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Bracing of the trunk and neck has a differential effect on head control during gait

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Morrison S, Russell DM, Kelleran K, Walker ML. Bracing of the trunk and neck has a differential effect on head control during gait. *J Neurophysiol* 114: 1773–1783, 2015. First published July 15, 2015; doi:10.1152/jn.00059.2015.—During gait, the trunk and neck are believed to play an important role in dissipating the transmission of forces from the ground to the head. This attenuation process is important to ensure head control is maintained. The aim of the present study was to assess the impact of externally restricting the motion of the trunk and/or neck segments on acceleration patterns of the upper body and head and related trunk muscle activity. Twelve healthy adults performed three walking trials on a flat, straight 65-m walkway, under four different bracing conditions: *1*) control-no brace; *2*) neck-braced; *3*) trunk-braced; and *4*) neck-trunk braced. Three-dimensional acceleration from the head, neck (C_7) and lower trunk (L_3) were collected, as was muscle activity from trunk. Results revealed that, when the neck and/or trunk were singularly braced, an overall decrease in the ability of the trunk to attenuate gait-related oscillations was observed, which led to increases in the amplitude of vertical acceleration for all segments. However, when the trunk and neck were braced together, acceleration amplitude across all segments decreased in line with increased attenuation from the neck to the head. Bracing was also reflected by increased activity in erector spinae, decreased abdominal muscle activity and lower trunk muscle coactivation. Overall, it would appear that the neuromuscular system of young, healthy individuals was able to maintain a consistent pattern of head acceleration, irrespective of the level of bracing, and that priority was placed over the control of vertical head accelerations during these gait tasks.

gait; bracing; acceleration; control; muscle

AN ONGOING CONTROL ISSUE FACED by the human neuromuscular system when walking is to ensure that the impact of gait-related oscillations on head motion is minimized. The rationale behind this control requirement is that maintaining head control is critical in ensuring the visual and vestibular systems function accurately during locomotion activities (Carlsen et al. 2005; Pozzo et al. 1990, 1991). One means by which the system actively ensures head stability during walking is through the trunk and the neck (Menz et al. 2003a, 2003b). Previous studies have reported that both of these segments play an important role in damping gait-related oscillations to ensure head control is maintained (Kavanagh et al. 2004, 2006a; Menz et al. 2003b). Given the significant contribution of vestibular, ocular and cervical reflexes to maintaining head control during locomotion (Cappozzo 1981; Carlsen et al. 2005; Leah et al. 2005; Mazza et al. 2008), stabilizing the head appears to be a critical component of gait (Berthoz and Pozzo 1988; Pozzo et al. 1990, 1991). Consequently, any factor which alters the intrinsic properties of the spine could lead to systematic changes in walking dynamics.

Healthy aging leads to changes in the role of the trunk and neck in stabilizing the head during gait (Kavanagh et al. 2004, 2005a). This may arise from the degeneration of the lumbar intervertebral disks in the elderly, increasing trunk stiffness (Boos et al. 2002; Videman et al. 2014), or it may represent an increase in trunk stiffness associated with a more cautious gait, as has been detected in individuals with low back pain (van den Hoorn et al. 2012). Any decline in capacity of the trunk-neck axis to effectively dampen oscillations related to walking is likely to negatively impact gait (Menz et al. 2003a, 2003b). However, age-related changes in other physiological functions often contribute to declines observed in walking performance. For example, general declines in muscle strength, flexibility, sensation, proprioception, and reaction times have all been linked with changes in walking function, which may be manifested by slower walking speed and cadence, shorter stride length, and greater stride width for healthy older adults (Kobsar et al. 2014; Menz et al. 2007; Oberg et al. 1993; Patterson et al. 2012). Hence, studying the influence of trunk and neck stiffness on the ability of these segments to dissipate the transmission of forces from the ground to the head in older adults cannot be readily separated from the other changes that occur with aging.

An alternative approach to understand the interaction between the trunk and head during gait and balance activities has been to assess what happens when the motion of the trunk and/or neck are experimentally restricted in young and/or older adults (Cholewicki et al. 1997, 2010; van der Burg et al. 2007; Wu et al. 2014). While not specifically addressing walking ability, Cholewicki and colleagues have undertaken several pivotal studies assessing the impact bracing the trunk has on neuromuscular control and/or balance ability in healthy persons and those with lower back pain (Cholewicki 2004; Cholewicki et al. 1997, 2003, 2007, 2010). They have reported that, under normal conditions, cocontraction of the trunk muscles provides stability to the spine (Cholewicki et al. 1997). Following on, when stiffness is increased artificially through using an external brace, the pattern of muscle activity about the trunk is affected in response to rapid trunk force release task (Cholewicki et al. 2010). With regards to gait, a recent study assessed the effect increasing trunk stiffness (either by voluntary cocontraction or by wearing an orthopedic brace) had on the pattern of coordination between the thorax, pelvis, and leg segments when walking on a treadmill at different speeds (Wu et al.

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2014). The results revealed that increasing trunk stiffness led to transient changes in coordination between these segments, similar to that found in patients with lower back pain. Similarly, van der Burg et al. (2007) examined the effects of externally increasing trunk stiffness (using an orthopedic corset) on balance recovery after tripping in young healthy adults. Their main finding was that, despite initial changes in trunk inclination, the overall effect of the corset was negligible, and that persons were able to compensate for any restrictions imposed by artificially stiffening the trunk when externally tripped. What is apparent from these studies is that individuals are often able to compensate for any external restrictions on trunk motion introduced by bracing to ensure the movement is performed. However, it should be noted that these studies were not designed to address what effect external restrictions would have on head motion. Furthermore, it has to be determined what effect restricting the trunk and/or neck motion has on the head. While it is likely that the neuromuscular system is able to compensate for constraints imposed by externally increasing trunk stiffness (i.e., through bracing) during gait, the issue of how this is achieved still needs to be addressed.

The aim of the present study is to assess the impact of restricting the motion of the trunk and/or neck (through external bracing) on the acceleration patterns for the lower trunk, neck and head and trunk muscle activity during overground walking. We predict that constraining the trunk and/or neck will negatively impact the intrinsic ability of these segments to compensate for gait-related oscillations. This will be reflected by a decline in the ability of the trunk and neck to attenuate gait-related oscillations, ultimately resulting in an increase in the amplitude and complexity of head acceleration during overground walking. Additionally, muscle activation about the trunk will be altered as a result of bracing, with changes in electromyographic (EMG) activity and decreased cocontraction reflecting compensatory adjustments generated to maintain trunk control during walking with the trunk/neck braced.

