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The functional roles of muscles during sloped walking

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ABSTRACT

Sloped walking is biomechanically different from level-ground walking, as evidenced by changes in joint kinematics and kinetics. However, the changes in muscle functional roles underlying these altered movement patterns have not been established. In this study, we developed a total of 273 muscle-actuated simulations to assess muscle functional roles, quantified by induced body center-of-mass accelerations and trunk and leg power, during walking on slopes of 0°, ±3°, ±6°, and ±9° at 1.25 m/s. The soleus and gastrocnemius both provided greater forward acceleration of the body parallel to the slope at +9° compared to level ground (+126% and +66%, respectively). However, while the power delivered to the trunk by the soleus varied with slope, the magnitude of net power delivered to the trunk and ipsilateral leg by the biarticular gastrocnemius was similar across all slopes. At +9°, the hip extensors absorbed more power from the trunk (230% hamstrings, 140% gluteus maximus) and generated more power to both legs (200% hamstrings, 160% gluteus maximus) compared to level ground. At −9°, the knee extensors (rectus femoris and vasti) accelerated the body upward perpendicular to the slope at least 50% more and backward parallel to the slope twice as much as on level ground. In addition, the knee extensors absorbed greater amounts of power from the ipsilateral leg on greater declines to control descent. Future studies can use these results to develop targeted rehabilitation programs and assistive devices aimed at restoring sloped walking ability in impaired populations.

1. Introduction

Sloped surfaces are encountered in both man-made and natural environments. When walking on slopes, the muscles raise or lower the body center-of-mass (COM) while maintaining balance. The altered biomechanical demands of sloped walking, particularly with large slope angles, require greater activity from lower-limb muscles, such as the gluteus maximus on inclines and rectus femoris and vasti on declines (Lay et al., 2007). On inclines, metabolic power is increased compared to level-ground walking (Jeffers et al., 2015). These results suggest that sloped walking is a more difficult biomechanical task than level-ground walking.

Previous sloped walking studies have established how joint kinematics and kinetics change compared to level-ground walking. For example, when walking on a 24% grade (13.5°) the hip and ankle ranges-of-motion increase by 20% and 59%, respectively, compared to level ground (Lange et al., 1996). In addition, when walking on a 39% grade (21.3°) the peak hip extension moment during early stance is nearly four times larger than on level ground and the peak ankle plantarflexion moment during late stance is 19% greater than on level ground (Lay et al., 2006). On declines, the knee joint mechanical power can be up to six times larger than on level ground (Kuster et al., 1995). However, the contributions of individual muscles to these altered biomechanics have not been established.

Musculoskeletal modeling and simulation allow for investigation of the functional roles of individual muscles during movement (Piazza, 2006). Muscles can accelerate all body segments (Zajac et al., 2002) and their functional roles can be difficult to predict based solely on anatomical classification (Hernández et al., 2008; Neptune et al., 2001). For example, the soleus generates power to the trunk while the gastrocnemius generates power to the ipsilateral leg for swing initiation during level-ground walking (Neptune et al., 2001). The hamstrings absorb power from the trunk and generate power to the ipsilateral leg during level-ground walking (Neptune et al., 2004; Silverman and Neptune, 2012), and the knee extensors provide braking acceleration of the COM (Pandy et al., 2010) while absorbing power from the ipsilateral leg and generating power to the trunk (Neptune et al., 2004; Silverman and Neptune, 2012). While these

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results provide an understanding of level-ground walking, muscle functional roles likely change in response to the biomechanical demands of sloped walking.

Our goal was to characterize the functional roles of the major lower-limb muscle groups in unimpaired adults during sloped walking using musculoskeletal modeling and simulation. We quantified muscle functional roles with induced COM acceleration and mechanical power delivered to the trunk and legs, and refer to each muscle's contribution to the net COM acceleration and segment mechanical power. Based on previous kinematic, kinetic, and musculoskeletal simulation results, we hypothesized that on inclines the ankle plantarflexors and hip extensors would accelerate the COM and generate power to the trunk and ipsilateral leg to a greater extent than in level-ground walking. We also hypothesized that during decline walking the knee extensors would provide greater braking acceleration and absorb more power from the ipsilateral leg relative to level-ground walking.

2. Methods

2.1. Experimental data collection

Thirteen healthy adults (4 female/9 male, 67 ± 10 kg, 173 ± 9 cm, 28 ± 7 years) provided written informed consent to participate in the protocol approved by the Department of Veterans Affairs' Human Subjects Institutional Review Board. Participants walked at 1.25 m/s on an instrumented dual-belt treadmill (Bertec Corp., Columbus, OH) on slopes of 0°, ±3°, ±6°, and ±9° in randomized order while we measured bilateral ground reaction forces (GRFs, 1500 Hz), whole-body kinematics (100 Hz, Vicon Inc., Centennial, CO) and electromyographic (EMG) signals (1500 Hz, Noraxon Corp., Scottsdale, AZ) from eight muscles of each leg (Table 1). For each person (13 total participants), three gait cycles were analyzed for each slope (7 total slopes), for a total of 273 simulations.

