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Musculoskeletal stiffness changes linearly in response to increasing load during walking gait

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ABSTRACT

Development of biologically inspired exoskeletons to assist soldiers in carrying load is a rapidly expanding field. Understanding how the body modulates stiffness in response to changing loads may inform the development of these exoskeletons and is the purpose of the present study. Seventeen subjects walked on a treadmill at a constant preferred walking velocity while nine different backpack loading conditions ranging from 12.5% to 40% bodyweight (BW) were introduced in an ascending and then descending order. Kinematic data were collected using Optotrak, a 3D motion analysis system, and used to estimate the position of the center of mass (COM). Two different estimates of stiffness were computed for the stance phase of gait. Both measures of stiffness were positively and linearly related to load magnitudes, with the slopes of the relationships being larger for the descending than the ascending conditions. These results indicate that changes in mechanical stiffness brought about in the musculoskeletal system vary systematically during increases in load to ensure that critical kinematic variables measured in a previous publication remain invariant (Caron et al., 2013). Changes in stiffness and other kinematics measured at the 40% BW condition suggest a boundary in which gait stiffness control limit is reached and a new gait pattern is required. Since soldiers are now carrying up to 96% of body weight, the need for research with even heavier loads is warranted. These findings have implications on the development of exoskeletons to assist in carrying loads.

1. Introduction

The United States Army has identified musculoskeletal injuries as the greatest threat to military readiness (McNulty, 2009) and the effects of load carriage substantially contribute to such injuries among soldiers (Knapik et al., 2004). Modern military forces are expected to carry increasingly heavy loads (Knapik et al., 2004) ranging from 27% to 45% of their bodyweight (BW) for typical combat loads and exceeding 96% BW for emergency situations (Dean, 2004). One solution is to build exoskeletons that aid the soldier in carrying heavy loads (Kazerooni et al., 2007). However, the complexity of the control system, weight, noise, durability and force-producing limitations of the actuators, and increased metabolic cost for the wearer (e.g. Gregorczyk et al., 2010), present substantial challenges for designing exoskeletons that satisfy the functional and energetic demands for military and rehabilitation applications (Herr, 2009).

A promising research direction in this regard is to develop soft exoskeletons using materials that have variable stiffness to help support and return elastic energy to the body. One example is the development of electroactive polymers with variable-stiffness properties akin to human soft tissues (Herr and Kornbluh, 2004; Mulgaonkar et al., 2008; Pelrine and Kornbluh, 2008). The goal of the present study is to explore how the human body modifies stiffness in response to varying loads to inform bioinspired development of assistive load-carriage technologies.

The role of stiffness and elastic energy return in the performance of tasks such as hopping (Farley and Gonzalez, 1996), walking (Holt et al., 1996; Kubo et al., 2006) and running (McMahon and Cheng, 1990) has been widely investigated (see Butler et al. (2003), for a review); however, little research has been conducted to determine how the body changes stiffness in response to carried load. An exception is our previous work showing that stiffness ('global', vertical and knee stiffness in the sagittal plane) increases linearly with walking speed both without load and while carrying a backpack holding a load of 40% BW (Holt et al., 2003). However, the relationship between stiffness and load has not been investigated across the range of 0–40% BW loads.

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A number of spring and pendulum models have been used to estimate stiffness. In this paper we include the findings from two models. Our own model includes a state-dependent forcing function with passive energy conservation through the pendular body segments and soft-tissue springs (Escapement-Driven, Inverted Pendulum and Spring [EDIPS]) (Fonseca et al., 2004; Holt et al., 1996, 2003). In the EDIPS model, the term 'global stiffness' is used to label a collective measure of torsional stiffness due to all active (muscular) and passive (fascial–muscular–tendinous) sources of elastic stiffness that influence the inverted body pendulum during the stance phase of gait. A second model calculates the body's vertical stiffness and has been used previously for hopping (Farley and Morgenroth, 1999), and running (Farley and Gonzalez, 1996; McMahon and Cheng, 1990). Vertical stiffness is considered a mechanism to resist vertical collapse of the body (McMahon and Cheng, 1990), as the center of mass descends after mid-stance and changes in vertical stiffness have been shown to originate, primarily, from adaptations in torsional stiffness about lower-extremity joints (Farley and Morgenroth, 1999). Thus the global stiffness measure captures more of the factors that influence the total body without assigning it to specific body parts, while vertical stiffness is more specific to the lower extremity joints during the loading stage.