METHODS AND MATERIALS

Participants

Twelve healthy adults (2 men, 10 women; average age 21.25 ± 1.95 yr; average height 1.68 ± 0.9 m; average weight 66.4 ± 2.65 kg) participated in this study. All participants were physically active, had normal or corrected-to-normal vision, and reported no known neurological/cognitive disorders, or history of neuromuscular injury that could influence performance. Participants provided informed consent prior to inclusion, and all procedures were approved by the University Internal Review Board.

Experimental Design

Participants were required to perform three straight-line walking trials along a 65-m level walkway under four different bracing conditions at their preferred, self-selected walking speed. All persons wore their preferred footwear (running shoes) for all walking trials. During all walking trials, individuals were instructed to look straight ahead. The different bracing conditions were as follows: *1*) control (no bracing); *2*) trunk braced only; *3*) neck braced only; and *4*) trunk and neck braced. The order in which the different bracing conditions were performed was counterbalanced with the view of avoiding order effects.

Each person's preferred walking speed (PWS) was determined using a 20-ft. straight GAITRite pressure-sensitive walking surface, which was positioned in the midpoint of the 65-m walkway (CIR Systems, Havertown, PA). Two practice (unbraced) trials were performed to determine each person's PWS (sample frequency 120 Hz). To minimize the influence of speed on the results, walking trials outside of the ±10% allotted range for the designated PWS, based on the no-bracing condition, were rejected and repeated. Following data collection, analysis was performed and confirmed there were no significant changes in gait speed across the four bracing conditions (control = 1.33 ± 0.08 m/s; neck braced only = 1.32 ± 0.13 m/s; trunk braced only = 1.31 ± 0.11 m/s; trunk and neck braced = 1.33 ± 0.10 m/s; $F_{3,33} = 0.31$, $P = 0.82$).

For all walking conditions, three lightweight wireless triaxial accelerometers (Delsys, Boston, MA) were attached to each subject to measure three-dimensional accelerations during walking. Accelerometers were attached over the head (vertex, with a firm fitting elastic headband), the neck (C_7 spinous process), and lower trunk (L_3 spinous process). These devices were attached to the neck and trunk using rigid sports tape. Bilateral surface EMG measures were also attained during the walking tasks using the same Delsys system. The selected muscles were as follows: *1*) rectus abdominus (RA; 4 cm lateral from the umbillicus); *2*) external oblique (EO; directly below most inferior point of costal margin, on a line to the opposite pubic tubercle); *3*) thoracic erector spinae (ES; over palpable bulge of muscle, ~3 cm lateral of midline, with lower electrode at level of L_1); and *4*) middle portion of trapezius (TZ; over muscle belly). Location of the electrodes for the RA, EO, TZ, and ES muscles were according to specifications outlined in previous studies (Anders et al. 2007; Swinnen et al. 2012). All EMG and accelerometer data were sampled at 1,000 Hz and synchronized using a 64-channel Delsys Trigno data collection system (Delsys, Boston, MA).

Two separate braces were used to brace the trunk and neck regions. For the trunk region, a Kendrick Extrication Device (K.E.D., Ferno-Washington, Wilmington, OH) covered the entire trunk region. For the neck, a cervical collar (StifNeck, Laerdal Medical, Wappingers Falls, NY) was used. The braces used were standard equipment in the field of emergency medicine. Each brace was able to be independently adjusted to fit each person. To ensure consistency across subjects, the same research assistant fitted this device to each person. Figure 1 provides an illustration of a subject walking with the neck and trunk braces attached.

Data Analysis

Prior to analysis, the accelerometer data were filtered by a second-order Butterworth low-pass digital filter with a cut-off frequency of 20 Hz. EMG data were full-wave rectified and band-pass filtered at 10–500 Hz. A tilt correction was also applied to the acceleration data prior to analysis, to account for deviations in accelerometer axes from the global vertical and horizontal axes while attached to the subject's body (Moe-Nilssen 1998a, 1998b). 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. The degree of accelerometer tilt was established from accelerometer data during each walking trial. The filtering, tilt correction procedure and all subsequent analyses were performed using software developed in Matlab version 7.0 (Mathworks R14).

Acceleration Signal Amplitude

The amplitude of head, neck, and trunk acceleration in the vertical, mediolateral (ML) and anterior-posterior (AP) directions were examined from root mean square (RMS) accelerations (*g*). Power spectral analysis was performed on all acceleration data using Welch's averaged, modified periodogram method (window size of 512 data points). For the acceleration data, the power from the dominant frequency peak (peak power, g^2), and frequency at which the peak power occurred (Hz) were calculated for each trial.

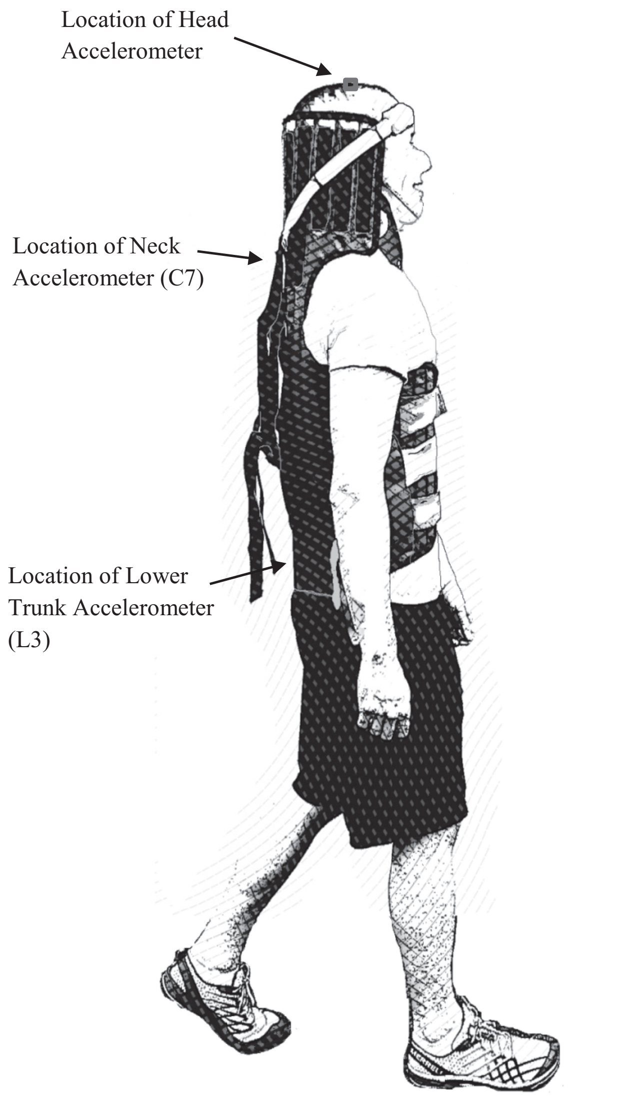

Fig. 1. Illustration of a subject walking with full neck and trunk brace attached. The approximate location of the accelerometers for the head, neck (C_7) and lower trunk (L_3) is also noted in this figure.

