Frequencies in Center of Pressure Time Series above 1Hz during Quiet Upright Stance Reflect the Use of a Hip Strategy

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Frequencies in Center of Pressure Time Series above 1Hz during Quiet Upright Stance Reflect the Use of a Hip Strategy

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Upright stance is primarily maintained through the use of ankle and hip strategies, involving respective rotations about the two joints. Choice of and coordination between these two strategies is regulated by the central nervous systems (CNS). Postural control mechanisms used to maintain upright stance are often assessed (or inferred) using a spectral decomposition of center of pressure (COP) time series. Existing evidence, however, is contradictory in some respects. Specifically, there is conflicting evidence whether COP frequencies above 1.0 Hz reflect CNS function or use of a hip strategy. In this study, fatigue was induced to the lumbar extensors and shoulder flexors through intermittent sub-maximal (60% MVC) isotonic exercises, and used to 'perturb' postural control. Use of hip strategies was assessed by entropy methods, which indicates the level of 'complexity'. Coordination between ankle and hip strategies was analyzed using a coherence method. Both power spectrum and Tsallis and entropies showed that the medio-lateral (ML) hip strategy was impaired by lumbar extensor fatigue. Coherence (i.e. coordination) between ankle and hip motions was reduced following fatigue in both muscles groups. Integrated with earlier evidence, these results indicate that, during quiet upright stance, COP frequencies above 1.0 Hz correspond to the hip strategy rather than reflecting CNS function in postural control.

1. INTRODUCTION

Occupational falls have drawn increased interests of researchers in recent decades (e.g. Kemmlert and Lundholm 2001; Layne and Pollack 2004). It was reported (Agnew et al. 1993) that between 1980 and 1986, there were over 4,000 fatalities that resulted from falls. More recent evidence (BLS, 2006) indicated that falls accounted for 14% of all occupational fatalities. Slipping, tripping, and loss of balance are major contributors to occupational falls (Courtney et al. 2001). The potential for a slip or trip is highly associated with the capabilities of the postural control system (e.g. peripheral and central systems), including both physical and cognitive aspects (Redfern et al. 2001). As such, investigating underlying factors involved in postural control is necessary in order to help prevent occupational falls.

Postural control in an upright stance is maintained primarily through ankle and hip strategies (e.g. Runge et al. 1999; Peterka, 2002; Creath et al. 2005), in which equilibrium is maintained by movements around the ankle and hip joints. It was suggested by Kuo (1995) that the choice of ankle or hip strategy depends on both the postural goal and environmental constraints, and is regulated by central selection.

Typically, upright stance stability can be quantified through processing displacements of center of pressure (COP) (e.g. Collins and De Luca 1993), which are derived from ground reaction forces. Thurner et al. (2000) suggested that the spectral content of COP >1.0 Hz reflects the contribution of the central nervous system (CNS) to posture control, based on spectral changes in COP with mental disorders (e.g. autism). However there also exist other factors which could account for their experimental results, for instance the decline of peripheral systems performance (e.g. time-delay of postural feedback control) due to the mental disorders. As these changes might affect underlying postural control mechanisms, there is thus some question about the validity of their conclusions. More recent evidence (Creath et al. 2005), suggests that the ankle strategy is related to frequencies below 1.0 Hz, and the hip strategy to frequencies above 1.0 Hz. Similarly, Zhang et al. (2007) demonstrated that above and below 1.0 Hz, the hip and ankle strategies respectively dominate sway. From this, it is apparent that there is a lack of consensus regarding what postural control mechanisms, during upright stance, are responsible for the COP signal content at frequencies above 1.0 Hz.

This research sought to clarify this issue and identify the postural control mechanisms related to the sway frequencies above 1.0 Hz. To this aim, lumbar and shoulder localized muscle fatigue (LMF), which has been demonstrated to result in central and peripheral changes (e.g. Benwell et al. 2006; Simo et al. 1999), were introduced in order to compromise the hip strategy and/or CNS postural control mechanisms. Effects of fatigue were assessed using two approaches. The first was based on COP signal entropy. Entropy methods were used to assess the associated changes of the hip strategy in the presence of lumbar LMF. Previous research (e.g. Rosso et al. 2003) has implied that entropy can be used to evaluate the complexity (order/disorder) of physiology signals. A lower complexity implies a smaller degree of disorder, which in turn suggests a loss of certain regulated information by underlying control mechanisms. For example, Hong et al. (2007) applied entropy methods to quantify the variability of postural sway following withdrawal of visual feedback and induced plantar desensitization. We hence expected a decreased disorder (decreased entropy) in COP data > 1.0 Hz following induction of LMF. Both power spectrum and Tsallis wavelet entropy methods were used to assess changes to underlying postural control mechanisms (e.g. CNS or the hip strategy). A second method involved assessing coherence between ankle and hip kinematics. Such coherence methods have been frequently applied to analyze the coupling between external stimulus (e.g. visual stimulus) and upright stance stability (e.g. Prioli et al. 2005) as well as the coordination between leg and trunk segments during upright stance (e.g. Zhang et al. 2007). In the present work, the coherence method was applied to analyze how coordination between ankle and hip motions was affected by fatigue. Since such coordination is regulated centrally (i.e. by the CNS), fatigue-related changes in coherence would reflect central changes in postural control.