In a previous study we have reported that the center of mass of body + load (COM_{TSYS}) trajectory and lower extremity joint angles in the sagittal plane remain invariant across a wide range of loads (Caron et al., 2013). Similarly, we have reported that the vertical orientation of the vector from the ankle to the COM_{TSYS} and from the knee to the COM_{TSYS} in the sagittal plane remains invariant (Caron et al., 2013). We have claimed that these invariants simplify the motor control process and we hypothesize that the mechanism for maintaining invariance with systematic increases and decreases of load is through linearly proportional increases in global and vertical stiffness.

2. Methods

Seventeen individuals participated in the study (9 males, 8 females, age 25.4 ± 5.2 years; mass 70.6 ± 11.0 kg; height 1.7 ± 0.07 m). Subjects had no history of cardiopulmonary disease, neurological impairment, or injury that would limit treadmill walking for longer than 1 h. The Institutional Review Board of Boston University approved the study and subjects provided written informed consent.

Body segments were measured for anthropometric calculations, using anatomical landmarks (Table 1), while subjects were supine. The subjects were then fitted with 20 Infrared Light Emitting Diodes on those anatomical landmarks. Subjects walked on a level treadmill (Kistler Instrument Corporation, Amherst, MA) without load to determine their preferred walking velocity, using the procedure detailed in Holt et al. (1991). Treadmill velocity was held at the subjects' preferred speed (mean ± SD = 1.11 ± 0.12 m/s) throughout the experiment. Three-dimensional kinematic data were collected through an Optotrak 3020 system (Northern Digital Inc., Waterloo, Ontario, Canada) at a sampling rate of 100 Hz.

Table 1
List of IRED marker locations placed on the body and backpack, else calculated, bilaterally from which all independent variables were reduced.

Markers	Anatomical or calculated location
Head	Zygomatic process
Shoulder	Acromioclavicular joint
Elbow	Lateral epicondyle of humerus
Wrist	Ulnar styloid process
Virtual hip	Greater trochanter = 3 × distance from knee to thigh
Thigh	1/3 distance from knee to greater trochanter
Knee	Lateral epicondyle of femur
Ankle	Lateral malleolus
Foot	5th metatarsal head
Upper pack	Upper mid-line of backpack tank
Lower pack	Lower mid-line of backpack tank

Load was carried using a water tank mounted to an aluminum backpack frame and secured to the thorax. Water was systematically added and drained, from the backpack while subjects walked continuously on the treadmill for approximately one hour. Detailed descriptions of the load manipulation apparatus and the experimental procedure are provided in Caron et al., (2013). Load conditions included in the present study began at 12.5% BW for all subjects, then increased in 2.5% increments to 30% BW (ascending load conditions). Load then increased to 40% BW, following which it was systematically decreased (descending load conditions) to 30% BW and then, in 2.5% increments, to 12.5% BW. The range and incremental changes in load between 12.5% and 30% BW were selected to best assess the linearity of stiffness changes between relatively light and heavier loads. The 40% BW condition was included to allow comparisons to previous studies that used a 40% BW loaded condition (Holt et al., 2003; LaFiandra et al., 2002, 2003), and to determine whether actual measures of stiffness at 40% BW were predictable from the lower range. The decision to systematically add and remove the loads versus randomize the loads was made so that the range and increment of loads could be accomplished in one experimental session with under 1 h of continuous walking with load in order to minimize the potential impacts of fatigue (Patton et al., 1991). In addition, we were interested in determining if there were sudden transitions in coordination patterns as load was systematically increased or decreased. Participants walked for 2.5 min in each load condition and data were collected during the last minute. All subjects completed all trials without requests for breaks.