Acceleration Signal Regularity

An indication of the pattern of regularity of the acceleration signals was determined using approximate entropy (ApEn). ApEn provides a measure of signal regularity (Pincus 1991, 1995), with this analysis returning a single value for the signal within the range of 0–2. Higher values indicate increased irregularity/complexity in the signal, while lower values (closer to 0) represent greater regularity or structure.

Segmental Gain

An estimation of the degree of attenuation or gain between the lower trunk-neck and neck-head combinations was determined by applying a transfer function to the amplitude (RMS) and regularity (ApEn) of acceleration data. The transfer function provides an estimation as to whether there was an overall gain (i.e., positive value) or attenuation (i.e., negative value) of the acceleration signal from the lower trunk to the head during the different walking conditions (Hamill et al. 1995; James et al. 2014; Kavanagh et al. 2006b). For this analysis, data from the neck were divided by data from the lower trunk to give a measure of gain or attenuation for the trunk segment. Similarly, head values (RMS and ApEn) were divided by those for the neck to determine the level of gain/attenuation between these two segments. Units for attenuation or gain are decibels (dB).

Muscle Activation Measures

EMG amplitude. An indication of the degree of activity for the selected trunk muscles was determined by calculating the RMS of each EMG signal (bin size 100 ms).

Coactivation index. Changes in the mean RMS values for each muscle were used to give an estimate of the degree of muscle coactivation via the calculation of the coactivation index (CI) (Hortobagyi and DeVita 2000).

$$CI = \frac{Rectus\ Abdominus(L,R) + Transverse\ Abdominus(L,R)}{Erector\ Spinae(L,R) + Trapezius(L,R)} \times 100$$

Coupling Analysis

Estimation of the degree of coupling between selected paired acceleration signals was determined by cross correlation (Pearson product moment). Correlation analysis was performed on the filtered accelerometer data prior to RMS conversion. For this analysis, the peak coefficient between two signals was calculated over a range of time-lags ($\pm$5 s), with the maximal value being used as a measure of the coupling strength. Coupling analyses was performed between the neck-trunk, head-neck, and head-trunk combinations across the different bracing conditions. Comparisons were limited to acceleration in a single direction (i.e., no comparisons were made for different directions within a single segment or between segments).

Statistical Analysis

Within-subjects general linear mixed models were used to assess for differences in the selected dependent measures across the four conditions. Where significant effects were reported, post hoc evaluations were performed using Tukey's honestly significant difference test. All statistical analyses were performed using SAS statistical software (SAS Institute), with the risk of Type I error set at $P < 0.05$.

RESULTS

Acceleration Patterns

Figure 2 provides an example of the typical differences in the acceleration pattern (in the AP, ML and vertical directions) for the head (*top*) and lower trunk (*bottom*) segments during the control (no bracing) condition. This figure also highlights the frequency profile for the respective accelerometer signals for each segment during this same condition.

RMS acceleration. VERTICAL. Figure 3 (*top*) highlights the overall pattern of change in mean RMS acceleration for each segment/direction and the transfer function results across bracing conditions. For motion in the vertical direction, a significant main effect for condition was found ($F_{3,31} = 10.88$, $P < 0.001$). Post hoc analyses revealed that differences were seen between the control condition and all braced conditions, indicating that bracing the neck or trunk led to increased vertical acceleration, while combined bracing (neck and trunk) decreased vertical acceleration compared with the unbraced conditions. Additionally, the vertical acceleration observed during the neck-trunk-braced condition was significantly less than when the neck or the trunk was singularly braced (all P values < 0.05).

AP. Analysis of the differences in the AP RMS acceleration revealed a significant condition effect ($F_{3,31} = 4.07$, $P <$

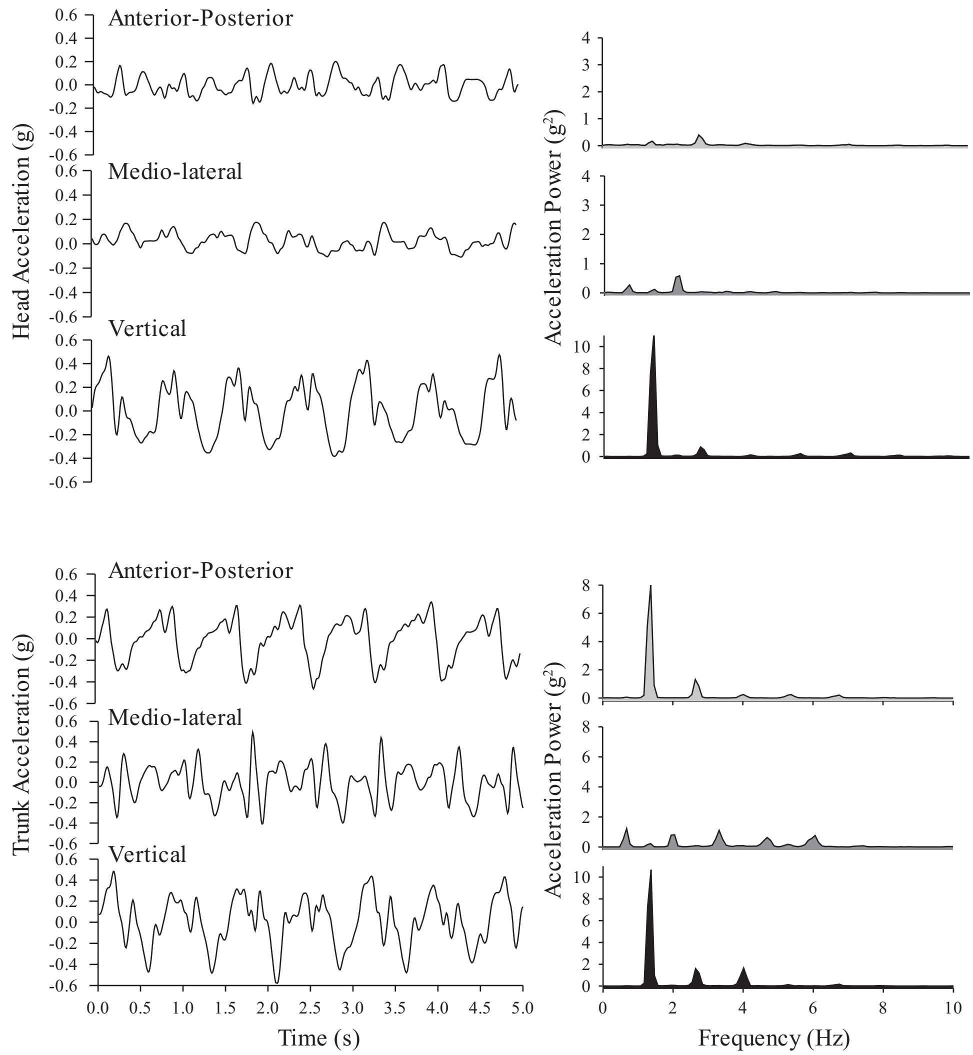

Fig. 2. Representative examples of the pattern of head (*top*) and lower trunk (*bottom*) acceleration [in the vertical, anterior-posterior (AP), and mediolateral (ML) directions] during the control (unbraced) walking condition. Power spectral profiles for each signal are also shown. All data were obtained from the same subject during a single trial within this condition.