2.2. Simulation development

Kinematic marker trajectories were low-pass filtered with a cutoff frequency of 6 Hz using a 4th-order bidirectional Butterworth filter in Visual3D (C-Motion, Inc., Germantown, MD). An inverse kinematics solution was computed using a least squares optimization approach (Lu and O'Connor, 1999), and the resulting joint angles were low-pass filtered with a 6 Hz cutoff frequency. Force data were also low-pass filtered with a 6 Hz cutoff to eliminate noise caused by treadmill vibrations (Antonsson and Mann, 1985; Kram et al., 1998; Riley et al., 2007) and maintain consistency between data types (Bisseling and Hof, 2006; Kristianslund et al., 2012). Musculoskeletal models were developed in OpenSim 3.1 (Delp et al., 2007) by scaling a generic model (Anderson and Pandy, 1999; Delp et al., 1990) with 21 degrees-of-freedom (DOF) and 92 Hill-type musculotendon actuators with force-length-velocity properties (Zajac, 1989). Scale factors for each segment were computed from a static trial in Visual3D. Passive structures were represented by torques applied to each rotational DOF as an exponential function of joint angle (Anderson, 1999; Davy and Audu, 1987). A residual reduction algorithm (RRA) was used to ensure dynamic consistency between the inverse kinematics solution, model and GRFs by adjusting the total model mass and torso COM location (Delp et al., 2007). After making model adjustments, we used a custom optimization algorithm to adjust the inverse kinematics solution and minimize a multi-objective cost function based on the root-mean-squared residual forces and kinematic tracking errors for each trial. We then used a computed muscle control (CMC) algorithm to determine muscle forces that reproduced the kinematic solution from RRA while minimizing the sum of squared muscle excitations. Measured EMG signal timing was used to constrain the minimum and maximum excitations of the corresponding muscles in the model (Table 1). The EMG signals were processed by subtracting the mean, rectifying the signal, and applying a bidirectional moving average filter with a window of 100 ms. For each muscle, the processed signal for each trial was normalized by the peak value across all trials. Simulated muscles were constrained to be "on" (excitation ≥ 0.5) if the normalized EMG was greater than 0.5 for more than 0.01 s, and "off" (excitation ≤ 0.1) if the normalized EMG was less than 0.05 for more than 0.01 s. Otherwise the minimum and maximum excitation bounds were 0.02 and 1.0, respectively.

Table 1
Muscle group definitions and abbreviations. Muscles for which EMG was measured experimentally are denoted by 'E', and muscles in the group which were constrained based on EMG are denoted by 'C'.

Abbreviation	Muscles in group
SOL	Soleus^{EC}
GAS	Lateral gastrocnemius^{EC}
	Medial gastrocnemius^{C}
TA	Tibialis anterior^{EC}
	Extensor digitorum longus
	Extensor hallucis longus
	Peroneus tertius
VAS	Vastus lateralis^{EC}
	Vastus medialis^{C}
	Vastus intermedius^{C}
RF	Rectus femoris^{EC}
HAMS	Biceps femoris long head^{EC}
	Gracilis
	Semimembranosus
	Semitendinosus
GMAX	Gluteus maximus (superior, middle, and inferior compartments)^{EC}
IL	Iliacus
	Psoas
GMED	Gluteus medius (anterior, middle, and posterior compartments)^{EC}
	Gluteus minimus (anterior, middle, and posterior compartments)
	Piriformis

2.3. Induced acceleration and segment power analyses

An induced acceleration analysis (IAA) was performed to determine muscle contributions to net COM acceleration. We used a "rolling without slipping" kinematic constraint between the foot and ground during stance (Hamner et al., 2010) and solved the equations of motion to compute accelerations due to each force acting on the model. Accelerations were reported in the directions parallel to the treadmill (braking [−]/propulsion [+]), perpendicular to the treadmill (normal [+]), and mediolateral (lateral [−]/medial [+]). The treadmill reference frame was chosen instead of the global reference frame because it describes the accelerations parallel and perpendicular to the direction of progression during sloped walking.

We calculated the instantaneous power delivered to the body segments by each force in the model (Fregly and Zajac, 1996). The power delivered to the trunk (pelvis and torso) and legs (toes, calcaneus, talus, tibia, and femur in each leg) was calculated by summing the power delivered to each segment. The power in each segment was normalized by that segment's mass. Because participants were unimpaired, we assumed symmetry between legs and reported left leg muscle contributions to induced COM acceleration and segment power.

2.4. Statistical comparisons

A linear mixed effects ANOVA (Pinheiro et al., 2015) with slope as a fixed effect and participant as a random effect was used to assess changes in muscle results across slopes. The mean induced COM acceleration and power delivered to the trunk and legs from each left leg muscle group (Table 1) during stance (defined as 0–60% of the gait cycle) were compared. When a significant slope effect was found, post-hoc comparisons were performed between each slope and level ground using least squares means and Dunnett's method for p-value adjustments (Lenth and Hervé, 2015).

3. Results

Overall, the experimental joint angles and net joint moments from RRA (Fig. 1) were consistent with the literature (Fradet et al., 2010; Lay et al., 2006; Silverman et al., 2012). The model changes were within acceptable bounds for measurement error in model mass properties, with an average mass change of 1.00 ± 0.76 kg and torso COM change of 6.84 ± 2.70 cm. The simulations had similar muscle excitations to the collected EMG signals (Fig. 2), low residuals (Supplementary Table 1), and low tracking errors (Supplementary Table 2).

3.1. Induced body COM acceleration

SOL and GAS had increased contributions to propulsion (positive acceleration parallel to the treadmill) on inclines relative to level ground, with a 126% increase in SOL ($p < 0.001$) and a 66% increase in GAS ($p < 0.001$) contributions at +9° compared to 0° (Fig. 3). RF and

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