Lumbar and cervical erector spinae fatigue elicit compensatory postural responses to assist in maintaining head stability during walking

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Kavanagh, Justin J., Steven Morrison, and Rod S. Barrett. Lumbar and cervical erector spinae fatigue elicit compensatory postural responses to assist in maintaining head stability during walking. J Appl Physiol 101: 1118–1126, 2006. First published June 8, 2006; doi:10.1152/japplphysiol.00165.2006.—The purpose of this study was to examine how inducing fatigue of the lumbar erector spinae (ES) and cervical erector spinae (CES) muscles affected the ability to maintain head stability during walking. Triaxial accelerometers were attached to the head, upper trunk, and lower trunk to measure accelerations in the vertical, anterior-posterior, and mediolateral directions during walking. Using three accelerometers enabled two adjacent upper body segments to be defined: the neck segment and trunk segment. A transfer function was applied to root mean square acceleration, peak power, and harmonic data derived from spectral analysis of accelerations to quantify segmental gain. The structure of upper body accelerations were examined using measures of signal regularity and smoothness. The main findings were that head stability was only affected in the anterior-posterior direction, as accelerations of the head were less regular following CES fatigue. Furthermore, following CES fatigue, the central nervous system altered the attenuation properties of the trunk segment in the anterior-posterior direction, presumably to enhance head stability. Following lumbar erector spinae fatigue, the trunk segment had greater gain and increased regularity and smoothness of accelerations in the mediolateral direction. Overall, the results of this study suggest that erector spinae fatigue differentially altered segmental attenuation during walking, according to the level of the upper body that was fatigued and the direction that oscillations were attenuated. A compensatory postural response was not only elicited in the sagittal plane, where greater segmental attenuation occurred, but also in the frontal plane, where greater segmental gain occurred.

accelerometry; attenuation; transfer function; postural control

Rather than acting as a passive passenger unit, the upper body is a complex sensorimotor system, whose many components interact to facilitate support and stability during walking (51, 64). Importantly, the upper body modulates segmental oscillations arising from gait-generated events to ensure a stable and consistent trajectory of head motion during walking (4, 21, 43). Maintaining head stability is considered a fundamental task of the postural system, as a stable head ensures optimal visual-vestibular-central nervous system (CNS) integration, which influences spatially oriented behavior associated with maintaining posture (11, 32, 39).

Gait-related oscillations are generally attenuated from inferior to superior locations within the body. Accelerometer-based studies have revealed that accelerations of the head are 60–90% of the lower trunk (24, 38) and 10–20% of tibial accelerations (27, 34) during normal walking on a level surface. The process of attenuation of upper body accelerations arises from a combination of active and passive mechanisms. Passive spinal structures such as intervertebral discs and ligaments assist in shock absorption, but acting alone have a limited capacity to enhance intrinsic stability of the spine in response to mechanical loading. It is generally accepted that active mechanisms have the greatest influence on controlling motion of the upper body, as altering patterns of neuromuscular activity enable posture and stability to be maintained according to the nature of the walking task (51, 64).

Viewing the upper body as two adjacent segments allows the neck and trunk segments to be examined in regard to how the postural system facilitates head stability at separate levels of the body. The neck segment contributes to stabilizing the head not only by voluntary activation of postural muscles but also via cervico- and vestibulo-collic reflexes, which rotate the head in a compensatory manner in the presence of self-generated or externally imposed oscillations (28, 29). The trunk segment appears to minimize the impact of gait-related oscillations before reaching the head by providing a stable platform for the neck, despite the activity of the lower extremities (4, 25). Furthermore, as the frequency content of head and trunk oscillations are different in the mediolateral (ML) compared with the anterior-posterior (AP) and vertical (VT) directions (26), the attenuation properties of the trunk and neck segments differ according to direction to achieve global head stability (25).