0.001). Subsequent analysis revealed that the mean AP RMS acceleration for the neck-braced-only condition was significantly less than for all other conditions (all P values < 0.05).

ML. No significant condition effects were found for the ML RMS acceleration (all P values > 0.05).

Segmental gain (RMS). VERTICAL. As illustrated in Fig. 3 (*top*), negative values (i.e., indicative of attenuation) were found across all conditions, except for the trunk-neck segment combination during the trunk-braced condition (values were positive here). Significant main effects for condition were found for the neck-head ($F_{3,31}$ = 10.31, $P < 0.001$) and trunk-neck ($F_{3,31}$ = 4.47, $P < 0.001$) combinations. Post hoc analysis revealed that, for both combinations, significant differences were seen between the neck/trunk-braced condition and all other conditions (all P values < 0.05). During this condition, neck-head attenuation values were significantly less than for the other three conditions, while the opposite was seen for the trunk-neck combination, where attenuation values were greater under the neck/trunk-braced condition.

AP. The results of this analysis revealed positive (i.e., gain) effects for the neck-head and attenuation effects (i.e., negative values) for the trunk-neck. Significant main effects for condition were seen for both the head-neck ($F_{3,31}$ = 71.50, $P <$ 0.001) and neck-trunk ($F_{3,31}$ = 85.51, $P < 0.001$). Between the neck-head, the overall gain was significantly less for the control and trunk-braced conditions compared with the conditions where the neck was braced (all P values < 0.05). For the trunk-neck, significant differences were seen between all conditions, with the largest attenuation being for the neck-braced-only condition and the lowest attenuation values seen for the trunk-braced condition (all P values < 0.05).

ML. Transfer function values for ML RMS values revealed a mix of gain and attenuation effects for the neck-head, while only attenuation effects (i.e., negative) were recorded for the trunk-neck combination. A significant condition effect was found for the neck-head ($F_{3,31}$ = 27.28, $P < 0.001$) and trunk-neck ($F_{3,31}$ = 71.78, $P < 0.001$) transfer values. Subsequent analysis revealed that, for both combinations, significant differences were between all conditions (all P values < 0.05). For the neck-head, attenuation decreased from the control condition to the point where positive value (i.e., gain in acceleration) was seen for the trunk/neck-braced condition.

ApEn. VERTICAL. Figure 4 (*top*) illustrates the overall pattern of change in ApEn values for each segment and direction across the four conditions. A significant condition effect was found ($F_{3,31}$ = 6.00, $P < 0.001$) with ApEn values for the control condition being significantly less than for those condi-

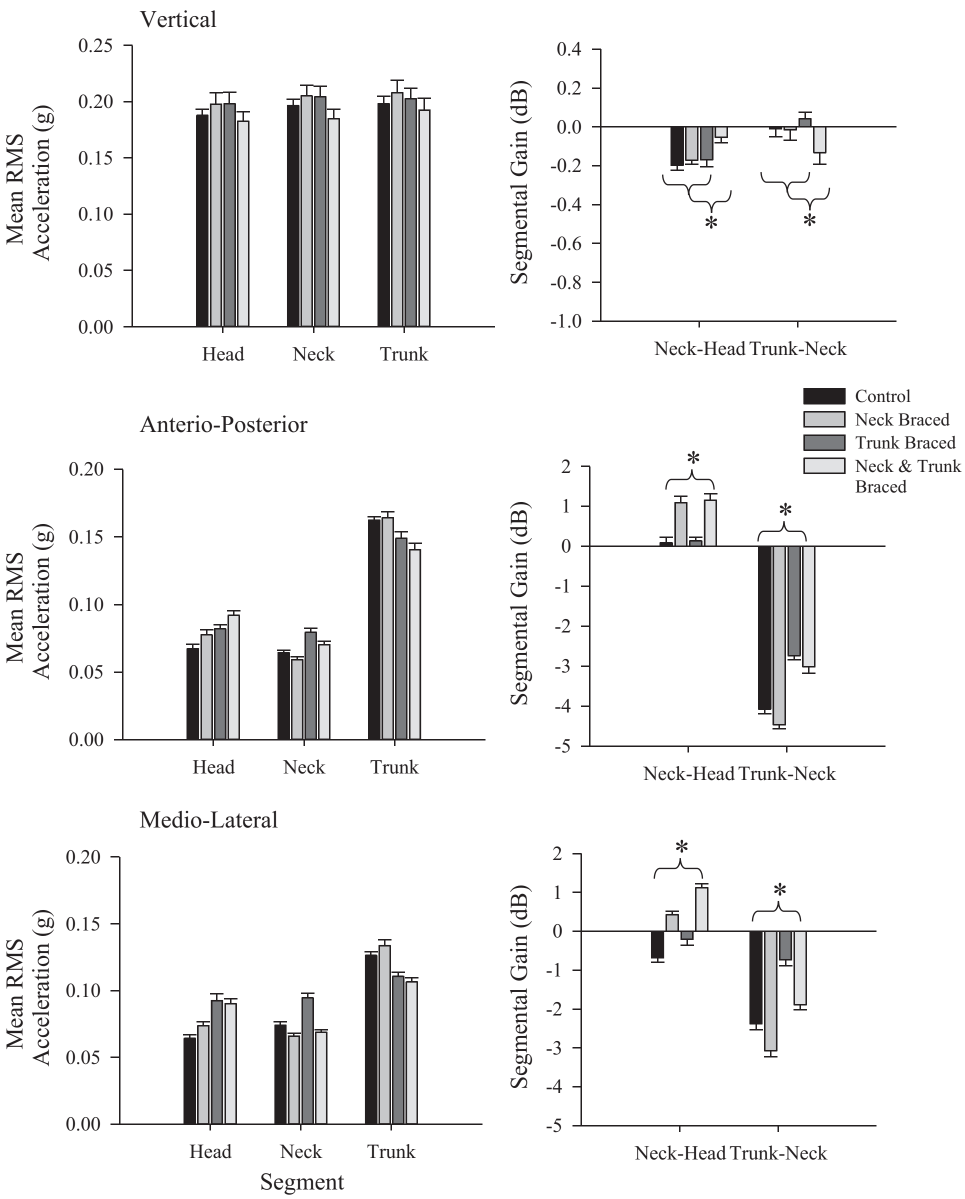

Fig. 3. Bar graph depicting differences in mean root mean square (RMS) acceleration for the head, neck and lower trunk segments (in the vertical, *top*; AP, *middle*; and ML, *bottom* directions) across the four different task conditions. Transfer function was also applied to the RMS values, and the results for the head-neck and neck-trunk transfer function are shown. Positive values indicate that accelerations increased from inferior to superior segments (i.e., gain), while a negative value indicates that RMS acceleration decreased between segments (i.e., illustrates overall attenuation). Values are means ± 1 SE. *For the transfer function results, significant differences between conditions are noted: $P < 0.05$.

tions where the trunk was braced (i.e., neck-trunk and trunk; all P values < 0.05).

