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Variations of handheld loads increase the range of motion of the lumbar spine without compromising local dynamic stability during walking

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ABSTRACT

Background: Walking is often considered a beneficial management strategy for certain populations of low back pain patients. However, little is known about how simple challenges that people often encounter, such as carrying loads in the hands, affect the low back during walking.

Research Question: How do variations in hand loading affect arm swing, lumbar spine range of motion (ROM), and lumbar spine local dynamic stability (LDS) during walking?

Methods: Sixteen young healthy participants (8 female) performed nine treadmill walking trials, each at 1.25 m/s for 3 consecutive minutes. Conditions manipulated the magnitude of hand loads (unloaded, low, high) and location of hand loads (directly in hands, in bags). Kinematic markers were used to measure sagittal plane arm swing, 3D lumbar spine ROM, and lumbar spine LDS during each trial.

Results: Arm swing was significantly ($p < 0.001$) reduced as load increased directly in the hands; however, when held in bags load magnitude had no effect. Further, arm swing was significantly ($p < 0.0001$) lower when loads were held in bags. Lumbar flexion/extension ROM was greatest with the low load compared to both unloaded ($p = 0.012$) and high load ($p = 0.0717$) conditions, and was also greater ($p < 0.0001$) with loads held directly in the hands compared to loads in bags. Despite these changes in lumbar spine ROM, lumbar spine LDS was not significantly affected by any of the variations in hand loading.

Significance: The greater lumbar spine cyclic motion, elicited by low hand loads held directly in the hands during walking, may be beneficial to the health of the low back. No changes in lumbar LDS were found, thereby suggesting that the small, likely beneficial, increases in lumbar spine ROM are well controlled by the motor control system and do not create an increased risk of injury.

1. Introduction

Walking is often recommended as a management strategy for certain populations of low back pain (LBP) patients, stemming from evidence that exercise can help to manage chronic pain [1,2], and that walking in particular elicits beneficial movement of the spine [3]. Low levels of cyclic loading and muscle activation, such as those experienced by the spine when walking, may have the potential to increase nutrient and molecular exchange through the intervertebral disc (IVD) [4]. Additionally, movement generated during walking may have the potential to mobilize the lower spine and pelvis in patients who've adopted a maladaptive stiffening strategy in response to their LBP [5]. Walking as a therapeutic programme is not only easy for patients to adhere to, but has a low cost on the individual and on the health care system, making it an accessible therapeutic tool [1,6]. A number of studies have examined the benefits of walking for people with chronic LBP. It has been found that a walking programme can improve function/disability, pain, quality of life, and fear avoidance within 8 weeks to 6 months [1,6–8], and is sometimes as effective or more effective than other interventions.

Local dynamic stability (LDS) has been used to examine the neuromuscular control of human movement by examining the behaviour of the kinematic variance. In walking, LDS is has been shown to negatively correlate with a risk of falling in elderly patients [9,10] and those with neuromuscular disorders such as paresis of the lower extremities [11]. LDS during walking has also been shown to improve with rehabilitation in multiple sclerosis patients who are at a higher risk of falls [12], thereby demonstrating the clinical utility of LDS measures [9]. LDS is sensitive to changes in walking speed [13] and arm swing magnitudes [14–16], while lumbar spine LDS during repetitive flexion/extension movements has been shown to change with fatigue [17], movement speed and load [18], and experimentally induced LBP [19].

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While brisk walking with natural arm swing has been advocated as useful in back exercise and rehabilitation programs, we were motivated to determine if additional strategies could be implemented to take further advantage of natural adaptations to walking variations (e.g. hand-held masses). We were also interested in understanding how other variations in hand-held masses, typically encountered in everyday tasks (e.g. carrying handheld loads in bags), could affect the potential efficacy of a walking intervention or strategy. The current work was designed to assess the effects of various hand loading conditions on low back movement and control during walking. Specifically, three dependent factors were examined: 1) lumbar spine angular range of motion (ROM), 2) lumbar spine LDS, and 3) arm swing magnitude. These factors were examined in relation to the independent factors of: load magnitude and load location (directly in the hands vs. in bags).

2. Materials and methods

2.1. Participants

Sixteen participants, eight male (mean (± SEM) age 21.6 ± 0.8 years, height 178.1 ± 1.8 cm, and weight 75.3 ± 2.10 kg) and eight female (22.0 ± 0.3 years, 164.4 ± 1.8 cm, and 65.6 ± 4.7 kg) completed the study. Exclusion criteria included any persistent pain, or treatment for pain or injury, to the lumbar spine, legs, or arms (causing absence from school, work or regular activity) within the past six months. All participants were rested and had not completed any intense physical activity in the 24 h preceding the experiment. Prior to any testing, participants were informed of the intent, procedures, and risks of the experiment before providing their informed consent. The experimental protocol was approved by the University Research Ethics Board.