Although the upper body dynamics of healthy individuals have been documented (6, 12, 15, 24, 26, 38, 61), little is known about how perturbation of the neuromuscular system, and in particular fatigue, influences the control of segmental attenuation and head stability during walking. This is important given that examining how the postural system operates under fatigued conditions can provide information on how the CNS organizes multiple muscle systems to facilitate posture and stability. For example, intense short-duration back-extension exercise has identified altered timing, or organisation, of erector spinae (ES) activation during manual lifting tasks (17). Central to facilitating posture and stability during walking, the ES muscle group acts to resist external flexion moments, which are greatest at heel contact, and exhibit consistent sequencing of activation during each gait cycle (4, 51, 63). Furthermore, the sequential “top down” activation of spinal extensors during normal walking is suggested to be an anticipatory control...
strategy that assists in attenuating upper body accelerations (51). In this study, it was of interest to examine how fatiguing ES at different levels of the upper body affects the ability to attenuate segmental oscillations and maintain head stability.

The purpose of this study was to examine the effect of ES muscle fatigue on head stability during walking. In particular, the complementary role that the neck and trunk segments play in modulating accelerations of the head after inducing neuromuscular fatigue of the lumbar ES (LES) and cervical ES (CES) was examined. Although changes due to fatigue can be viewed from different perspectives, a fatigue protocol was selected that altered patterns of neuromuscular activity rather than force-producing characteristics of the postural system. It was hypothesized that 1) CES fatigue would result in a diminished capacity to maintain head stability due to an altered attenuation capability of the neck segment and 2) LES fatigue would not affect head stability as the unfatigued, intervening neck segment would compensate for the diminished attenuation properties of the trunk segment to preserve head stability.

METHODS

Subjects. Eight healthy male subjects (age: 23 ± 4 yr; height: 1.81 ± 0.09 m; mass: 77 ± 12 kg) with no history of musculoskeletal pathology or injury, or formal resistance training involving back extensor muscle groups, were recruited from the university community. Written, informed consent was obtained from each participant before testing. All experimental procedures complied with the guidelines of the Griffith University Human Research Ethics Committee.

Experimental protocol. Subjects were required to attend two testing sessions to undertake preferred speed straight-line walking trials along a 30-m level walkway. In the first session, five walking trials were performed before and after subjects were subjected to trunk or neck fatigue. Forty-eight hours later, subjects performed the above-mentioned walking trials; however, the previously nonfatigued muscle group (trunk or neck) was fatigued. The sequence in which trunk or neck fatigue was induced was counterbalanced to avoid order effects. During the fatiguing and walking tasks, all subjects were encouraged to inform the examiner if any pain or severe discomfort was experienced. One week after experimental testing of each participant, a questionnaire was administered to determine if any pain or severe discomfort was experienced. One week after experimental testing of each participant, a questionnaire was administered to determine if any pain or severe discomfort was experienced.

Instrumentation. Four custom-designed, lightweight triaxial accelerometers were attached to each subject to measure three-dimensional accelerations during walking (27). An individual accelerometer node erometers were attached to each subject to measure three-dimensional accelerometers during walking (27). An individual accelerometer node consisted of two biaxial accelerometers (Analog Devices ADXL202, range ±2 g) mounted perpendicular to each other on a 1.5 × 2.6-cm printed circuit board (total mass: 6 g). Three of the four available sensing axes were used to measure linear acceleration along the VT, AP, and ML axes of the accelerometer based coordinate system. Accelerometers were attached over the occipital pole of the head with a firm-fitting elastic headband, as well as the C7 spinous process (upper trunk), the L3 spinous process (lower trunk), and 30 mm proximal to the lateral malleolus of the right leg (shank) with Leuko Sportstape Premium Plus. The C7 spinous process was selected because it represents the upper geometrical limit of the trunk (13). Using three accelerometers to measure head, upper trunk, and lower trunk accelerations enabled two upper body segments to be defined. For the purpose of this study, the segment between the lower and upper trunk will be referred to as the trunk segment, and the segment between the upper trunk and the head will be referred to as the neck segment. Each accelerometer sent data through a cable connected to a lightweight processing box (mass: 110 g) fixed to the subject's waist. The processor box contained a Hitachi microprocessor (HB/300H), a Bluetooth Personal Area Network device (Cambridge) for external communications, two AAA batteries, and a power regulation module (27). The processor box transmitted binary data at 250 Hz to a remote personal computer (PC). Custom-designed software developed in Visual Basic 6.0 (Microsoft) synchronized and displayed all acceleration data transmitted to the PC in real time. Gait velocity was monitored using three pairs of Omron (E3JK-R4M2) photoelectric light gates spaced at 5-m intervals along in the middle of the 30-m walkway.