AP. No significant condition effect was seen for ApEn values for the AP accelerations.

ML. No significant changes in ApEn for the ML accelerations were found for the different conditions.

Segmental gain (ApEn). VERTICAL DIRECTION. As shown in Fig. 4 (*top*), predominantly negative values were found across conditions, with the attenuation generally being greater between the trunk and neck segments. The only indication of gain was observed between the neck and head under the fully braced conditions. A significant condition effect was found for neck-head ($F_{3,31} = 14.75$, $P < 0.001$) and trunk-neck ($F_{3,31} = 19.16$, $P < 0.05$). For the head-neck combination, post hoc analysis revealed that the differences were between all conditions, except for control and neck braced. Between the trunk-neck, differences were between all conditions except for trunk braced and neck/trunk braced, with the attenuation being greatest for the control condition and decreasing as a function of bracing (all P values < 0.05).

AP DIRECTION. A significant condition effect was found for neck-head ($F_{3,31} = 8.63$, $P < 0.001$) and trunk-neck ($F_{3,31} = 10.20$, $P < 0.001$). For transfer between the neck-head, values were greater (positive) for the neck-trunk-braced conditions compared with all other conditions (all negative). For the trunk-neck, differences were between all conditions except for control and neck braced (P values < 0.05, see Fig. 4, *middle*).

ML DIRECTION. A significant main effect of condition was observed between the neck-head ($F_{3,31} = 14.49$, $P < 0.001$) and trunk-neck ($F_{3,31} = 57.75$, $P < 0.001$). For the neck-head, transfer values were significantly less for the neck-trunk braced conditions compared with all other conditions (all values were negative). Positive (gain) values were observed for the trunk-neck, with the control and neck braced conditions having higher gain values compared with trunk and neck-trunk braced conditions (all P values < 0.05, see Fig. 4, *bottom right*).

Frequency. VERTICAL. The frequency profile for the vertical component of the each segment (see Fig. 2, *top*) under control

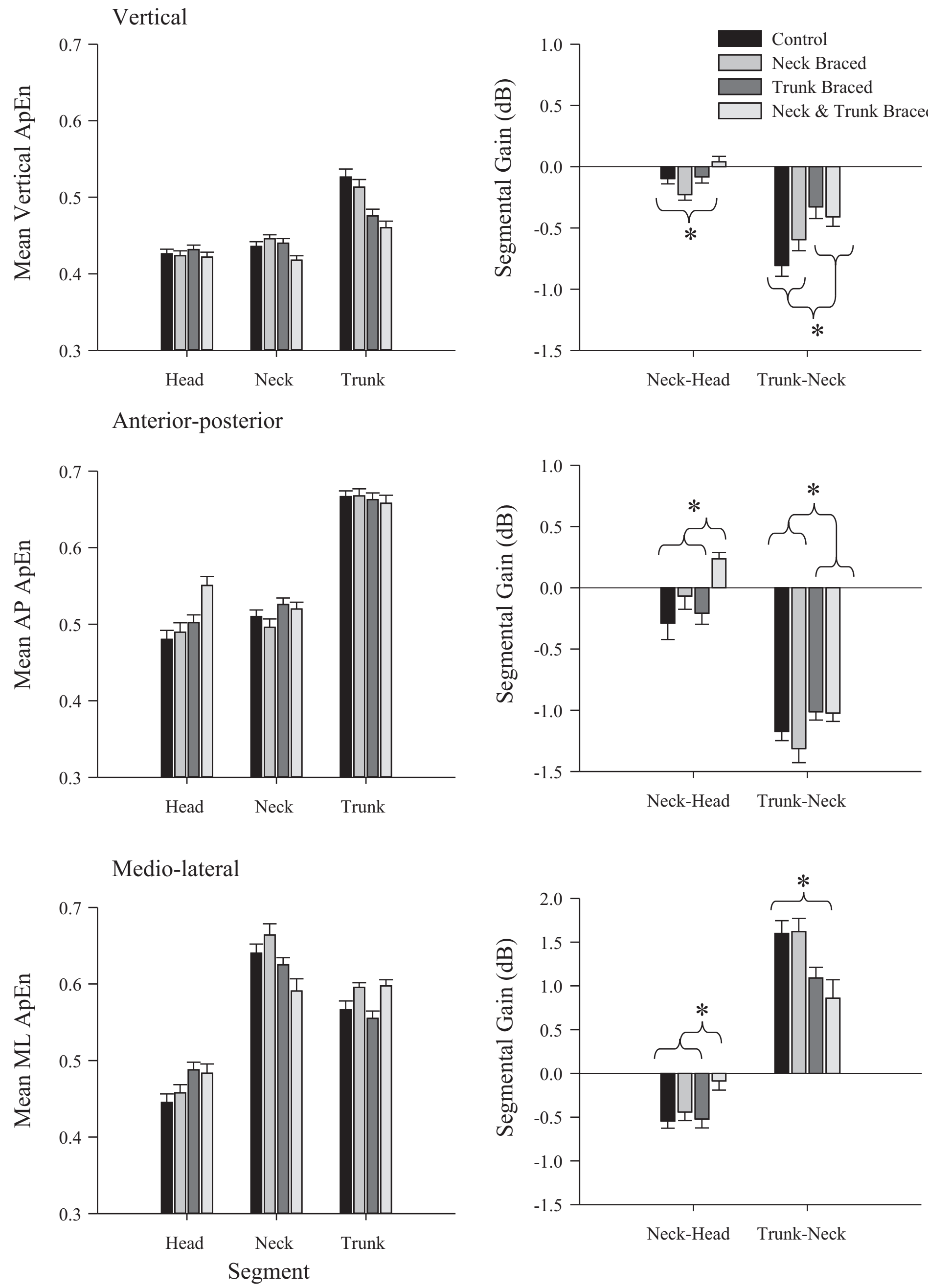

Fig. 4. Changes in pattern of regularity [mean approximate entropy (ApEn) values] for the head, neck and lower trunk segments across the four different conditions. *Top*: the changes in vertical ApEn values. *Middle*: AP changes. *Bottom*: ML changes. Transfer function was also applied to the ApEn values, and the results for the head-neck and neck-trunk combinations are shown. Positive values indicate that ApEn increased from inferior to superior segments (i.e., gain), while a negative value indicates that ApEn decreased (i.e., attenuation). Values are means ± 1 SE. *For the transfer function results, significant differences between conditions are noted.

conditions was characterized by a prominent peak between 1 and 2 Hz (head 1.87 ± 0.03 Hz; neck 1.57 ± 0.08 Hz; trunk 1.57 ± 0.08 Hz). A significant condition effect was found for the frequency of these peaks ($F_{3,31}$ = 14.65, $P < 0.001$), with differences being found between all conditions except for control-neck braced and trunk-braced trunk/neck-braced conditions (all P values < 0.05). Overall, the frequency of the peak vertical accelerations was greatest during the control, unbraced conditions. No significant change in peak power were observed between the different bracing conditions.

