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Adaptive changes in running kinematics as a function of head stability demands and their effect on shock transmission

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ABSTRACT

This study aimed to identify adaptive changes in running kinematics and impact shock transmission as a function of head stability requirements. Fifteen strides from twelve recreational runners were collected during preferred speed treadmill running. Head stability demands were manipulated through real-time visual feedback that required head-gaze orientation to maintain within boxes of different sizes, ranging from 21° to 3° of visual angle with 3° decrements. The main outcome measures were tibial and head peak accelerations in the time and frequency domains (impact and active phases), shock transmission from tibia to head, stride parameters, and sagittal plane joint kinematics. Increasing head stability requirements resulted in decreases in the amplitude and integrated power of head acceleration during the active phase of stance. During the impact portion of stance tibial and head acceleration and shock transmission remained similar across visual conditions. In response to increased head stability requirements, participants increased stride frequency approximately 8% above preferred, as well as hip flexion angle at impact; stance time and knee and ankle joint angles at impact did not change. Changes in lower limb joint configurations (smaller hip extension and ankle plantar-flexion and greater knee flexion) occurred at toe-off and likely contributed to reducing the vertical displacement of the center of mass with increased head stability demands. These adaptive changes in the lower limb enabled runners to increase the time that voluntary control is allowed without embedding additional impact loadings, and therefore active control of the head orientation was facilitated in response to different visual task constraints.

1. Introduction

Head stabilization during locomotion affords a consistent visual base during ambulation (Mulavara and Bloomberg, 2002; Pozzo et al., 1990). Head stabilization, defined as the maintenance of head-in-space equilibrium (Cromwell et al., 2001), facilitates optimal conditions for vestibular and visual function during locomotion (Keshner et al., 1995; Pozzo et al., 1990). Although the vestibulo-ocular reflex (VOR) can compensate for head rotations during locomotion through eye rotations, this reflexive adjustment is limited and only works appropriately when head velocities are under 350° s^{-1} (Pulaski et al., 1981). Stabilizing the head in space optimizes the effect of the VOR (Leigh and Zee, 2006), which can be accomplished through both active and passive regulation of the head and other body segments.

During normal walking, healthy individuals maintain a high degree of head stability through compensatory movements such as adjustments in head pitch that counteract linear and angular motions imposed by the whole body (Grossman et al., 1988; Pozzo et al., 1990). The degree to which the head is stabilized during locomotion is determined predominantly by frequency and velocity of head perturbations (Grossman et al., 1988; Keshner et al., 1995). The impact of different lower limb extremity movements as a function of gait speed (Hirasaki et al., 1999; Latt et al., 2008), stride rate (Holt et al., 1995; Latt et al., 2008), and step length (Latt et al., 2008) on the frequency characteristics of the head have been well documented in walking. Upper body changes in arm swing and trunk rotation are also associated with head stability (Cromwell et al., 2004a; Cromwell et al., 2001). These gait changes require modulation of the range and amplitude of frequency contents of the impact shock transmitted to the head (Grossman et al., 1988; Grossman et al., 1989). As frequency and amplitude of many gait variables are upregulated during running, the resulting increase in perturbations to the head will require greater compensation to stabilize the head.

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The shock induced from the foot-ground collision during running is dissipated by the combination of passive (e.g., deformation of running shoe and heel pad, ligaments, muscle oscillation, increase in knee flexion, and limited pronation of the foot) and active mechanisms (e.g., increased muscle activation) in both the upper (Cromwell et al., 2001; Kavanagh et al., 2006) and lower body (Boyer and Nigg, 2007; De Clercq et al., 1994; Edwards et al., 2012; Paul et al., 1978; Perry and Lafortune, 1995). Low frequency (4–8 Hz) tibial accelerations represent voluntary lower extremity motion during stance, while accelerations in the higher frequency range (10–20 Hz) represent the rapid deceleration that accompanies initial foot-ground contact (Bobbert et al., 1991; Edwards et al., 2012; Hamill et al., 1995). The signal power contained in these lower and higher frequency ranges differ with changes in running dynamics and how these frequencies are attenuated is specific to the mechanisms and strategies available (Boyer and Nigg, 2007; Paul et al., 1978; Simon et al., 1981).

Previous studies on running have demonstrated modulation of shock transmitted through the body across a range of impulse loads, resulting in stable head accelerations (Gruber et al., 2014). Increased shock attenuation has been observed in response to changes in gait parameters such as higher speeds (Mercer et al., 2010), lower stride frequency (Hamill et al., 1995), and longer stride length (Derrick et al., 1998). The knee joint has consistently been reported to play an important role in shock attenuation, specifically through greater knee flexion (Derrick et al., 1998; Edwards et al., 2012). The stabilization of the head in response to altered stride parameters and running speed has been well established; the inverse, how the body adapts and attenuates impact shock to different visual task or head stability demands is less clear.