ES maximum voluntary isometric contraction measurement. A minimum of 24 h before experimental testing, participants attended a familiarization session where maximum voluntary isometric contraction (MVIC) for the LES and CES were determined using a custom-built apparatus (Fig. 1). This apparatus was designed to seat the subject so that the upper body was perpendicular to the walkway surface. To remove the potential effect of hip extensor activity contributing to LES MVIC, the apparatus seat height was adjusted so that the trunk-thigh angle was 135° and the shank-thigh angle was 90°. Orienting the upper and lower body in this position assisted in maintaining elastic equilibrium at the hip and neutral lumbar lordosis (2). MVIC force for LES was measured using a Xtran 1-kN S-beam load cell transducer (Applied Measurements) connected in series with a fixed VT post and a chest harness. An adjustable beam attached to the fixed post supported the pelvis to ensure the hips remained stationary. Both the support beam and harness-load cell transducer setup remained parallel to the walkway surface during the fatiguing task. Overall, this configuration created a force couple between the pelvis and trunk when the subject pulled posteriorly via the chest harness and resulted in contraction of the LES. Procedures to determine CES MVIC force were similar; however, a head harness replaced the chest harness, and the adjustable support beam previously stabilizing the pelvis was placed over the sternum, so that a force couple was created between the upper thoracic region and the head. The MVIC protocol consisted of five maximal contractions of 5-s duration, with 5 min of rest between contractions. The greatest value from the load-cell data collected from all trials was taken to represent the subject’s MVIC.

Fig. 1. Apparatus used to determine maximum voluntary isometric contraction (MVIC) and induce neuromuscular fatigue of the lumbar erector spinae (LES). The setup of the apparatus was able to be modified to elicit contraction of the cervical erector spinae (CES).
ES fatigue protocol. Following the control walking trials, and immediately before the experimental trials, each subject performed the ES fatiguing procedure. The level of ES contraction throughout the fatiguing procedures was set at 60% of MVIC. To induce fatigue, a 30-s contraction was performed, followed by 30 s of rest, then a further 20-s contraction. The fatigue protocol selected for this study was similar to those used previously to study the effects of muscle relaxants on lumbar musculature (2) and to discriminate between individuals with and without lower back pain (46), according to changes in EMG spectral properties.

LES and CES fatigue were induced with the apparatus and experimental configuration outlined for the measurement of MVIC. A PC monitor was placed in front of the subject and displayed real-time force data collected from the load cell. Contraction of ES led to an increase in load-cell force, which was displayed on a continually updating graph (50 Hz), thus providing feedback regarding the level of ES contraction. For increased simplicity, a basic visual cue was also provided on the PC monitor. A green circular light (30-mm diameter) was illuminated when the force generated by the subject was within 5% of the desired target force. Any time the light was not active, the subject was instructed to either increase or decrease muscular contraction to achieve their target force output.

Bipolar Ag/AgCl EMG surface electrodes were positioned over the right LES and CES during the respective fatigue interventions to measure muscle activity during each isometric contraction. Electrodes were placed with an interelectrode distance of 20 mm and in parallel with the underlying muscle fibers of the ES. LES electrodes were placed at the level of L4, ~30 mm lateral to the spinous process. CES electrodes were placed at the level of C3, ~20 mm lateral to the spinous process. To ensure muscle activity was recorded under steady-state contraction and not during the progressive force development phase, EMG data collection commenced after the target force level had been reached. Similarly, EMG data collection ceased before the subject was instructed to stop the isometric contraction. All EMG signals were sampled at 1 kHz and amplified using a Coulbourn isolated bioamplifiers (V75-02). Electrode-skin impedance was below 5 kΩ during all test conditions. A second-order Butterworth band-pass filter with a range of 5–500 Hz was applied to all EMG data.