AP. As highlighted in Fig. 2, a prominent peak was seen in the AP frequency profile for all segments around 1–2 Hz with multiple harmonics also being observed (head 1.85 ± 0.04 Hz; neck 1.44 ± 0.02 Hz; trunk 1.25 ± 0.02 Hz). A significant condition effect was found for the frequency of peak power ($F_{3,31}$ = 14.90, $P < 0.001$) within the 0–3 Hz range. Post hoc analysis revealed that, under the control conditions, the peak power was observed at a lower frequency compared with the braced conditions. Additionally, the frequency of peak power was significantly greater for the neck/trunk-braced conditions compared with the other two bracing conditions (all P values < 0.05). No significant differences in peak power were observed between the different bracing conditions.

ML. The frequency profile for the ML component of each segment under control conditions was characterized by a peak between 1–2 Hz (head 1.21 ± 0.11 Hz; neck 1.22 ± 0.10 Hz; trunk 1.57 ± 0.08 Hz) and multiple harmonic components. A significant condition effect was found for the peak power ($F_{3,31}$ = 6.22, $P < 0.002$) and frequency of peak power ($F_{3,31}$ = 3.84, $P < 0.019$) within the 0–3 Hz range. Post hoc analysis revealed that peak power for the trunk-braced condition was significantly greater to that seen during all other conditions (all P values < 0.05). In addition, the frequency of peak power was significantly less during the two less constrained conditions (control, neck braced only) compared with the two conditions where the trunk was braced (all P values $<$ 0.05).

Muscle Activation Measures

Mean RMS EMG. Figure 5 illustrates the general changes in the mean RMS values for each muscle as a function of the bracing conditions. As the initial inferential analyses revealed no left-right differences for any muscle group, the results were collapsed across body side and are presented below.

For the RA and EO muscle groups, the results revealed a significant effect for condition (RA $F_{3,31} = 19.282$; EO, $F_{3,31} = 27.12$, P values < 0.001). For both muscles, activity was greatest during the control condition and significantly decreased during the trunk-braced-only and the trunk/neck-braced conditions (P values < 0.05).

A significant main effect for condition was found for the activity in TZ ($F_{3,31} = 6.52$, $P < 0.002$) and ES ($F_{3,31} = 25.70$, $P < 0.001$). For the TZ muscle, the neck-trunk-braced condition led to significantly reduced activity compared with all other conditions. In contrast to the results for the abdominal muscles, activity in the ES muscles was lowest for the control and neck-braced conditions and increased significantly as a function of bracing of the trunk (i.e., for the trunk-braced and trunk/neck-braced conditions).

CI. Analysis of the level of coupling between selected muscle pairs (CI) revealed a significant condition effect ($F_{3,31} = 62.51$, $P < 0.05$). Post hoc analyses revealed the level of coactivation between the selected muscles was greatest during the control, no-bracing condition (mean CI 78.4 $\pm$ 4.9%), and it decreased significantly for the trunk-braced (CI 50.9 $\pm$ 3.1%) and neck-trunk braced conditions (CI 59.2 $\pm$ 4.1%). No significant difference was observed between the neck-braced only and the control condition ($P > 0.05$).

Coupling Analyses

Cross correlation. This analysis revealed that the strength of the coupling relations was greatest between the head, neck and trunk segments in the vertical direction (head-neck $r = 0.90$, neck-trunk $r = 0.88$, head-trunk $r = 0.91$). There was no significant condition effect, as these specific head-neck-trunk relations remained consistently high across all bracing conditions (r range 0.89–0.97). Notable correlations were also seen between the neck and head in the ML direction ($r = 0.32$),

