ARTICLE IN PRESS

Journal of Biomechanics ∎ (∎∎∎∎) ∎∎∎–∎∎∎

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Leg joint function during walking acceleration and deceleration

Mu Qiao [a], Devin L. Jindrich [b,*]

[a] Kinesiology Program, School of Nutrition and Health Promotion, Arizona State University, Phoenix, AZ 85004-0698, United States
[b] Department of Kinesiology, California State University, San Marcos, CA 92096 United States

ARTICLE INFO

Article history:
Accepted 15 November 2015

Keywords:
Locomotion
Maneuver
Gait
Initiation
Termination

ABSTRACT

Although constant-average-velocity walking has been extensively studied, less is known about walking maneuvers that change speed. We investigated the function of individual leg joints when humans walked at a constant speed, accelerated or decelerated. We hypothesized that leg joints make different functional contributions to maneuvers. Specifically, we hypothesized that the hip generates positive mechanical work (acting like a "motor"), the knee generates little mechanical work (acting like a "strut"), and the ankle absorbs energy during the first half of stance and generates energy during the second half (consistent with "spring"-like function). We recorded full body kinematics and kinetics, used inverse dynamics to estimate net joint moments, and decomposed joint function into strut-, motor-, damper-, and spring-like components using indices based on net joint work. Although overall leg mechanics were primarily strut-like, individual joints did not act as struts during stance. The hip functioned as a power generating "motor," and ankle function was consistent with spring-like behavior. Even though net knee work was small, the knee did not behave solely as a strut but also showed motor-, and damper-like function. Acceleration involved increased motor-like function of the hip and ankle. Deceleration involved decreased hip motor-like function and ankle spring-like function and increased damping at the knee and ankle. Changes to joint mechanical work were primarily due to changes in joint angular displacements and not net moments. Overall, joints maintain different functional roles during unsteady locomotion.

1. Introduction

Activities of daily living require obstacle avoidance, changing speed, turning, and maintaining balance after perturbations (Glaister et al., 2007; Jindrich and Qiao, 2009). This "unsteady" locomotion may require changes to leg and joint function, and may reduce energy exchange that could reduce locomotion economy (Cavagna and Kaneko, 1977; Rome et al., 2005). However, the specific biomechanical and motor control mechanisms used to maintain stability or maneuver are often not well understood (Qiao and Jindrich, 2014).

One important question is how joint mechanics are coordinated to achieve desired movements of the entire body. Joints can be coordinated to achieve overall limb function necessary to task demands (Yen et al., 2009). However, joints may also exhibit different functions. For example, switching from walking to running involves power increases at distal more than proximal joints (Farris and Sawicki, 2012; Sawicki and Ferris, 2009).

In contrast, the hip may primarily contribute to uphill locomotion (Roberts and Belliveau, 2005). Proximal joints may act as power producing motors whereas distal joints act more as springs (Daley et al., 2007). However, downhill and uneven-terrain walking involve changes to both hip and knee joint moments, suggesting that different tasks may involve functional changes to different joints (Lay et al., 2006; Voloshina et al., 2013).

Several important questions remain for both constant-average-velocity (CAV) and unsteady walking. During walking, the stance leg acts as a stiff "strut," reducing overall work production by the leg (Cavagna et al., 1976). However, it is unclear whether individual leg joints also exhibit primarily strut-like function, or also act as "motors" (energy producers), "springs" (storing and releasing energy) or "dampers" (energy absorbers) (Dickinson et al., 2000). Moreover, unsteady locomotion often requires energy production or absorption beyond that observed during CAV locomotion (Jindrich and Qiao, 2009). Joints may all contribute to changing COM energy in similar ways, or joints may make different functional contributions (Lee et al., 2008).

Therefore, we seek to test the general hypothesis that joints make different functional contributions to unsteady locomotion. We chose to study CAV walking and changing speed because changing speed is a functionally important behavior that involves changes to COM energy. Moreover, to our knowledge the mechanical changes to joint function associated with acceleration and deceleration have not been characterized.

* Corresponding author. Tel.: +1 760 750 7334; fax: +1 760 750 3237.
E-mail address: djindrich@csusm.edu (D.L. Jindrich).

http://dx.doi.org/10.1016/j.jbiomech.2015.11.022
0021-9290/© 2015 Elsevier Ltd. All rights reserved.

Please cite this article as: Qiao, M., Jindrich, D.L., Leg joint function during walking acceleration and deceleration. Journal of Biomechanics (2015), http://dx.doi.org/10.1016/j.jbiomech.2015.11.022

We tested two specific hypotheses. First, we hypothesized that leg joints have distinct functional roles during CAV locomotion, acceleration and deceleration: the hip acts as a motor producing positive work, the knee acts as a strut with minimal work production and angular movement, and the ankle acts as a torsional spring with alternating work absorption and production (the potential for storage and return). Second, we tested the null hypothesis that acceleration and deceleration would involve changes to the magnitudes of moments or energies produced by joints, but not to joint functional roles (strut, motor, damper or spring).

2. Methods

2.1. Participants

Sixteen male college students (age 27 ± 4 years, mass (m) 70 ± 8 kg, and height (bh) 1.77 ± 0.07 m, mean ± s.d.) participated in the study. All participants were healthy without neuromuscular disorders or injury history and gave informed consent. The same participants also performed a second set of running tasks (Qiao and Jindrich, 2012).