Data analysis: neuromuscular fatigue. Median power frequency (MF) was extracted from LES and CES EMG data collected during each contraction task and was used as an indicator of neuromuscular fatigue (31, 42, 58). Power spectral analysis was performed on a sliding 1-s window (1,000 data points) of raw EMG data. MF was determined for a window beginning at time 0, before the analysis window was shifted forward by 0.1-s epochs for the entire duration of the fatiguing task. This procedure provided 290 MF values over the duration of the 30-s isometric contraction and 190 MF values for the 20-s isometric contraction task.

Basic gait parameters. All analyses of gait patterns were performed on acceleration data collected from the middle 15 strides of each trial. A tilt correction was applied to acceleration data before analysis to correct for deviations in accelerometer axes from the global VT and horizontal planes while attached to the subject’s body. Under static conditions, the output of each accelerometer reflects the degree of tilt in the device, which can be determined and corrected for using basic trigonometry (24). To determine the effect of ES fatigue on upper body posture and hence accelerometer orientation during gait, the average accelerometer tilt relative to global horizontal during each walking trial was calculated using methods described by Moe-Nilssen (41).

Stride duration was determined from heel contact events, calculated from the zero crossing following peak negative accelerations for the Shank in the AP direction. Stride length was calculated from the gait velocity for each trial, multiplied by the subject’s average stride duration for that trial. Step length was assumed to be one-half of the stride length. Cadence was computed from the time taken to perform 15 strides and expressed in steps per minute (i.e., 120/average stride duration).

Segmental gain. A profile of gain through the upper body was determined by applying a transfer function to acceleration data collected during the walking trials. The transfer function describes the relationship between the input and output of a system. When gait patterns are examined, the frequency content from superiorly located oscillations are typically divided by the frequency content of inferiorly located oscillations. Given that the fundamental frequency of the power spectrum is governed by the step frequency, it is appropriate to consider the transfer function in regard to signal harmonics rather than individual frequencies. This enhances the validity of comparing between trials and subjects walking at different speeds. Therefore, the transfer function was defined as:

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\text{Transfer function} = 10 \log_{10} \left[ \frac{\text{Superior}(f_0)}{\text{Inferior}(f_0)} \cdot \frac{(2f_0)}{(3f_0)} \cdot \frac{(15f_0)}{(15f_0)} \right] \tag{1}
\]

where \(f_0\) is the fundamental frequency, and \(2f_0\) and \(3f_0\) are the second and third harmonics of the fundamental frequency. Using a fast Fourier transform, acceleration signal harmonics were extracted up to the 15th harmonic (15\(f_0\)). The transfer function determined whether accelerations gain (positive) or attenuate (negative) from inferior to superior upper body locations. Two signals of identical frequency composition will produce a gain of zero for all harmonics.

The transfer function was applied to the harmonics extracted from acceleration data collected during the control, LES fatigue, and CES fatigue walking trials. To provide greater detail about how the upper body modulates gait-related accelerations, gain was calculated for root mean square (RMS) acceleration, and peak power was derived from power spectral analysis of accelerations. For all walking conditions, data from the upper trunk were divided by data from the lower trunk to give a measure of gain for the trunk segment. Similarly, head data were divided by upper trunk data to determine the gain associated with the neck segment.

Signal regularity. The degree of acceleration signal regularity was determined using approximate entropy (ApEn). ApEn is a probability statistic based on the logarithmic likelihood that a sample of data will remain within a tolerance window defined as 20% of the standard deviation (\(r = 0.2\)) in subsequent data increments of one data point (\(m = 1\)) within a serial signal (48, 50). ApEn analysis returns a scalar value that approaches zero with increased signal regularity and approaches two with increased signal irregularity (48, 49). Increased regularity in a signal, or a signal containing a large degree of repeatable pattern features such as a pure low-frequency sine wave, will return a low ApEn value. In contrast, an irregular signal where time series events are unrelated to previous event (such as white noise) will return a high ApEn value.

