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Training conditions that best reproduce the joint powers of unsupported walking

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ABSTRACT

Objective: To identify the clinically relevant combinations of body weight support and speed that best reproduce the joint powers of unsupported walking.
Methods: Timing and magnitude of lower extremity joint powers were calculated for 8 neurologically intact volunteers (4M/4F) walking with 0%, 30% and 50% body weight support at three speeds (slow, comfortable, and fast). Lower extremity joint power absorption was analyzed during weight acceptance and forward propulsion. In addition, power generation was analyzed during forward propulsion. Timings and magnitudes of joint powers per condition were evaluated to identify the training combinations of body weight support and speed that best preserved the powers of unsupported walking at slow, comfortable and fast speeds.
Results: For all speeds examined, increasing body weight support to 30% without changing speed provided the best match. In general, changes in speed disrupted the joint power magnitudes and timings more than application of body weight support. Increasing body weight support when faster training speeds were used proved a viable method for reproducing the joint powers of slow, unsupported walking.
Conclusions: These data provide a reference for understanding the effect of potential training conditions on power absorption and generation within the lower extremity joints during walking. It is possible to reproduce the joint powers of unsupported walking with certain combinations of body weight support and speed. We recommend applying adequate levels of BWS when training speeds are faster than the overground speed goal, as occurs during treadmill-based locomotor rehabilitation of individuals with incomplete spinal cord injury.

1. Introduction

Treadmill-based walking rehabilitation using body weight support (BWS) seeks to engage the nervous system and reproduce the afferent stimulation associated with normal locomotion [1–7]. Having been shown to trigger muscle activity in the absence of supraspinal input [4–7], body weight supported treadmill training (BWSTT) has pioneered a shift in spinal cord injury (SCI) neurorehabilitation, moving beyond compensation into recovery of function [8–16]. Current clinical practice of BWSTT provides individuals with incomplete SCI repetitive stepping practice at the *highest gait speed* and *lowest BWS level* tolerated by patients [8,9]. These clinical guidelines are supported by research addressing the effects of speed and BWS on efferent signals [5,7]. For instance, training at speeds of 0.89 m/s and higher engages the spinal cord and induces stepping after injury [7], so rehabilitation at this speed and faster – as high as 3.58 m/s [15] – occurs clinically. In addition, relatively lower BWS levels are motivated by evidence that ankle plantarflexor muscle activity decreases at BWS levels above about 50% [5]. Despite clinical implementation of these best known practices, some BWSTT participants do not improve as rapidly or as much as others, and this difference in outcomes is only partly explained by factors such as time since injury and functional status of participants at enrollment [10,14,15].

Prior work demonstrates that muscle activation patterns in humans, including individuals with SCI, depend on peripheral feedback related to both kinematics and kinetics [6]. Because the spinal cord learns what it practices [1], this suggests that fine-tuning

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training conditions to reproduce an effective combination of the kinematics and kinetics of unsupported walking may enable neurorehabilitationists to tap into additional neurologic potential and extend recovery for individuals with SCI and other debilitating conditions. Identifying that optimal combination is challenging, though, because gait kinematics and kinetics provide varied, interrelated, and redundant sources of afferent stimulation [1]. This makes it difficult to quantitatively assess the relative contributions of these sources of peripheral information to muscle activity during stepping [5]. Furthermore, motor control factors may also play an important role. For example, eccentric and concentric activity can trigger different neural events. Repetitive eccentric contractions of one quadriceps produce greater eccentric but not concentric maximum voluntary muscle contraction in the contralateral quadriceps [25]. Muscle activity alone does not reveal whether muscle contractions are lengthening or shortening. Joint power, however, does reveal whether the net effect of the muscles acting across the joint is eccentric or concentric [26,27]. Since its calculation combines kinetic (i.e. joint moment) and kinematic (i.e. joint angular velocities) terms, joint power is a promising measure to provide integrated analysis about the motor control, kinematics, and kinetics occurring during walking. This may explain in part why it has been proposed elsewhere as a key determinant of gait [28].

Even among uninjured individuals walking with body weight supported by an overhead harness, though, the impact of clinically relevant BWS and speed parameters on joint power is largely unknown. Research to date has been conducted almost exclusively at speeds $\leq$1.4 m/s [17–22,24], with only a few studies examining speeds between 1.4 and 1.7 m/s [23,30] and none employing the higher speeds that can be found in clinical practice [15]. Among these studies, two focus exclusively on kinematics [17,18], while most examine some combination of kinematics and kinetics [19–24]. Lewek [23] is the only prior work to report joint power in the context of varying BWS and speed and did so exclusively at the ankle. Moreover, Lewek's analysis is limited to power *generation* magnitude during forward propulsion. Power *absorption*, which represents net eccentric activity about a joint, has been unexamined in terms of magnitude or timing at any joint, even though it is prevalent during the weight acceptance phase of walking [27] and may potentiate muscle activity on the contralateral leg [25]. In summary, no prior study has reported the training conditions that best reproduce the joint powers of unsupported walking.