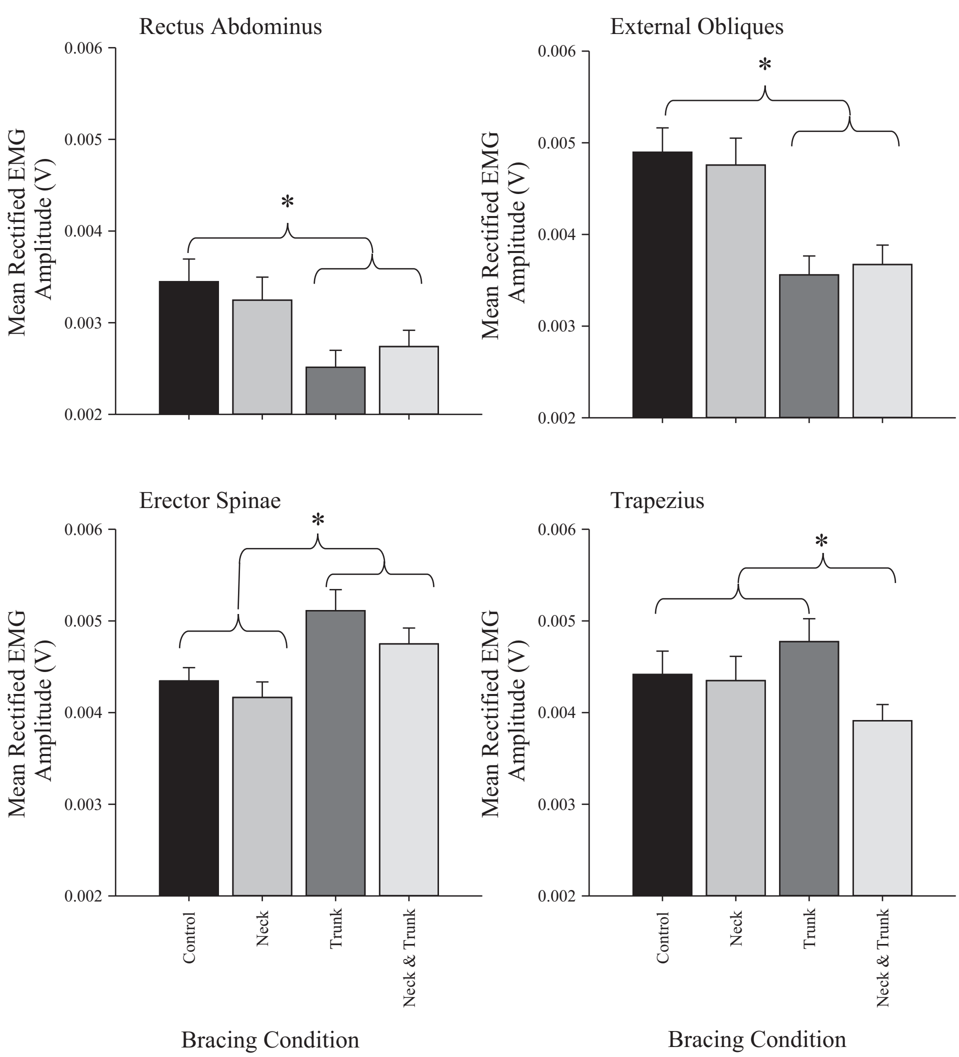

Fig. 5. Changes in the mean amplitude of muscle activity across the four different conditions. As there were no significant left-right differences for any of the selected muscle groups, the average values collapsed across both body sides are shown. Values are means $\pm$ 1 SE. *Significant differences between conditions: $P < 0.05$.

although the strength of this relation decreased for other segmental relations (ML neck-trunk $r = -0.23$, ML head-trunk $r = 0.13$). The strength of these correlations remained constant across conditions. For coupling in the AP direction, no notable relations were observed for these same segments across all conditions (head-neck r range $= -0.06$-0.04; neck-trunk $r = 0.01$–0.25, head-trunk $r = 0.12$ to -0.02).

DISCUSSION

This study was designed to examine the effect of external bracing on the attenuation and pattern of acceleration from the trunk, neck and head and related trunk muscle activity during overground walking. When the neck and/or trunk were singularly braced, an overall decrease in the ability of the lower trunk to attenuate gait-related oscillations was observed, which led to increases in the amplitude in vertical acceleration for all segments. However, when the trunk and neck were braced together, the level of acceleration across all segments decreased in line with increased attenuation from the neck to the head. With regards to muscle activity, activity in ES tended to increase, while activity in the trunk flexors (i.e., RA and EO muscles) decreased as a function of bracing. This led to an overall decline in the level of trunk muscle cocontraction. The described changes seen in the acceleration and muscle activation patterns were observed, even though there were no significant changes in gait speed across conditions. Consequently, it is more likely that these changes reflect adaptive responses to compensate for the restrictions imposed by the bracing conditions, rather than as a function of changes in walking speed per se.

Pattern of Acceleration Attenuation during Gait

Under control (no bracing) conditions, the pattern of acceleration for the head, neck and lower trunk segments in three axes of motion were consistent with the findings of previous studies (Kavanagh 2009; Kavanagh et al. 2004, 2005a; Mazzà et al. 2009; Moe-Nilssen and Helbostad 2004). In general, acceleration amplitude was greatest for the lower trunk and smallest for the head. The reduced acceleration amplitude between the trunk and head occurred through smaller peak power and fewer/smaller harmonics across all axes of motion. During the walking trials, the trunk played a major role in the attenuation of gait-related oscillations in all direction, as evidenced by the large negative transfer values (see Fig. 3). This result, which is consistent with previous reports (Kavanagh et al. 2006b; Mazzà et al. 2009; no. 234; Prince et al. 1994), confirms that the trunk, through both active and passive mechanisms, can act to damp the transmission of oscillations during walking. However, while the largest accelerations occurred in the vertical direction, the largest attenuation of accelerations occurred between the lower trunk (L_3) and neck (C_7) in the AP direction, followed by the ML direction, with only a small relative change in attenuation occurring for vertical oscillations. Interestingly, the general increases in attenuation of AP and ML accelerations from the trunk to the neck seen with bracing were counterbalanced by moderate gain increases (i.e., increased transmission of accelerations) between the neck and head segments. These results indicate that oscillations present at the lower trunk are attenuated rather than simply transmitted to the head, a process that is probably mediated through the combination of activity in specific musculature and/or the intrinsic ability of the spinal structures to absorb gait-related oscillations (Kavanagh et al. 2006b; Mazza et al. 2008; Ratcliffe and Holt 1997). The results for the ApEn analysis tend to support this claim. Across all three directions, the acceleration signals for the head were more regular or predictable (ApEn was lowest). There was also a tendency for attenuation of ApEn acceleration values from inferior to superior. Given that lower ApEn values reflect increased regularity and hence greater control being exerted (Kavanagh et al. 2005b, 2006a; Pincus 1995), attenuation of the accelerometer signal from the lower trunk to the head supports the view that controlling head motion was prioritized during walking. The ability to maintain a more regular and predictable degree of head motion would optimize the accuracy of the visual and vestibular sensory systems located in the head (Berthoz and Pozzo 1988; Pozzo et al. 1990, 1991).

Impact of Bracing on Vertical Head Acceleration

The importance of specifically controlling vertical head accelerations is undergirded by the findings for the three different bracing conditions. Singularly bracing the neck and/or the trunk was expected to have negative consequences for attenuation of gait-related oscillations and hence would impact on stabilization of the head during walking. This prediction was supported, except for a notable exception in the vertical axis. For motion in the AP and ML axes, increasing the degree of bracing led to a general decrease in the degree of attenuation, particularly from the lower trunk to the head, with larger amplitude accelerations at the head developing, as was expected. Along with larger oscillations, the regularity of head motion also decreased with bracing (ApEn increased), as did the level of attenuation. In general, the combination of bracing both the trunk and neck together led to an overall gain effect (or reduced attenuation) for both RMS and ApEn between the neck and head segments.