Signal smoothness. The smoothness of each acceleration signal was calculated using the harmonic ratio (16, 56). Using Fourier transformation, harmonic coefficients were extracted from each acceleration signal, where the fundamental frequency was determined from stride duration. For the VT and AP directions, the sum of the first 20 even harmonics was divided by the sum of the first 20 odd harmonics to give the harmonic ratio. Odd harmonics were divided by even harmonics to calculate the harmonic ratio for the ML direction, so that in all directions higher harmonic ratios reflect increased signal smoothness (38).

Statistical analysis. Typically, MF decreases in a linear manner during fatiguing isometric muscle contraction (9). Therefore, linear regression was used to determine whether the fatigueguing tasks caused a decrease in the MF collected over 0.1-s epochs for each subject. A significant change in the slope of the least squares regression line from zero was taken as an indicator that the tasks induced neuromuscular fatigue.
A two-way ANOVA was applied to basic gait parameter data (gait velocity, stride duration, cadence, and step length) to determine whether differences existed within testing sessions (control vs. fatigue trials) or between testing sessions (control LES fatigue vs. control CES fatigue trials).

One-way ANOVA was applied to RMS and peak power of head accelerations to examine head stability. No significant differences between head accelerations in the control and fatigue walking trials suggested that head stability was not compromised due to fatigue. Furthermore, paired t-tests were used to determine whether differences existed pre- and postfatigue for the accelerometer tilt for the head, upper trunk, and lower trunk.

The effect of fatigue on segmental attenuation was examined using interaction effects (fatigue condition by upper body segment) derived from a two-way ANOVA. Planned contrasts were used to determine whether differences existed pre- and postfatigue for both the neck segment and trunk segment.

The effect of acceleration location (head, upper trunk, lower trunk) on ApEn and harmonic ratio was examined with separate ANOVAs for the VT, AP, and ML directions. Planned repeated contrasts were used to examine relations between adjacent segments in each direction, such as head VT-upper trunk VT, and upper trunk VT-lower trunk VT. All statistical analyses were performed using SAS for Windows (version 8.0), with the level of significance set at 0.05.

RESULTS

Basic gait parameters. Data for gait velocity, stride duration, cadence, and step length for each walking speed condition are presented in Table 1. No differences were detected in basic gait parameters between control and corresponding fatigue trials. Furthermore, no differences were detected in basic gait parameters between control trials and fatigue trials from different testing sessions.

Neuromuscular fatigue. When normalized to body weight, the average force production at 60% MVIC for lumbar extension and cervical extension was 41±9 and 16±5% of body weight, respectively. For all subjects, sustained ES contraction resulted in a time-dependent compression of the power spectrum toward lower frequencies, as illustrated in the representative MF-time histories in Fig. 2, top. Linear regression revealed that, during the initial 30 s of isometric contraction and the following 20 s of contraction, the slope of the MF significantly decreased for both the LES and CES for all subjects (all P < 0.001) (Fig. 2, bottom).

Accelerometer tilt. The maximum mean accelerometer tilt observed during walking across all conditions and trials was 2.31 degrees (upper trunk accelerometer in the AP direction). No significant differences were observed in mean accelerometer tilt relative to global horizontal between control and fatigue walking trials.

Segmental gain: harmonic profile. The harmonic profiles presented in Fig. 3 indicate that patterns of oscillatory gain were similar for the control, CES fatigue, and LES fatigue walking conditions. The most notable harmonic profiles were for the trunk segment in the AP and ML directions, where all but the first harmonic demonstrated a negative gain. Although neck segment gain in the ML direction tended toward zero for all harmonics, the remaining directions and locations had less discernable patterns of harmonic gain.