Therefore, the purpose of this study was to identify how individuals adapt their running kinematics and the subsequent alterations in shock transmission in response to increases in head

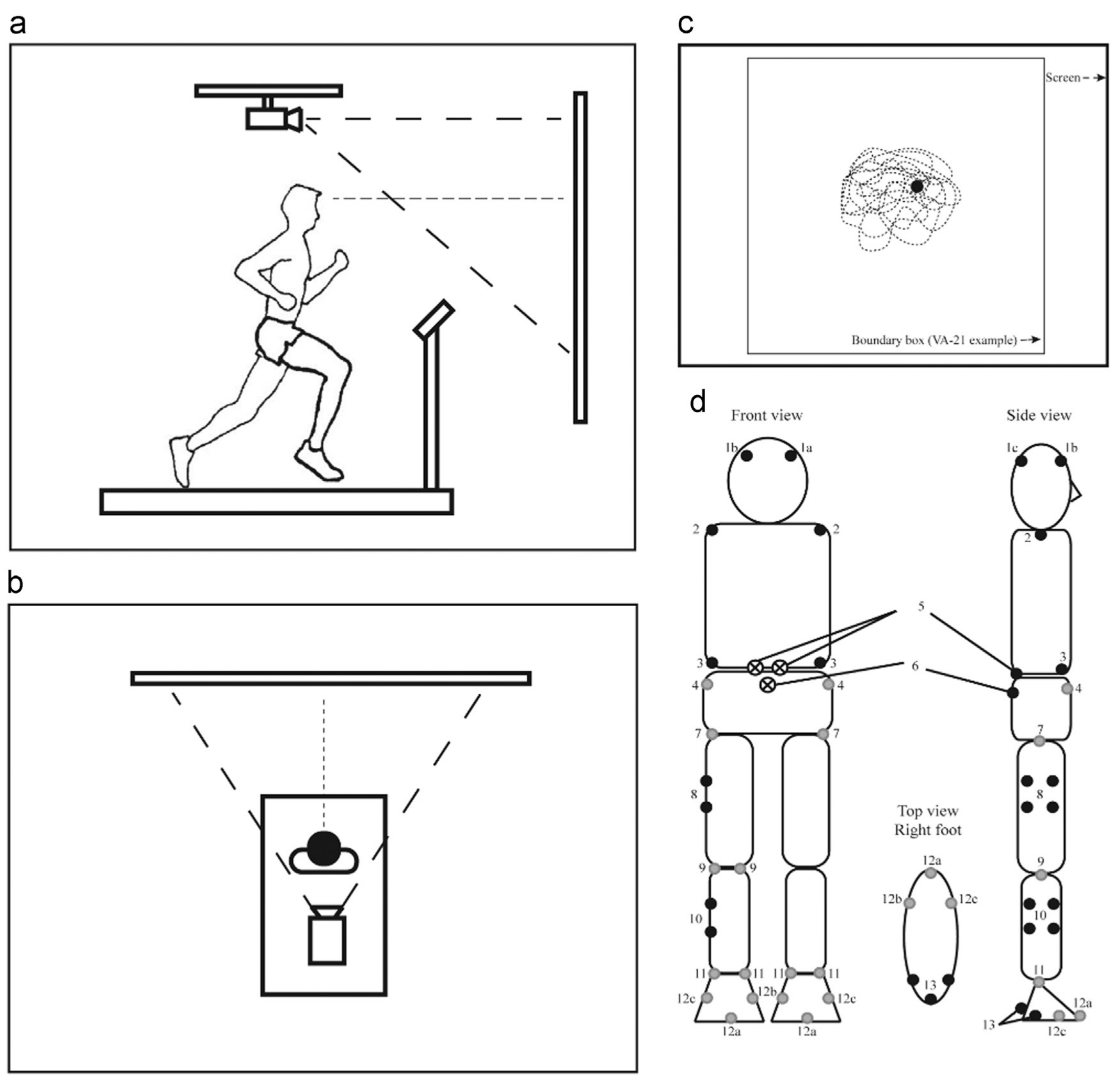

Fig. 1. Experimental setup: (a) side and (b) top view. A screen was positioned 2.5 m away from the treadmill center. A thin dashed line from the participant's head in (a) and (b) indicates an imaginary line of an extended head vector created from the 6-DOF object worn on the head. The interaction point of head gaze vector on the screen (black dot in (c)) was displayed during running. The thin dashed line in (c) represents an imaginary trajectory of the head gaze point on the screen, but this line was not displayed during the testing. A square box (light solid line in (c)) represents the area in which the head projection vector had to be maintain, and the angles subtended by the box horizontally and vertically were varied in each condition from 21° to 3° of visual angle with 3° decrements. Retro-reflective marker placement (d). Four markers were placed on the (1) head ((a) left and (b) right front, left (not shown in the figure) and (c) right back); four on the trunk ((2) left and right acromion, (3) left and right iliac crest); five on the pelvis ((4) left and right ASIS, (5) left and right PSIS, (6) sacrum); (7) the left and right greater trochanter; (8) a four marker cluster on the right thigh; (9) the medial and lateral epicondyle of the right femur; (10) a four-marker cluster on the right shank; (11) the medial and lateral malleolus of the left and right fibula; and (12) six on the left and right shoe ((12) (a) the head of the second phalanx, the head of the (b) first and (c) fifth metatarsal, (13) a three-marker cluster on the calcaneous). Markers colored grey were attached only during the calibration period and used for creating the anatomical model. Markers with cross-hairs indicate the markers attached on posterior side.