Gait & Posture 50 (2016) 207–211

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

# Effects of load carriage and footwear on lower extremity kinetics and kinematics during overground walking

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ARTICLE INFO

*Article history:*
Received 19 May 2016
Received in revised form 8 September 2016
Accepted 9 September 2016

*Keywords:*
Barefoot
Shod
Walking gait
Loading

ABSTRACT

Kinetic and kinematic responses during walking vary by footwear condition. Load carriage also influences gait patterns, but it is unclear how an external load influences barefoot walking. Twelve healthy adults (5 women, 7 men) with no known gait abnormalities participated in this study (age = 23 ± 3 years, height = 1.73 ± 0.11 m, and mass = 70.90 ± 12.67 kg). Ground reaction forces and 3D motion were simultaneously collected during overground walking at 1.5 m $s^{-1}$ in four conditions: Barefoot Unloaded, Shod Unloaded, Barefoot Loaded, and Shod Loaded. Barefoot walking reduced knee and hip joint ranges of motion, as well as stride length, stance time, swing time, and double support time. Load carriage increased stance and double support times. The 15% body weight load increased GRFs ~15%. Walking barefoot reduced peak anteroposterior GRFs but not peak vertical GRFs. Load carriage increased hip, knee, and ankle joint moments and powers, while walking barefoot increased knee and hip moments and powers. Thus, spatiotemporal and kinematic adjustments to walking barefoot decrease GRFs but increase knee and hip kinetic measures during overground walking. The ankle seems to be less affected by these footwear conditions. Regardless of footwear, loading requires larger GRFs, joint loads, and joint powers.


## 1. Introduction

Overground walking is a common task young adults perform with backpack loads. These loads influence spatiotemporal, kinematic, and kinetic parameters of walking. For example, trunk loads promote shorter stride lengths at higher stride frequencies to maintain a fixed speed [1–3]. Kinematic differences with loads include greater hip and knee flexion during stance [4]. Kinetic differences include increased ground reaction force (GRF) magnitudes that are proportional to the added mass [5,6] and greater hip and knee extensor moments, hip power generation, and power absorption at the knee and ankle [4].

Walking mechanics also differ between barefoot and shod conditions. Compared to walking shod, shorter stride lengths with an increased stride frequency [7–10] and decreased ranges of motion (ROM) at the hip, knee, and ankle joints have been reported [7]. These adjustments to barefoot walking are associated with decreased braking and vertical GRFs during early stance [8]. Additionally, peak knee flexor moments increase, and hip flexor moments decrease during early stance while barefoot [8]. Those altered joint kinetics may be a response to the increased plantar pressures experienced while barefoot [11]. Increasing body mass via an external load may exacerbate these changes, but currently the simultaneous impact of load carriage across footwear conditions is unclear.

Titchenal, Asay, Favre, Andriachi and Chu [12] compared knee kinetic responses to three footwear conditions (athletic shoe, 3.8 cm heels, and 8 cm heels) with and without a 20% bodyweight load. They reported increased knee extensor and abductor moments during stance. Compared to the athletic shoe, the 3.8 cm heel shoe promoted larger knee flexor and extensor moments [12]. Rather than raising heel height, Dames and Smith [9] investigated the kinematic and metabolic effects of treadmill walking barefoot vs. shod with trunk loads. Dames and Smith [9] used a treadmill with a rubberized slat design that likely improved comfort during barefoot walking, but may have attenuated responses to the loading condition. Additionally, the treadmill did not have force measuring capabilities, which limited the authors' ability to provide insights into lower extremity kinetics.

The present study seeks to understand how simultaneously imposing external loads and varying footwear conditions impact overground walking mechanics in young, healthy adults. Compared to previous research [9] focused on treadmill walking,

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http://dx.doi.org/10.1016/j.gaitpost.2016.09.012
0966-6362/

studying overground walking is likely to provide insights that better reflect actual tasks people may be exposed to in everyday life. Presently, it is unknown how external loads and varying footwear interact when experienced simultaneously on a level surface. The literature suggests GRFs, gait temporal measures, and lower extremity joint ROM, joint moments, and joint powers increase with loading but decrease when barefoot. Given these opposing effects, it was hypothesized that footwear*loading interactions would be evident in these kinetic and kinematic measures. Main effects were only considered if no interaction was present and would indicate that the conditions were sufficient to alter walking patterns.