RMS acceleration. No significant differences were identified between control and fatigue walking trials for RMS accelerations of the head. Fatiguing the CES resulted in a significant change in the gain of the trunk segment, as a significant condition by location interaction was detected in the AP direction [F(7, 49) = 19.63, P < 0.001]. Contrasts revealed that, for the trunk segment in the AP direction, RMS accel-
that, following CES fatigue, ApEn values significantly increased for lower trunk accelerations in the VT direction \( F(1,77) = 4.20, P = 0.033 \) and head accelerations in the AP direction \( F(1,77) = 3.72, P = 0.047 \). The increase in ApEn in the AP direction indicates the structure of head accelerations became more irregular during walking following CES fatigue. Following LES fatigue, ApEn values significantly decreased for upper trunk accelerations in the ML direction \( F(1,77) = 3.98, P = 0.048 \). ApEn data for pre- and postfatigue walking conditions are presented in Fig. 6. Since no significant differences were identified between ApEn of the two control trials, control data presented in the figure is an average of all prefatigue walking trials.

**Signal smoothness.** A significant condition by location interaction was detected for the harmonic ratio in the AP \( F(11,77) = 13.26, P < 0.001 \) and ML \( F(11,77) = 11.69, P = 0.011 \) directions. Planned contrasts revealed that, for the AP direction following LES fatigue, the harmonic ratio for the upper trunk was significantly decreased \( F(1,77) = 6.25, P = 0.014 \). Contrasts revealed that, for the ML direction following LES fatigue, the harmonic ratio for the lower trunk was significantly decreased \( F(1,77) = 4.67, P = 0.033 \). Harmonic ratios for pre- and postfatigue walking conditions are presented in Fig. 6.

**DISCUSSION**

The purpose of this study was to examine the effect of ES muscle fatigue on head stability during walking. In particular, the complementary role that the neck and trunk segments played in modulating accelerations of the head after inducing neuromuscular fatigue of the LES and CES was examined. In the context of this study, optimal stability was defined as the pattern of upper body accelerations exhibited during the self-selected preferred speed control walking conditions. Any differences in terms of acceleration amplitude, frequency content, signal regularity, or smoothness of the head following ES fatigue was interpreted as being indicative of a change in upper body stability.

**Neuromuscular fatigue.** The results of the EMG analysis suggest that neuromuscular fatigue was induced for both the LES and CES during each submaximal isometric contraction. A progressive linear decrease in the MF of EMG activity during sustained isometric contraction is considered a quantitative indicator of neuromuscular fatigue (9, 44, 54). During the initial 30 s and the refatiguing 20-s isometric contraction period of the fatigue protocol, MF of the LES and CES significantly decreased in a time-dependent linear manner. In contrast to the LES, the CES MF for the refatiguing 20-s contraction decreased at a greater rate compared with the initial 30-s contraction. This suggests that the CES was continuing to fatigue during the refatiguing 20-s period, whereas the LES may have been approaching a point of contraction failure during the 20-s contraction (9). The differences in MF observed between the LES and CES may be explained by the load-sharing role of muscles in ES muscle groups. Extension of the lumbar spine is predominantly moderated by longissimus, iliocostalis, and, to a lesser extent, multifidus groups. In contrast, multiple muscles have the potential to extend the cervical spine, such as the trapezius, longissimus, iliocostalis, splenius, and spinalis groups among others. Although variable load...
sharing has been illustrated for LES muscle to prolong the onset of fatigue and maintain force output (33, 37, 60), the CES may be even more reliant on recruiting synergistic muscles to prolong fatigue if the CES contains a greater proportion of type II fibers (14, 36).

An intermittent fatigue protocol, such as the one employed in the present study, has been suggested to induce LES muscular fatigue without compromising force output at submaximal loads (2). Therefore, differences in upper body segmental attenuation characteristics in the present study are more likely the result of altered patterns of neuromuscular activity rather than changes in force-generating properties. Given that accelerometer tilt was not different following LES and CES fatigue, it is also likely that any fatigue-induced differences in the present study were not influenced by changes in upper body posture.

