ORIGINAL ARTICLE

The Influence of Mechanically and Physiologically Imposed Stiff-Knee Gait Patterns on the Energy Cost of Walking

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ABSTRACT. Lewek MD, Osborn AJ, Wutzke CJ. The influence of mechanically and physiologically imposed stiff-knee gait patterns on the energy cost of walking. Arch Phys Med Rehabil 2012;93:123-8.

Objective: To investigate the relative roles of mechanically imposed and physiologically imposed stiff-knee gait (SKG) patterns on energy cost.

Design: Repeated-measures, within-subjects design.

Setting: Research laboratory.

Participants: Individuals (N=20) without musculoskeletal, neuromuscular, or cardiorespiratory limitations.

Interventions: Participants walked on an instrumented treadmill at their self-selected overground gait speed for 3 randomly ordered conditions: (1) control, (2) mechanically imposed stiff-knee gait (SKG-M) using a lockable knee brace, and (3) physiologically imposed stiff-knee gait (SKG-P) using electrical stimulation to the quadriceps. Each condition was performed with 0% and 20% body weight support. Indirect calorimetry determined net metabolic power, and motion capture measured lower extremity joint kinematics and kinetics.

Main Outcome Measures: Net metabolic power, knee flexion angle, circumduction, hip hiking, and hip flexion and ankle plantarflexion moments.

Results: Participants walked at 1.25±.09m/s. Net metabolic power was significantly increased by 17% in SKG-M and 37% in SKG-P compared with control (mean increase: .66±.60W/kg for SKG-M; 1.39±.79W/kg for SKG-P; both $P<.001$). Furthermore, SKG-P required greater net metabolic power than SKG-M ($P<.001$). Simulated SKG was associated with increased circumduction and hip hiking. Despite no change in ankle plantarflexion moments ($P=.280$), the hip flexion moment was increased during SKG-P (.43±.15Nm/kg·m) compared with control (.31±.08Nm/kg·m; $P<.001$).

Conclusions: The increase in energy cost associated with simulated SKG was due in part to abnormal mechanical compensations, and in part to an increase in quadriceps activity. Understanding the mechanisms underlying the increase in quadriceps activity will enable a reduction in the energy cost of walking with SKG.

Key Words: Biomechanics; Gait; Oxygen consumption; Rehabilitation.

INDIVIDUALS WITH STROKE, cerebral palsy, and acquired brain injury often exhibit deficits in hip, knee, and ankle flexion during swing.[1-3] This movement pattern, termed "spastic paretic stiff knee gait pattern," is among the most common types of abnormal gait patterns, presenting in approximately 25% to 30% of individuals after stroke.[3,4] The presence of a stiff-knee gait (SKG) pattern is purported to require a greater energy cost,[5-7] with evidence from unimpaired subjects to support this contention.[8-10] If walking with SKG is energy inefficient, it would be prudent to determine the underlying cause of the increase in energy cost.

The abnormal mechanics associated with SKG are often assumed to contribute to the greater energy cost.[11] For instance, with the knee extended, the swinging limb has a greater moment of inertia,[7,12] requiring greater hip moments that can increase energy cost. In addition, concomitant compensatory limb movements (eg, hip circumduction, hip hiking, lateral trunk lean, and contralateral vaulting) are commonly adopted to overcome the increased length of the swinging limb and reduce the risk of tripping.[13] That a contralateral shoe lift, intended to counteract these compensatory movements, reduces the energy cost associated with simulated SKG[8] provides evidence that abnormal mechanics may impact the energy cost during walking.

The presence of abnormal compensatory movements requires additional hip or ankle muscle activity, or both, which can further elevate energy cost. More importantly, perhaps, from a muscle perspective is that the SKG pattern itself has been attributed to inappropriate muscle activity during late stance/early swing.[1,14-16] Inappropriate thigh (eg, quadriceps) and shank (eg, soleus) muscle activity during late stance is purported to reduce knee flexion velocity, subsequently limiting knee flexion during swing.[17] Inappropriate quadriceps activity is most commonly believed to cause SKG,[1,14-16] and thus may further elevate energy cost. Therefore, the elevated energy cost may be due to the presence of (1) abnormal compensatory limb mechanics (ie, mechanical cause) and/or (2) the under-

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List of Abbreviations

ANOVA	analysis of variance
BWS	body weight support
GRF	ground reaction force
NMP	net metabolic power
SKG	stiff-knee gait
SKG-M	mechanically imposed stiff-knee gait
SKG-P	physiologically imposed stiff-knee gait
$\dot{V}CO_2$	carbon dioxide production
$\dot{V}O_2$	oxygen consumption

lying increase in knee extensor muscle activity (ie, physiologic cause).

Reducing elevated energy costs during walking may be accomplished with the use of body weight support (BWS), which is used clinically to assist with locomotor training.[18,19] Greater amounts of BWS are known to reduce the energy costs associated with normal walking.[20,21] If BWS can reduce energy costs associated with SKG, then patients with SKG might have the ability to walk further, faster, or longer during training to enhance rehabilitation practice.

The purpose of this study was to determine the influence of both mechanically imposed and physiologically imposed SKG patterns on energy cost. Based on previous work,[8-10] it was hypothesized that walking with a simulated SKG pattern would increase energy costs. Furthermore, we hypothesized that energy cost would be even greater if the simulated SKG pattern was imposed with an increase in quadriceps activity, compared with mechanically imposed, without an increase in quadriceps activity. In addition, we hypothesized that the use of BWS would reduce metabolic costs while walking both with and without simulated SKG. Finally, a determination of the compensations created in response to the simulated SKG patterns will provide guidance for treatment planning.