## 2. Methods

### 2.1. Participants

Twelve young, healthy individuals (5 women, 7 men) with no known musculoskeletal or neurological issues that would compromise gait were recruited for this study (age = 23 ± 3 years, height = 1.73 ± 0.11 m, and mass = 70.90 ± 12.67 kg). The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

### 2.2. Experimental protocol

Participants wore tight-fitting clothing throughout the experiment so that anatomical landmarks could be easily identified and to minimize marker movements. Anthropometric data (i.e., body mass, height and various segment widths and lengths) were measured based on VICON's Plug-in-Gait model. Retroreflective markers were attached to anatomical landmarks using double-sided tape. Participants then performed overground walking trials at 1.5 $m\,s^{-1}$, which is slightly faster than the preferred speed of young adults [13], in four walking conditions: Barefoot Unloaded (BU), Shod Unloaded (SU), Barefoot Loaded (BL), and Shod Loaded (SL). Two pairs of timing gates (BROWER Timing Systems, Draper, UT) were used to ensure walking speed was within ±5% of the target speed. A backpack (Dakine, Hood River, OR) loaded with lead weights equal to 15% of the participant's body mass was worn during loaded trials. This mass was chosen for comparison with previous investigations [9,14]. The design of the backpack allowed for a stable placement of the lead weights with minimal vertical movement of the mass relative to the body. Participants wore their own athletic shoe (average shoe mass = 272 ± 68 g) for the shod conditions. The order of these four conditions was individually randomized and each participant completed all four conditions. During each trial, 3D motion (100 Hz) (VICON, Englewood, CO) and ground reaction force (GRF) (2000 Hz) data were collected. GRFs were measured by a tandem-belt instrumented treadmill (AMTI, Watertown, MA) embedded in the center of the walkway. Trials included in the data analysis were within the expected velocity range and clean foot contacts were made with the force plates (i.e., a single, whole foot contact per force plate).

### 2.3. Data analysis

Markers were labeled within VICON Nexus, but all subsequent processing of data was performed using a custom Visual 3D (C-Motion, Germantown, MD) script. Marker data were filtered using a recursive, Butterworth lowpass filter ($F_c$ = 6 Hz). This cutoff frequency was confirmed by a residual analysis performed in MATLAB (MathWorks, Natick, MA) as described by Winter [15]. GRF data were filtered using a recursive, Butterworth lowpass filter ($F_c$ = 50 Hz). Motion and GRF data were combined through inverse dynamics to estimate joint reaction forces and moments for the ankle, knee, and hip in the sagittal plane. Joint angular positions at contact, ranges of motion, and velocities were also included as dependent measures. Joint powers were calculated as the product of the joint moment and angular velocity. Joint power peaks from the phases defined by Winter [16] were used in statistical analyses. These include two ankle phases (A1, A2), four knee phases (K1-K4), and three hip phases (H1-H3). A1 is the initial weight acceptance and A2 the propulsive peak at toe-off. K1 is the energy absorption phase during weight acceptance. K2 is the only power generation phase and occurs during mid-stance. K3 is power absorption during terminal stance and early swing, and K4 is terminal swing power absorption. H1 phase occurs during early stance, H2 is an absorption phase during mid-stance, and H3 a power generation phase prior to toe-off. Finally, spatiotemporal measures including stance time, double support time, swing time, and stride length were included to further characterize the effects of loading and varying footwear.

### 2.4. Statistical analysis

Dependent variables were determined from three successful strides and then averaged. Using a single group design, a series of 2 × 2 (loading, footwear) ANOVAs with repeated measures were performed using IBM SPSS 23.0 (Armonk, NY). Interaction and main effects were investigated with an alpha level of 0.05 for set for significance. Bonferroni post hoc tests were performed where pairwise comparisons were appropriate.

## 3. Results

### 3.1. Combined loading & footwear

While loaded and barefoot, the expected increases with load and decreases while barefoot were offsetting for the braking and propulsive GRFs (Fig. 1), double support time (Table 1), and hip ROM (Table 2). These counteracting responses resulted in no difference from the shod, unloaded condition for these measures (i.e., BL = SU). The only footwear by load interaction was observed for hip ROM ($p$ = 0.016). Hip ROM increased with load while walking shod, but when load was added while walking barefoot, hip ROM did not change.

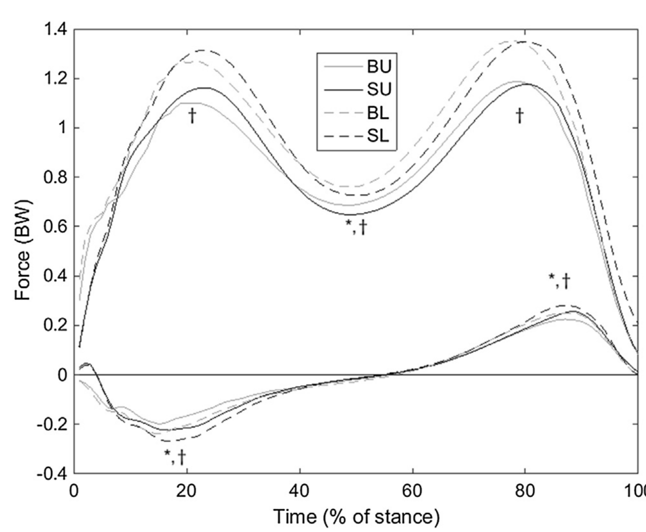


**Fig. 1.** Vertical and anteroposterior ground reaction force profiles. * indicates significant footwear effect and † indicates significant load effect ($p$ < 0.05).