**Head stability following fatigue.** The results of the present study indicate that walking speed did not differ between any condition. This suggests that not only was the preferred walking speed consistent between control walking trials but that the level of induced LES and CES fatigue did not affect locomotor system’s ability to repeat the preferred walking speed. One characteristic of walking often observed with acute and chronic postural deficits is that walking speed is decreased, with a view of minimizing potentially perturbing segmental oscillations, thus enhancing stability. Compared with healthy adults, the elderly (5, 45), Parkinson’s disease (40), and cerebral and spinal palsy (7, 21) individuals often exhibit a more cautious walking strategy, which is characterized by adaptations in walking speed and segmental interactions that assist in minimizing the amplitude and variability of head motion. Studying
postural control in populations with musculoskeletal degeneration, or any type of coordination deficit, can be problematic due to the cause-and-effect relationship of walking speed and stability. It is often difficult to determine whether adaptations of the neuromuscular system cause reductions in gait speed (22) or whether the slower speed is due to self-imposed constraints to provide a greater perception of stability (20). The absence of speed differences between walking conditions in the present study suggests that differences observed for head stability and segmental attenuation are more likely to reflect the singular effect of fatigue rather than velocity-dependent changes in gait patterns.

The first hypothesis of this study was that CES fatigue would result in a diminished capacity to maintain head stability due to an altered attenuation capability of the neck segment. This was in part confirmed as the regularity of head accelerations, and therefore head stability was different between pre- and postfatigue walking conditions. However, the attenuation properties of the trunk, and not the neck, were affected following CES fatigue. The second hypothesis of this study was that LES fatigue would not affect head stability as the unfatigued, intervening neck segment would compensate for the diminished attenuation properties of the trunk segment. This was confirmed in the plane of progression; however, in the ML direction the dynamics of the trunk segment were different between pre- and postfatigue walking conditions.

Head accelerations were unaffected by fatigue in the VT and ML directions, indicating the postural system was able to preserve head stability in the frontal plane despite LES or CES fatigue. Similarly, head stability was preserved in the AP direction following LES fatigue. However, the results of the signal regularity analysis indicate that, following CES fatigue, AP head accelerations were less regular. This change in head AP regularity was observed despite no postfatigue differences in the regularity of lower or upper trunk accelerations in the AP direction. Potentially, CES fatigue perturbed sensorimotor feedback mechanisms of the neck segment that contribute to stabilizing head motion during walking. Neuromuscular fatigue can affect the afferent flow from a multitude of peripheral receptors to the CNS (52, 59, 62), which in turn may alter neural drive and motor output of the same region of the body (23, 47). This point is particularly relevant in regard to regulating upper body movement, as active spinal stability is primarily controlled by muscle recruitment patterns, active muscle stiffness, and reflex response, all of which are modifiable with fatigue (18). It has also been speculated that postural orientation in the sagittal plane may be altered following neck fatigue due to changes in afferent outflow to areas of the CNS that are responsible for building spatial reference frames for body segments (55). With the majority of dependent variables not affected following the isometric fatiguing contractions, the postural system was in general able to preserve head stability despite the presence of upper body fatigue. It is perhaps not surprising that the only difference detected in head accelerations following fatigue was for signal regularity. ApEn has been reported to be a highly sensitive analysis in detecting differences in upper body accelerations during walking on a level surface compared with measures of signal amplitude, frequency content, or smoothness (25). The question now arises: How did the trunk and neck segments modulate gait-related oscillations to contribute to stabilizing the head following fatigue of ES muscle groups?

**Segmental gain following fatigue.** The general profile of trunk and neck segment harmonic gain was similar for the control, CES fatigue, and LES fatigue walking trials (Fig. 3). However, fatiguing ES at one level of the upper body altered the gain of the adjacent upper body segment in the AP direction. During the CES fatigue condition, there was greater attenuation for the trunk segment in the AP direction for RMS and peak power compared with the control condition. Although it was not possible to determine the mechanism underlying the observed compensatory motor strategy, since multiple muscle systems are responsible for maintaining posture, it is likely that certain sensorimotor events occurred. It is probable that the nature of fatigue altered afferent flow from sensory receptors of the fatigued muscle to the CNS, which in turn elicited an adaptive motor response (3, 19). Recruitment of unfatigued synergistic muscle groups have been observed in knee extensors during submaximal fatiguing contractions (1, 30, 53) and intrinsic finger muscles during isometric contraction tasks (8, 66). These studies demonstrate that muscular fatigue need not compromise the performance of simple movement tasks, as altered neural drive can generate alternative motor strategies to
perform the task. Although supraspinal factors are likely to be involved, it cannot be ruled out that spinal reflexes may have contributed to the observed motor patterns. Although not yet demonstrated in humans, animal models have revealed that neuromuscular fatigue increases fusimotor activity not only in the contracting muscle but in synergistic muscles as well (35).

