the medial longitudinal arch is decreased, which may contribute to deformity. Yet, specific knowledge of diabetic foot kinetics (i.e. midfoot moments and power), during a common task like walking and prior to the onset of deformity, is limited.

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Understanding of in-vivo foot kinetics in people with DMPN is derived from investigations utilizing single-segment foot modeling approaches. These studies demonstrate that people with DMPN have reduced peak ankle power generation during gait (Mueller et al., 1994; Rao et al., 2006, 2010; Yavuzer et al., 2006). Multi-segment foot modeling studies that assess both midfoot (MF) and rearfoot (RF) power have only been performed in the healthy adolescent population. These studies demonstrate that MF power contributes to forward propulsion during gait and that single-segment foot modeling overestimates RF power generation (Dixon et al., 2012; MacWilliams et al., 2003). Multi-joint foot modeling offers specific insight into muscle performance at the MF and a more accurate representation of RF function. The assessment of multi-joint foot kinetics is an advancement of single segment modeling approaches and a necessary next step when evaluating pathology at the forefoot and MF. Yet, an investigation of multi-joint power and work has not been performed in people with DMPN or healthy adults.

It is hypothesized that both RF and MF power generation are deficient during gait in people with DMPN. A reduction of MF power generation or positive work, as well as a greater amount of power absorption, would indicate less active muscle support of the midfoot, and potentially greater loading on passive structures. If supported, this finding in patients prior to deformity would link anticipated foot muscle and ligament changes to foot biomechanics during walking. Pathologies such as toe/midfoot deformity and forefoot tissue breakdown that are catalyzed by neuropathy may be expedited by the repetition of abnormal MF function during daily weight-bearing activity.

The purpose of this study was to compare the multi-joint kinetic profile of people with DMPN without deformity or ulceration, to healthy matched controls during walking. It was anticipated that people with DMPN would demonstrate 1) decreased MF positive work, 2) increased MF and RF negative peak power and 3) an increased negative work ratio at both the MF and RF. Detection of an abnormal kinetic pattern of MF function prior to the development of pathology may guide formulation of early foot-specific interventions aimed at reducing the progression of deformity and disability.

2. Methods

2.1. Subjects

Thirty subjects, 15 people with DMPN and 15 healthy matched controls (age, gender, BMI), participated in this case control study (Table 1). Sample size was determined from kinetic data of a pilot study of people with DMPN and controls (N=6) [α=0.05; 1−β=0.8; Cohen's d range 1.0–5.2] and is consistent with prior investigations of DMPN foot function (Rao et al., 2007).

Table 1
Subject characteristics. Means and standard deviations (SD), as well as p-values of group comparisons, for the matching variables of age, gender, and body mass index are displayed. Additional information, specific to Diabetes Mellitus, is provided for the DMPN group.

	DMPN (n=15) mean (SD)	Control (n=15) mean (SD)	p-Value
Age (years)	57.4 (9.9)	55.7 (10.2)	0.64
Gender (% male)	80%	80%	1.0
BMI (kg/m^2)	30.9 (5.7)	31.9 (5.8)	0.67
Type of DM (% type II)	73.3%		
Duration of DM (years)	19.6 (12.4)		
HbA1c	7.6 (1.3)		

DMPN=Diabetes Mellitus and peripheral neuropathy; BMI=Body Mass Index; DM=Diabetes Mellitus; HbA1c=Glycated Hemoglobin Test.

Subjects in the DMPN group were recruited from a university health system. Review of medical records confirmed a history of DM without ulcer history (fasting blood glucose 126 ≥ mg/dl, HbA1c ≥ 6.5%) (American Diabetes Association, 2014). A clinical exam determined the absence of an active ulcer or rigid foot deformities. Positive findings on 2/3 clinical tests for loss of protective sensation (unable to detect a Semmes Weinstein 5.07 monofilament at 1+ location on the plantar foot, vibration from a 128 Hz tuning fork at the dorsal hallux, pin prick at 1+ location at the dorsal hallux/plantar foot) confirmed PN (Boulton et al., 2008). The foot with the greatest loss of protective sensation was the test foot. Convenience sampling was used to recruit and match control subjects to the DMPN group. Control subjects did not have current foot/ankle pathology or surgical history. The test foot was determined by a coin flip. The University of Rochester and Ithaca College Human Subjects Review Boards approved procedures and subjects provided written informed consent. Data were collected at the former Ithaca College Movement Analysis Laboratory and Center for Foot and Ankle Research in Rochester, NY.