Missing three-dimensional position data were interpolated using a cubic spline (MATLAB, The MathWorks, Natick, MA). All data were then filtered using a zero-phase shift, second order, low-pass Butterworth filter at 5 Hz. Gait events were determined from kinematics using custom algorithms adapted from Hreljac and Marshall (2000) for heel-strike (HS) and from De Witt (2010) for toe-off (TO). Mid-stance (MS) was calculated as 50% time between HS and TO (Gibson et al., 2006). All gait events and stiffness measures were calculated using only left-side data. HS was used to separate data into strides (average 51 strides per condition). For each subject, strides with more than 20 consecutive frames of interpolated data were excluded (average 3 strides per condition). An average of 48 strides per condition was included after exclusion procedures were completed. All dependent variables were calculated using custom MATLAB programs.

Global stiffness (k_G) is estimated from the mass (M_{TSYS}), length (L_e) and natural frequency ($2\pi/\tau$) parameters of the EDIPS model oscillating at its natural (resonant) frequency. The EDIPS model was applied to the kinematic data measured during the stance phase of gait (Holt, 1998; Holt et al., 2003, 2000, 1996) to estimate the system's global stiffness using the following equation:

$$k_G = (2\pi/\tau)^2 M_{TSYS} L_e{}^2 + M_{TSYS} L_e g, \qquad (1)$$

where τ is double the stance period (the time interval from HS to ipsilateral TO of the stance leg), M_{TSYS} is the total mass of the system (body + load) located at its center of mass (COM_{TSYS}), L_e is the simple pendulum equivalent length estimated as the mean distance from the COM_{TSYS} to the axis of rotation (ankle) across each stance period, and g is the acceleration due to gravity. The position of COM_{TSYS} was computed by combining the position vector of the body center of mass (estimated using kinematic data specifying the locations and orientations of the body segment, along with known segmental mass and anthropometric ratios from de Leva (1996) with the position vector of the load center of mass (estimated using kinematic data of the water-tank's position and orientation, and the known volume and calculated geometric distribution of the water in the tank). Global stiffness was measured for each stance period, and the average global stiffness across all measured strides in each loaded condition was used for analysis.

Stance period (SP) and L_e were independently assessed because they are parameters that influence global stiffness in addition to M_{TSYS} (Eq. (1)). Increased stride frequency (SF) has been reported at higher levels of load when walking velocity is constrained (LaFiandra et al., 2003). We would expect SF to behave similarly to SP; however, we presented both measures to confirm this expectation. SF was measured as the inverse of stride period (time from successive HS values of the left leg). SP was measured as the time interval from HS to ipsilateral TO of the stance leg. The means of SP, SF and L_e for each loaded condition were used for analysis.

Vertical stiffness (k_v) is a measure of the collective forces acting to resist the vertical collapse of the COM_{TSYS} as it descends after mid-stance (see Fig. 1) (Holt et al., 2003). The time period of COM_{TSYS} descent (max to min vertical COM_{TSYS} position) includes both the time periods directly after mid-stance and the initial loading phase of the contralateral limb after heel strike. Thus k_v is measured during a period of both single and double support. The total force acting upon the COM_{TSYS} ($M\ddot{x}$) in the vertical dimension and the resulting displacement (x–x_0) are used to calculate vertical stiffness (k_v) according to the following equation:

$$-k_v = M_{TSYS}\ddot{x}/(x - x_0), \qquad (2)$$

where M_{TSYS} is the total mass of the system, $\ddot{x}$ is the vertical acceleration of the COM_{TSYS}, x is the current vertical position of the COM_{TSYS}, and x_0 is the equilibrium position of the COM_{TSYS} assumed to be located at the mean vertical position of the COM_{TSYS} between the maximum COM_{TSYS} vertical position and the subsequent minimum COM_{TSYS} position during the stance phase (Fig. 1). Mean vertical stiffness