This study seeks to fill that gap by evaluating how well speed and BWS combinations that are representative of the clinical training parameters for individuals with SCI (i.e. training speeds above 0.89 m/s and BWS less than or equal to 50%) reproduce the lower extremity (LE) joint powers of unsupported walking at a range of speeds among healthy subjects. Identifying the edges, or contours, of training conditions that match the joint powers associated with unsupported walking at a variety of speeds will help remove a critical barrier to clinical translation. Prior work has shown that joint powers, especially at the knee, increase with increasing walking speed on the treadmill [26] and more generally that interaction effects exist between BWS and speed [22,23,30]. Joint moment sensitivity data indicate that hip and knee joint extension moments were more sensitive to changes in speed (<1.4 m/s) than BWS (0–60%), with the opposite observed for knee flexion and ankle plantarflexion moments [22]. Lewek's data appear to show greater changes in ankle power generation with variations in speed ($\leq$1.6 m/s) than with variations in BWS (0–40%), although these relationships were not the focus of the paper [23]. Based on our reading of these data [22,23,26], we hypothesized that the timings and magnitudes of LE joint powers associated with unsupported walking during weight acceptance and forward propulsion would be preserved better (i.e. show fewer disruptions) when BWS was applied than when speed was increased or decreased relative to the overground speed goal.

2. Methods

The protocol for this study was approved by The Ohio State University's Institutional Review Board.

2.1. Subjects

Twelve healthy subjects free of neurologic or orthopedic impairment were consented. Eight had complete data sets and were used for this analysis (4F/4M; 23.4 $\pm$ 2.7 year; 171.1 $\pm$ 10.0 cm; 69.0 $\pm$ 13.6 kg).

2.2. Data collection

Kinematic (Vicon, 7 camera; 200 Hz) and force (Bertec, split-belt instrumented treadmill, 1000 Hz) data were collected while subjects walked on a treadmill. Subjects wore a body-weight support (BWS) harness and, with a pneumatic system that is used clinically, were provided 0, 30 or 50% BWS in randomized order while walking at their slow, fast or comfortable speed. Comfortable speed was recorded from a 10 m walk test (1.23 $\pm$ 0.13 m/s). Slow and fast speeds were calculated as 50% (0.64 $\pm$ 0.07 m/s) and 150% (1.93 $\pm$ 0.21 m/s) of comfortable speed, respectively. Data collected with 0% BWS at each of the three speeds from these subjects have been previously reported in an analysis that examined the kinematic markers that best represent the kinetics of weight acceptance [27]. Point cluster marker sets [29] were applied to the foot, shank and thigh unilaterally, with additional markers of anatomical landmarks placed on the contralateral side, pelvis, and trunk. Four-to-12 steady-state gait cycles per subject were analyzed. Low-pass filters were applied to kinematic (Butterworth, 6 Hz) and debiased force (critically damped, 10 Hz) data [27]. Using Visual 3D software, joint powers were calculated throughout the gait cycle.

2.3. Biomechanical data reduction

Using custom MATLAB code, joint power absorption was calculated during the weight acceptance phase of gait (WA−) from initial contact to peak hip abduction angle [27]. During the propulsive phase of gait, both power absorption (PR−) and power generation (PR+) were analyzed. The propulsive phase was defined as the last 35% of stance. Peak (Watts/bodyweight in kg) and time to peak power (% of gait cycle) were analyzed within WA−, PR+, and PR− for the following joint aspects: sagittal ankle (AS), sagittal knee (KS), frontal hip (HF) and sagittal hip (HS).

2.4. Statistical analysis

Potential training conditions were parametrically varied across clinically relevant ranges and compared to each speed goal (slow, comfortable, and fast, all unsupported). Mean peak magnitude and mean time to peak power in WA−, PR−, PR+ were compared (ANOVA). Where significance was indicated, Tukey's post hoc analysis was applied.

2.5. Identification of "best fit" conditions

Conditions were identified in which all variables examined matched the speed goal (+++), all but 1–2 variables matched (++) and all but 3–4 variables matched (+). All variables that differed from the speed goal ($p < 0.05$), including the magnitude and timing of each joint examined, were equally weighted in determination of best fit.