2.2. Materials

For all trials, participants were required to walk barefoot on a commercially available treadmill (Free Spirit, Sears, Canada; walking surface size of 141.5 by 50.5 cm). Optoelectronic kinematic data (Optotrak 3D Investigator, Northern Digital, Waterloo ON, Canada) were collected from six rigid bodies each consisting of three non-collinear, infrared emitting markers. These rigid bodies were placed on the back superior and inferior to the lumbar spine (12th thoracic and 1st sacral levels), the medial side of both wrists (ulnar styloid), and on the posterior aspect of both heels (Fig. 1). All kinematic data were sampled at 64 Hz.

2.3. Procedure

Participants completed eight separate load carriage trials and one unloaded control trial at the same velocity and duration (1.25 m/s for 3 min), while focusing their gaze at a point on a wall in front of them. The order of the trials was randomized. Load carriage trials were designed to manipulate two independent variables: load magnitude (1.14 kg or 2.27 kg in each hand), and load location (directly in the hands vs. in bags). For the hand-held loads, wrist weights were held in the hands. For the bag loads, the wrist weights were placed in small (30.5 cm × 24 cm × 10 cm) cloth bags that measured 54 cm from the handle in the hands to where the load sat in the bottom of the bag. Participants were instructed to walk naturally, and were given two minutes of rest between conditions to mitigate the effects of fatigue, as fatigue is known to impact LDS [17]. A treadmill was used to ensure a constant walking speed, as LDS is also affected by movement speed [18,20].

2.4. Data processing

All kinematic data were analyzed using MATLAB v8.3 (MathWorks,

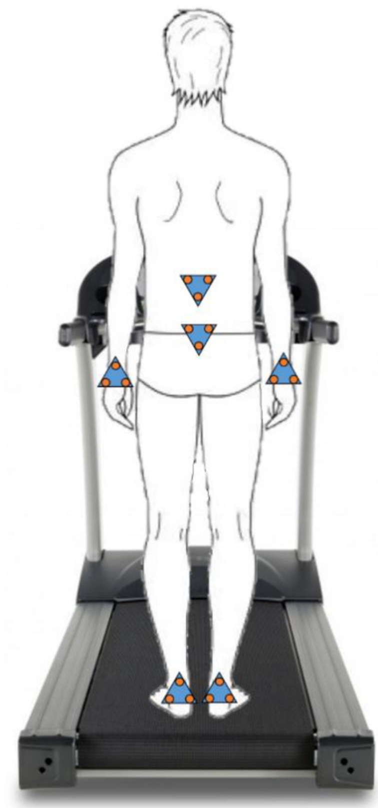

Fig. 1. Depiction of the participant on the treadmill with the location of motion capture rigid bodies shown.

Natick, MA, USA). Raw kinematic data were interpolated using a five-point cubic spline to replace any small gaps of missing data points caused by infrared marker occlusion. Kinematic data from each of the walking trials were then low-pass filtered (4th order dual-pass Butterworth, 6 Hz cut-off) to attenuate higher frequency noise picked up from the treadmill and ensure that the remaining signal was representative of true human movement. The first 3500 frames, or 54.7 s, which corresponded to approximately 50 strides [21] were removed to ensure that subjects reached a state steady of movement. Sagittal plane arm swing magnitudes were computed for each arm as the differences between the maximum and minimum position of the arm, for each stride, in the antero-posterior (AP) direction.

Three dimensional (3D) lumbar spine angles were computed between the T12 and S1 rigid bodies using a flexion-extension (FE)/lateral-bend (LB)/axial-twist (AT) Cardan rotational sequence [22]. The lumbar spine ROM in these three anatomical directions were then computed for each stride in the same manner as arm swing magnitude.

2.5. Quantifying local dynamic stability

LDS of the lumbar spine was calculated as done previously [22]. Briefly, the 3D lumbar spine angles were first biased into positive Cartesian space [23] and then converted into a single dimensional state variable by way of a Euclidean norm transformation (calculated at each instant in time) [17]. The Euclidean norm time-series was then time-delay reconstructed (10% of a stride cycle) into 6-dimensional state space. Finally, short term maximal finite-time Lyapunov exponents (λ_{max}) were calculated as the exponential rate of divergence between nearest neighboring trajectories within the reconstructed state space [24]. A larger λ_{max} indicates greater divergence and therefore greater instability, while a smaller λ_{max} indicates convergence and therefore greater stability.