METHODS

Participants

Twenty unimpaired individuals (10 males, 10 females; mean age ± SD, 21.7±1.4y; mean height ± SD, 1.74±.10m; mean weight ± SD, 70.5±15.6kg; 19 right-leg dominant) were recruited to undergo testing. Participants were not included if they were pregnant, had a history of ligament deficiency, cardiovascular disease, neurologic impairment, impaired balance or history of unexplained falls, or other orthopedic problems in the lower extremities or spine. All subjects gave informed consent that was approved by the Human Subjects Review Board of UNC-Chapel Hill before participation.

Data Collection

All subjects underwent a single gait analysis to evaluate metabolic, limb kinematic, kinetic, and spatiotemporal measures. Subjects walked on a dual-belt treadmill[a] with two 6-component forceplates to collect ground reaction force (GRF) data. The treadmill speed was set individually for each subject to match the subject's self-selected overground gait speed, which was calculated as the average speed from 3 passes across a 14-ft pressure mat.[b] While walking on the treadmill, all participants wore a safety harness,[c] which did not restrict lower extremity movements. The harness was attached overhead to a custom-designed unweighting system (similar to that used in the study by Donelan and Kram[22]).

Each subject participated in 3 walking conditions (control, mechanically imposed SKG [SKG-M], and physiologically imposed SKG [SKG-P]), which were each performed at 2 BWS conditions (0%BWS and 20%BWS). These 6 trials were block randomized by walking condition, such that both randomly ordered BWS conditions were completed for a particular walking condition before performing the next randomly selected walking condition. SKG-M was achieved with an adjustable knee brace[d] placed on the dominant leg. The leg brace was worn for all trials: fixed at full extension for SKG-M, but left unlocked to allow unrestricted sagittal plane knee motion during the control and SKG-P conditions. A Grass S48 electrical stimulator[e] applied trains of electrical pulses (75 pulses/s, 400-μs pulse duration, 33-pulse tetanic train) to the dominant quadriceps to achieve the SKG-P condition. Large pregelled, self-adhesive 3 × 5-inch electrodes[f] were applied superficial to the proximal vastus lateralis and the distal vastus medialis muscles. Voltage amplitude during walking was sufficient to fully extend the knee against gravity while sitting. During walking, the stimulation was applied during each step, beginning during late stance (ie, the peak of anteriorly directed GRF) and persisting for 400 to 450msec.

During all trials, the rates of oxygen consumption ($\dot{V}O_2$; mL· kg^{-1} · min^{-1}) and carbon dioxide production ($\dot{V}CO_2$) were collected with a portable metabolic cart.[g] Before each data collection, the system was calibrated using known concentrations of gas. In addition to the walking trials, all subjects began testing with 5 minutes of quiet standing without BWS to measure baseline energy cost.

While walking, an 8-camera passive motion analysis system,[h] sampling at 120Hz, tracked 14-mm retroreflective markers attached to the lower extremities and pelvis. Bilaterally, markers were placed on the iliac crests, greater trochanters, lateral and medial femoral condyles, lateral and medial malleoli, dorsal surface of the foot (over the second metatarsal head), the posterior heel counters of the shoe, and the first and fifth metatarsal heads. Rigid thermoplastic shells were attached to the posterior pelvis (3 markers), and bilateral thighs and shanks posterolaterally (4 markers each). Each shell was attached firmly with elastic bandages to reduce movement between the bone and marker. Importantly, the shells on the dominant limb were placed under the knee brace to appropriately track the motion of the thigh and shank within the brace. After a static standing calibration to locate joint centers with respect to each segment coordinate system, all joint markers were removed.

Metabolic data were collected continuously on a breath-by-breath basis as subjects walked on the treadmill for approximately 4 or 5 minutes, to achieve "steady state." Once at metabolic steady state, subjects continued to walk for an additional 30 seconds, while kinematic and kinetic data were recorded. Steady state was determined visually during testing and later confirmed with techniques described previously.[23,24] Between conditions, subjects remained standing until metabolic rates returned to resting values.

Data Management

Measurements of $\dot{V}O_2$ and $\dot{V}CO_2$ were averaged over the 30 seconds that coincided with motion capture collection (ie, after 4–5 min of walking) and used to calculate metabolic power.[25] Net metabolic power (NMP) was determined by subtracting the baseline standing metabolic measures from the walking data and normalizing to body mass. Data analysis software (Vicon Nexus)[h] was used to identify the locations of the markers in the lab coordinate system. The markers defined a 7-segment kinematic model for tracking the 3-dimensional motion of the pelvis and lower limb segments. All segment coordinate systems were defined with the positive x-axis to the right, positive y-axis facing anteriorly, and positive z-axis pointing superiorly. Visual3D software[i] estimated segment properties from measured anthropometric values.[26] All segments were modeled as a frustra of right cones, except for the pelvis, which was modeled as a cylinder.

Although kinematic and kinetic data were collected bilaterally, only data from the dominant side were analyzed. Marker trajectory and GRF data were filtered with 6- and 20-Hz low pass filters, respectively. Joint angles were calculated using Euler angles. Sagittal plane internal joint moments were calculated using an inverse dynamics approach and normalized to body mass and height. Limb circumduction was defined as the peak lateral excursion of the foot during swing relative to the