2.2. Experimental set-up and data collection

A 3D motion capture system (VICON®, model 612, 10 cameras, Oxford Metrics, Oxford, UK) recorded full body kinematics from 39 reflective markers (Plug-In-Gait marker set, 14mm in diameter, 120 Hz). Three force platforms (FPs) (400×600 mm^2, FP4060-NC, Bertec Corp., Columbus, OH, USA) acquired ground reaction forces (GRFs) at 3000 Hz (Fig. 1). The platforms were ground-mounted and covered by a 2 mm thick rubber mat (Ironcompany, Lafayette, CA, USA) to prevent vision of the platforms. The mats caused negligible peak vertical force attenuation (1.5%) and cross-talk (1.0%) (Qiao and Jindrich, 2012). Touchdown (TD) and takeoff (TO) were determined as the instants when the vertical GRF increased or decreased continuously for 5 ms (Qiao et al., 2014). We analyzed the stance phase from TD to TO of the left foot on the middle platform (FP2), ensuring that GRFs from the right foot were also recorded during double support (Fig. 1B).

2.3. Experimental protocol

Before data collection, participants performed trials from different initial positions to determine starting locations likely to result in success. We asked participants to perform three speed-change tasks (ΔSPEED): constant-average-velocity walking (CAV), acceleration (ACC), and deceleration (DEC). During CAV we asked participants to walk at their preferred speed; during ACC we asked participants to first walk at CAV then speed up as much as possible at FP1; during DEC we asked participants to first walk at CAV then slow down at FP1 to stop one step after FP3 (Fig. 1B). We recorded five trials for each task and randomized the order of the three tasks.

2.4. Calculations

We expressed kinematics and performed inverse kinematics and dynamics with a 15-segment, 34-degree of freedom (DOF) human model (Qiao and Jindrich, 2012). The hip was modeled as a ball-and-socket joint, the knee and ankle as hinge joints (Fig. 1A), and the foot as a single rigid body.

To calculate inverse kinematics, we first filtered the Cartesian coordinate time series for each marker with a Woltring filter (mean square error of 8), then calculated joint angles at each time sample using inverse kinematics. The inverse kinematics algorithm iteratively searched for joint angles that minimized a cost function (the sum of squares of the differences between measured markers and markers calculated from joint angles) (Qiao and Jindrich, 2012). We determined anthropometric parameters for each participant by allometric scaling (Pandy and Andriacchi, 2010). We assumed that each participant had the same percentage of segment mass as the reference model (Huston and Passerello, 1982). We calculated the dimensions of each body segment in vertical, anterior–posterior (AP) and medio-lateral (ML) directions by multiplying the referent model's segment length in each direction by bh/bh_{model} and the segment dimension ratio r in the horizontal plane, using the equation $r^2 \cdot bh/bh_{model} = m/m_{model}$ (Qiao et al., 2014). For each joint, we calculated the angular "displacement" as the joint angle at TO – joint angle at TD.

For inverse dynamics, we filtered joint angle time series with a 4th-order zero-lag low-pass Butterworth digital filter at 11 Hz before using a 3-point difference method to calculate the 1st-order time derivatives (angular velocities) (Qiao and Jindrich, 2014). We then

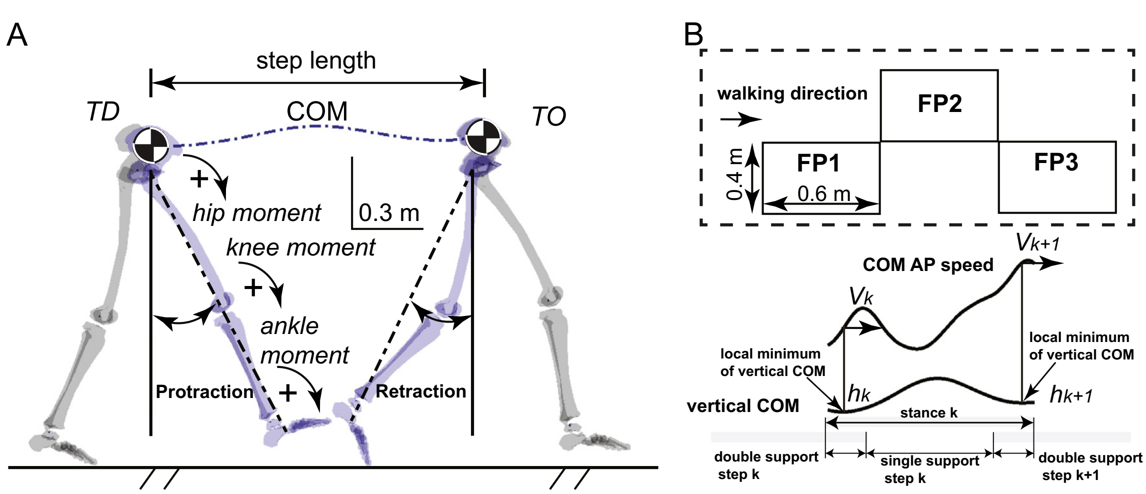

Fig. 1. Experiment setup. (A) Lower extremity postures at touchdown (TD) and takeoff (TO) of FP2. Dashed and dotted line is the COM trajectory. For the definition of joint moments in the lower extremities, hip extension(+)/flexion(−), knee flexion(+)/extension(−), ankle plantar-extension(+)/dorsiflexion(−). (B) Force platform arrangement. The dashed area was covered by a rubber mat. In FP2 left leg was in stance. v_k and v_{k+1} are the COM speeds at the local minimum of vertical COM during double supports k and $k+1$, where COM heights are h_k and h_{k+1} (Table 1, Fig. 4B).

Please cite this article as: Qiao, M., Jindrich, D.L., Leg joint function during walking acceleration and deceleration. Journal of Biomechanics (2015), http://dx.doi.org/10.1016/j.jbiomech.2015.11.022