2. METHODS

2.1. Participants

Sixteen young adults participated in this experiment, with an equal number of male and female. All participants completed an informed consent procedure approved by the Virginia Tech Institute Review Board, and had no self-reported injuries, illness, musculoskeletal disorders or falls in the year prior to the study.

2.2. Experimental Procedures

Data analyzed here were obtained from a prior experiment (Singh et al. 2005), in which trials of quiet, upright stance were conducted both prior to and after fatiguing exercises were conducted. Segmental kinematics were monitored using a set of 13 reflective markers attached at the ankle, knee, hip, shoulder, elbow, wrist, chin, and temple. Marker positions were tracked using a commercial system (Vicon, 460, Vicon Peak Motion Systems Inc., Lake Forest, CA). COP time series were derived from ground reaction forces obtained using a force platform (AMTI OR6-7-1000, Watertown, MA). During all trials, participants were instructed to stand without shoes on the force platform, as still as possible, with their feet together, arms by their sides, head straight and eyes closed for 75 seconds. For pre-fatigue, three replications were conducted with one minute in between each.

LMF was induced separately in the torso extensors and unilateral (dominant side) shoulder flexors on different days with at least 24 hours rest to minimize residual fatigue. Participants first performed warm up exercises, consisting of two sets of 10 repetitions each: stooping for the torso and arm raising for the shoulder. Subsequently, five replications of an isokinetic maximum voluntary contractions (MVCs) were conducted at a speed of 60°/sec using a commercial dynamometer (Biodex System 3 Pro, Biodex Medical Systems Inc., New York, USA), with at least one minute between each. The range of motion for MVCs was 45°, from 45° flexion to the upright position for the torso, and from 0° to 45° flexion for the shoulder.

Participants were then asked to perform sub-maximal (60% MVC) isotonic exercise at 12 repetitions/min using the dynamometer to induce LMF in the torso extensors and shoulder flexors with the same range of motion as that used in MVC. Participants were instructed to start the exertion at the sound of a computer-generated tone, and to reach the end range of motion by the second beep. These exercises were continued until participants were not able to finish three consecutive isotonic efforts over the full range of motion. From this, participants were fatigued to a level of ~60% of their pre-fatigue isokinetic capacity. Following LMF induction, standing trials were conducted, the first within 45 sec. After exercise termination, the remaining post-fatigue trials (11 trials) were performed over 15 minutes.

Triaxial ground reaction forces and moments were sampled at 100 Hz and the raw signal was low-pass filtered (Butterworth, 5 Hz cut-off frequency, 4thorder, zero lag) and transformed to obtain COP data in the antero-posterior (AP) and medial-lateral (ML) directions. Marker locations were sampled at 20 Hz, and were low-pass filtered (Butterworth, 5 Hz cut-off frequency, 4th order, zero lag) to obtain the ankle, knee, hip, and shoulder positions. For both COP and marker time series, the first 10 sec were removed in order to eliminate postural changes that might occur while participants closed their eyes, and the last 5 sec were removed to eliminate any potential effects of anticipation at the end of the trial, resulting in a duration of 60 sec. The three pre-fatigue trials and the first three post-fatigue trials were included for analysis, the latter in order to avoid substantial recovery effects in subsequent trials.

2.3. Tsallis and Power Spectrum Entropy Methods

Wavelet transforms are often used to transform signals from the time domain to the frequency domain. A generic wavelet transform is given by:

$$WT(a,b) = |a|^{-1/2} \int f(t)\varphi\left(\frac{t-b}{a}\right)dt \qquad (1)$$

where f(t) is the signal value at time t, and $\phi_{a,b}(t)$ is the wavelet function (the Haar wavelet function was applied here); *b* corresponds to the translation parameter (time) and *a* is a scale parameter (frequency); WT(*a, b*) is the wavelet coefficient. The power spectrum of the signal at frequency a and location b is defined as:

$$\mathrm{PS}(a,b) = |\mathrm{WT}(a,b)|^2 \qquad (2)$$

The total energy under scale (frequency) *a* over time *T* is:

$$E_a = \int_0^T WT(a,b)^2\, dt \qquad (3)$$

Subsequently, the total energy *E* is given by:

$$E = \sum_{a=1}^{W} E_a \qquad (4)$$

Tsallis (1988) proposed a generalized Boltzmann-Gibbs entropic measure as below:

$$S_T \equiv k\left(1-\sum_{a=1}^{W} p_a\right)(q-1)^{-1} \qquad (5)$$

where (for simplicity), given k equal to 1, W is the number of microscopic configurations whose probability belongs to $\{p_a\}$. $q \in R$ specifies the particular statistics, with 1.5 used here. The relative wavelet energy p_a is formulated as the percent of energy in configuration versus the total energy:

$$p_a = \frac{E_a}{E} \qquad (6)$$

The power spectrum entropy method is given by:

$$S_P \equiv -\sum_{a=1}^{W} p_a \log_2(p_a) \qquad (7)$$

When calculating Tsallis and power spectrum wavelet entropy, COP content in the 1.0-2.1 Hz band was chosen to analyze the hip strategy (note, there is limited power in the COP signal above this band).

2.4. Coherence

Coherence is defined by:

$$C_{AH}(f) = \frac{|s_{AH}(f)|^2}{s_{AA}(f)s_{HH}(f)} \qquad (8)$$

where $s_{AA}(f)$ is the autospectral density function of ankle angles, $s_{HH}(f)$is the autospectral density function of hip angles, and$s_{AH}(f)$is the cross-spectral density function of ankle and hip angles.

A confidence limit for coherence at level α was determined using (Lowery et al. 2007):

$$C_{AH}^{limit}(f) = 1-(1-\alpha)^{\frac{1}{M-1}} \qquad (9)$$

where M is the total number of overlapping sections as a result of the fixed length of a window sliding through the data, and α = 0.95. It was observed that for M=4.73, where the sliding window length is 256, the confidence limit is 0.5547 represented by the thick solid line shown in Figure 1. Ankle and hip angle energy concentrated below 0.02 Hz. It was also observed that above 1.0 Hz, coherence was approximately equal to 1.0. As a result, below 1.0 Hz, the mean of coherences, the values of which were larger than the confidence limit, was chosen as the final coherence.

Entropies (power spectrum and Tsallis) and coherence were determined for the three pre-fatigue and three post-fatigue trials. Effects of LMF on these measures were assessed separately using two-way mixed-factor ANOVA, with gender and LMF as independent variables. Statistical significance was determined when p< 0.05.

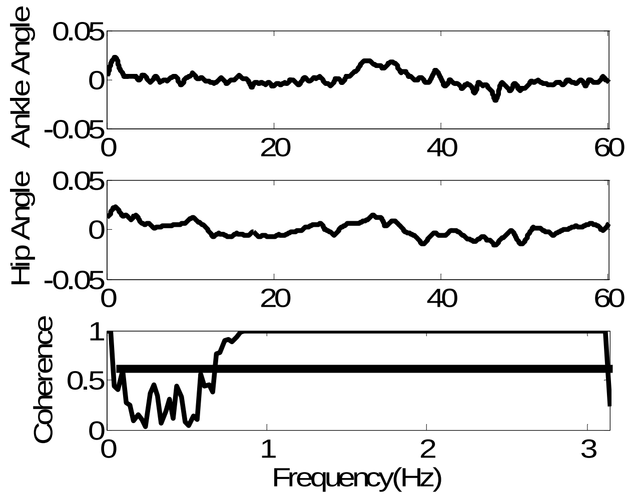

Figure 1: Representative hip and ankle angles (radians), and coherence for a post-fatigue trial (lumbar extensors).

3. RESULTS

3.1. Power Spectrum and Tsallis Entropies

Following lumbar LMF, ML power spectrum entropy above 1.0 Hz became significantly smaller (p = 0.046), but was unchanged (p = 0.220) in the AP direction (Figure2). Following shoulder LMF, power spectrum entropies above 1.0 Hz were unchanged in both the ML (p = 0.316) and AP (p = 0.553) directions. Results for Tsallis entropy were similar, in that ML Tsallis entropies above 1.0 Hz decreased significantly (p = 0.048) after lumbar LMF, and were unchanged (p = 0.221) in the AP direction (Figure 2). For other frequencies bands (e.g. 0.5-1.0 Hz), and following shoulder LMF, no significant changes in power spectrum and Tsallis entropy changes were observed.