2.2. Data acquisition

A second-step protocol was used to collect data as subjects walked barefoot over a force plate embedded in the level 25 ft walkway floor for a minimum of three trials (Nawoczenski et al., 1999; Rao et al., 2009). Similar to prior DMPN foot function studies (Rao et al., 2006, 2007), speed traps (Brower Timing Systems, Knoxville, TN) were used to control speed at 0.9 m/s (±5%) because walking speed influences joint power (Chen et al., 1997). The Flock of Bird™ (MiniBird™ hardware) six degree of freedom electromagnetic sensor motion capture system (Ascension Technology Corporation, Burlington, VT) was used to record multi-segment kinematics (100 Hz). A force plate (4060-NC; Bertec Corporation, Columbus, OH) measured ground reaction force data (1000 Hz) and was used to demarcate stance phase (±20 N). Kinematic and force plate data were smoothed using a 4th order, zero phase lag Butterworth filter with a cut off frequency of 6 and 50 Hz, respectively. MotionMonitor™ software (Innovative Sport Training, Chicago, IL) was used to synchronize and integrate data.

2.2.1. Kinetic model

Skin surface markers were attached with adhesive tape to skin overlying the 1st, 3rd and 5th metatarsals, middle cuneiform, calcaneus and tibia of one foot of each participant (Fig. 1a). Anatomical landmarks were digitized and subject-specific models were developed via transformation of the technical coordinate system of the sensors into local, anatomically-based, coordinate systems for each segment. A unified forefoot segment was created from anatomical landmarks corresponding to the sensor data of metatarsals 1, 3 and 5 (Supplemental file 1) (Fig. 1b). The coordinate system of each anatomical segment was oriented/rotated so that the y-axis was vertical (superior positive), x-axis was anterior/posterior (anterior positive), and the z-axis was medial/lateral (positive toward subjects right and parallel to the floor). A similar multi-segment foot model with the same approach has been previously validated (Umberger et al., 1999).

The three segment model used for kinetic analysis consisted of the tibia, rearfoot (calcaneus) and unified forefoot (Fig. 1c). The midpoint between the medial and lateral malleoli was used as the ankle (RF) center of rotation. The MF center of rotation was located at the center of the middle cuneiform and offset 11.3 mm inferiorly to place the center of rotation within the foot. The magnitude of this offset was based on anthropomorphic data (Harris and Case, 2012).

Prior multi-segment approaches have demonstrated value in using a MF joint center(s) to quantify forefoot to rearfoot sagittal plane motion during forward progression of gait (Dixon et al., 2012; Leardini et al., 2007; MacWilliams et al., 2003). Yet, the location of a single MF joint center is controversial because motion occurs at multiple joints in the MF (Nester et al., 2007) and the placement will influence measurement of MF kinetics. Our placement within the MF and at the middle cuneiform estimates a short forefoot moment arm and therefore a lower MF moment and consequent power. Alternatively, a more proximal placement of the MF joint center would increase the midfoot moment and consequent power. Dixon et al. (2012) used a three segment foot model to measure MF kinetics but placed the MF center of rotation on the surface of the transverse tarsometatarsal joint. We have also selected a relatively distal MF joint center that allows for inference of muscle and ligament action about the MF in the sagittal plane.

2.2.2. Calculations

Relaxed standing, with the feet oriented anterior–posterior, was used as a zero reference point for kinematic data analysis and walking trial data were normalized across stance phase (0–100%) (Leardini et al., 1999). Stance time (s) was acquired from force plate data and used in the calculation of work. The relative angular displacements and velocities of each segment were calculated using an Euler rotation sequence of Z, X', Y'' (Rao et al., 2007). Specifically, forefoot with respect to rearfoot and rearfoot with respect to tibia velocities were used in MF and RF power calculations.

All inverse dynamic calculations were conducted following the time point when the anterior–posterior position of the center of pressure was equal to the midfoot center of rotation (i.e. after heel off). There were minimal between-group differences in time to heel off (as % of stance) [Mean (SD): DMPN 54.5 (12.0) vs.