In contrast to the AP and VT directions, fatiguing the LES resulted in a greater gain in peak power for the trunk segment in the ML direction and an increase in signal regularity for acceleration of the upper trunk. A similar profile of upper body attenuation has been observed in previous research, where head stability was preserved despite peak power and regularity of upper body accelerations increasing from the lower to upper trunk during self-selected slow walking speeds (25). Differences in the trunk segment dynamics due to LES fatigue indicate that the effects of perturbing spinal extensors were not confined to movement characteristics in the sagittal plane. Since the postural system must coordinate segmental interactions in each direction, it is possible there is a trade-off in trunk attenuation in the ML direction to enhance head stability in the direction that was perturbed. Regardless of the motor strategy employed, following LES, fatigue accelerations were more regular for the upper trunk in the ML direction and smoother in the AP direction. A tightly controlled upper trunk is beneficial to head stability, since it creates a stable platform for the neck and head to oscillate on, a feature observed during normal walking (4, 25, 65). Furthermore, the differences observed for gain and signal regularity of the trunk segment, combined with increased smoothness of lower trunk accelerations in the ML direction following LES fatigue, reinforce the notion that the postural system attempts to control gait-related oscillations before reaching the head by coordinating trunk movement in each direction (25).

Limitations. Since this is the first known study to investigate the effect of LES and CES fatigue on head stability during walking, several experimental features can potentially be enhanced in future work. The methods employed in the present study were noninvasive; therefore, the number of subjects recruited in future investigations could be increased. In regard to the ES fatigue, it is of interest to determine the reliability of the fatiguing protocol. Although similar intermittent isometric LES contractions have been reported, the ability to replicate both the LES and CES fatigue protocol employed in the present study has yet to be determined. Furthermore, the transfer of fatigue from the isometric contraction task to the walking task is of interest, since the extent to which each subject recovered over the duration of each walking trial is unknown. No subjective measures of fatigue were included in the present study; however, this may have provided further information pertaining to each participant’s level of perceived exertion. Although the relationship between subjective and objective measures of fatigue for the CES are suggested to be moderately correlated (57), subjective measures of LES fatigue have a good correlation with features of the EMG signal (10). Although changes in the attenuation properties of trunk and neck segments in general assisted in preserving head stability following ES fatigue, inducing greater levels of fatigue may have resulted in more pronounced changes in head stability.

In conclusion, the findings of this study indicate that ES fatigue differentially affected upper body segmental attenuation during walking according to the level of the upper body that was fatigued and the direction that oscillations were attenuated. Head stability was preserved in all directions following LES fatigue and in the VT and ML directions following CES fatigue. However, accelerations of the head were less regular after CES fatigue was induced. Fatiguing ES at one level of the upper body altered the attenuation characteristics of the adjacent upper body segment in the AP direction, presumably to minimize the impact of gait-related accelerations on head movement. The effects of fatiguing spinal extensors were not confined to changes in segmental gain in the sagittal plane, since LES fatigue resulted in greater gain for the trunk segment in the ML direction. Signal regularity and smoothness analysis suggested that, following LES fatigue, the postural system reorganized segmental motion to enhance stability of the upper trunk, thus providing a stable platform for the neck and head to move on. Overall, the findings in this study suggest that head stability has a high priority during walking and that the CNS reorganizes segmental interactions to facilitate head stability in the presence of ES muscle fatigue.

GRANTS

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