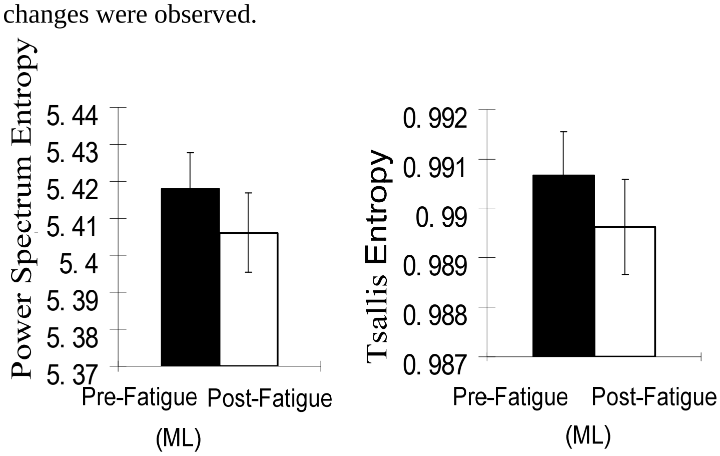

Figure 2: ML power spectrum and Tsallis entropy pre and post torso LMF.

Mean coherence was significantly reduced (Figure 3) following both lumbar (p < 0.001) and shoulder (p = 0.0014) LMF.

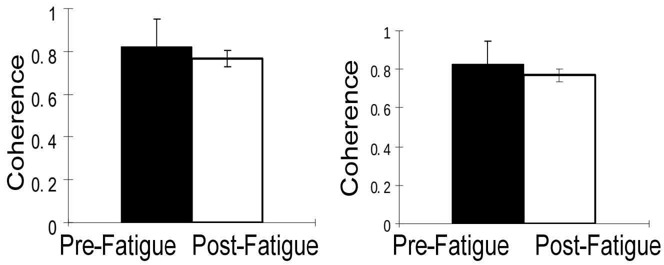

Figure 3: Coherence pre and post torso (left) and shoulder (right) LMF.

4. DISCUSSION

Ankle and hip coordination was reduced in the presence of lumbar extensor and shoulder flexor LMF. This reduced coordination might be because of fatigue-induced changes to peripheral systems or central systems performance. Several muscle groups are employed in the hip strategy, including the trunk flexors/extensors, gastrocnemius, tibialis anterior, and ankle dorsi/plantar flexors (Rungea et al. 1999). As such, major muscle groups used to maintain the hip strategy were not impaired due to shoulder LMF in our experiment. Further, only participants with healthy vestibular functions were involved, and the visual system was removed through eye closure. It is therefore concluded that the significantly reduced coordination between ankle and hip following shoulder LMF is most likely attributed to a change in central (CNS) control, rather than effects on afferent or efferent systems. Theoretically, Kuo (1995) has already pointed out that the CNS is responsible for selecting the ankle and/or hip strategy to maintain postural stability. Hence, ankle-hip coordination is closely regulated by CNS function. Some experiments (e.g. Brasil-Neto et al. 1993; Gandevia 2001; Benwell et al. 2006; Weir et al. 2006) have also shown that central function is altered following LMF, specifically with reduced descending drive resulting in reduced muscle power. In general, it is believed that central fatigue results from a depression of activity in the sensorimotor cortex (Benwell et al. 2006), which is highly correlated with the reduced muscle force output (Mima et al. 2001).

In light of the proceeding discussion, changes in coordination between ankle and hip following shoulder LMF, as indicated by coherence, are attributed to central fatigue. Entropy results, however, showed both power spectrum and Tsallis entropies for COP frequencies above 1.0 Hz did not show significant changes following shoulder LMF. In contrast to the suggestion by Thurner et al. (2000), these results indicate that the CNS was not significantly compromised, and implies that frequencies above 1.0 Hz is not related to CNS function.

On the other hand, both power spectrum and Tsallis entropies significantly decreased following lumbar extensor fatigue, indicating that the hip strategy was impaired. From this, it can be concluded that COP frequencies above 1.0 Hz do not correspond to CNS function, but reflect the hip strategy. This result is justified because the primary muscle groups (e.g. trunk extensors) used to maintain the hip strategy were fatigued in experimental trials, which thereby lead to the changed hip strategy. Other research (e.g. Simo et al. 1999) has also shown that torso LMF compromises the ability to sense a change in lumbar position, which may be an underlying source of the deteriorated hip strategy. Collectively, our results support that of Zhang et al (2007) and Creath et al. (2005), who indicated that COP frequencies above 1.0 Hz are related to the hip strategy during maintenance of quiet upright stance.

A potential limitation of this study should be noted. Brain functions and spinal cord activities were not directly measured (e.g. with transcranial magnetic stimulation or functional imaging techniques). Such measures would provide more direct evidence for our somewhat indirect inferences regarding CNS involvement in postural. Future work is clearly needed to address this weakness.

5. CONCLUSIONS

From this study, it was concluded that, in the context of quiet upright stance, the spectral content of COP time series >1.0 Hz correspond to the use of hip strategy rather than CNS function. This result may be of value in future studies, where spectral analysis of COP data is used to investigate postural control stability and mechanisms. It was also shown that shoulder LMF leads to central changes characterized by a deterioration in ankle-hip coordination. This result suggests the potential for a loss of balance in occupational situations where shoulder fatigue is present.

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