Findings for the vertical axis suggest a change in strategy when both the neck and trunk are braced. While bracing either of these segments singularly led to significant increases in the amplitude of vertical acceleration and no significant change in attenuation, bracing both segments together led to significantly increased attenuation between the neck-head (and paradoxically, decreased attenuation between the trunk-neck). The result of this was an overall decrease in vertical head accelerations compared with the control and other bracing conditions. This result would indicate that, rather than allowing vertical head acceleration amplitude to increase further when both the neck and trunk were braced, participants were able to adapt their gait through increasing the damping ability of the trunk to ensure the vertical acceleration remained low. The high correlations between segments were uninfluenced by bracing, indicating that only small adjustments to acceleration could be achieved through the trunk and neck, with or without bracing. This ability to effectively damp gait-related oscillations under the neck/trunk-braced conditions could have been achieved through altering the pattern of trunk muscle activity and/or by compensatory actions within the lower limb (Light et al. 1980; MacKinnon and Winter 1993; Prince et al. 1994; Ratcliffe and Holt 1997). Given that muscles of the trunk (particularly ES) play an important role in stabilizing the spine during walking and minimizing vertical motion of the body (Klemetti et al. 2014; MacKinnon and Winter 1993; Saunders et al. 1953; White and McNair 2002; Winter et al. 1993), the finding of increased activity in ES during conditions where the trunk was braced may reflect that this muscle group played a more active role in attenuating gait-related oscillations. Similarly, the lower limb also plays a significant role in the damping of oscillations during locomotion (James et al. 2014; Kavanagh et al. 2006b; Light et al. 1980; Ratcliffe and Holt 1997). It has been previously reported that individuals are able to minimize the transmission of impact-related forces during walking by adjusting motion about the knee and ankle joints (James et al. 2014). Within the context of the present study, it is possible that participants may have adapted their walking pattern (while maintaining the required gait speed) utilizing the muscles within the lower limb to counteract the stiffening effects cause by externally bracing of the trunk. Indeed, the finding that the vertical acceleration was also reduced at the lower trunk level during the combined bracing condition compared with the control conditions supports the position that the overall reduction was, in part, achieved by segments of the lower limb, up to and including L_3. In contrast, the difference in accelerations between L_3 and the head decreased with increasing degree of bracing for AP and ML axes of motion. This was driven by increasing accelerations at the head and decreasing accelerations at the lower trunk level. As was predicted, bracing makes it more difficult to attenuate accelerations between the trunk and head, especially in the AP and ML axes. For the majority of bracing conditions, decreases in attenuation of acceleration in the ML and AP directions and, in some instance, increased gain were observed compared with the control condition responses. As with the vertical accelerations, utilizing the lower limbs to minimize accelerations in AP and ML directions may be an effective response for controlling the increasing accelerations seen at the head in these axes (James et al. 2014; Kavanagh et al. 2006b; Light et al. 1980; Ratcliffe and Holt 1997).

As speed was controlled across all conditions, participants were able to attenuate vertical gait-related oscillations to minimize the impact on the head. The importance of being able to effectively dampen vertical accelerations is also indicated by the lowest ApEn values for vertical head signals. While the predictability of head acceleration decreased in the AP and ML axes (i.e., greater ApEn values) with greater bracing, vertical head acceleration regularity was maintained (lowest ApEn) of all segments across the bracing conditions, suggesting that maintaining a regular level of vertical head acceleration is critical to the task of walking. Given that accelerations in the vertical axis are larger than in AP or ML axes, having a more regular motion of the vertical axis may serve to provide a more consistent and stable platform for the vestibular and visual systems to operate (Berthoz and Pozzo 1988; Cappozzo 1981; Carlsen et al. 2005; Leah et al. 2005; Mazza et al. 2008; Pozzo et al. 1990, 1991).

Changes in Muscle Activity with Bracing

The changes in the pattern of acceleration as a function of bracing were also reflected by a similar changes in muscle activity. For the abdominal muscles (RA and EO), the amplitude of muscle activity was greatest during the control/nobracing condition and decreased significantly as the trunk (either singularly or in combination with the neck) was braced. For these muscles, the lowest amount of activity was seen during those conditions where motion within the trunk was restricted (i.e., the trunk braced or the neck/trunk braced conditions). In contrast, activity in the ES muscle followed an opposite direction, with activity being greatest under conditions where the trunk was braced (singularly or in combination with the neck) and was least during the control and neck-braced only conditions. As a result of this reciprocal pattern of change across the selected muscle groups, there was an overall decline in the degree of muscle coactivation from the control (nonbraced) condition to the most restrictive conditions (trunk braced only and trunk-neck braced). However, although there was an overall decrease in trunk muscle coactivation, this does not imply that trunk stiffness decreased similarly. Rather, it seems likely that bracing provided a degree of external stability that led to reduced activity for the abdominal muscles (RA and EO) and hence lower levels of muscle coactivation. Previous work by Cholewicki and colleagues (2006, 2007, 2010) reported similar findings during the performance of a rapid trunk force release task. For these studies, it was reported that overall trunk stiffness was increased through the wearing of a lumbosacral orthosis, even though muscle coactivation patterns decreased during the same postural perturbation tasks. The observed increases in ES activity with trunk bracing coupled with the reduced activity in the abdominal muscles (RA and EO) may reflect a neuromuscular adaptation to the added mechanical support provided by the external brace. Indeed, the finding of increased activity for ES is consistent with the general premise linking increased activity in the trunk extensors to greater trunk stability (Cholewicki et al. 1997; Klemetti et al. 2014; Lee et al. 2006; Moorhouse and Granata 2005). One further consideration is that the increased activity in the back extensors could have been related to an increase in the amount of forward inclination (lean) of the trunk during the more constrained (braced) conditions. Previous research has reported that increasing forward lean leads to increased lumbar extensor moments (Leteneur et al. 2009), which can lead to increased activity in the trunk extensor muscles (Kluger et al. 2014; Saha et al. 2008). While any change in inclination was compensated for during the processing of the accelerometer signals, there is the possibility that small increases in trunk inclination due to bracing may have also contributed to the increased activity in the back extensor muscles. Irrespective of the actual mechanism, an overall decrease in trunk muscle coactivation was subsequently observed in response to external bracing. Under the conditions where additional external stability is provided by the brace(s), increasing ES activity was sufficient to stabilize the spine while still ensuring the gait task was performed to the desired level.

There are some limitations to our design. Of the total of 12 subjects enrolled in this study, only 2 were men. While preliminary initial analysis revealed no differences between the male and female participants in this study (in terms of RMS amplitude), sex differences have been previously reported in relation to the ability to attenuate gait-related oscillations (Mazzà et al. 2009) and, in a sample of young and older adults, with regards to differences in gait symmetry (Kobayashi et al. 2014). In particular, Mazzà and colleagues (2009) reported that healthy young women exhibited greater ability to control both

AP and ML head accelerations (Mazzà et al. 2009), exhibiting a more effective ability to attenuate gait-related oscillations from the pelvis to the shoulder during overground walking. It is possible that interindividual differences in attenuation ability may have been revealed by including equal proportion and numbers of male and female subjects.

In conclusion, externally restricting the motion of the neck or trunk, which were singularly restricted through external bracing, impacted the ability of the upper body to attenuate gait-related oscillations, typically leading to significant increases in the amplitude and complexity of the vertical head acceleration. In contrast, when the neck and trunk were braced in combination, a significant decline in vertical accelerations was found across all segments. While this decline was coupled with increased attenuation between the trunk and neck segments, the decrease in oscillations about the lower trunk (L_3) indicates that the lower limbs also played a role in damping oscillations during this gait task. Bracing the trunk likely led to increased trunk stiffness and was reflected by decreased activity in the trunk flexors (RA, EO), increased activity in the trunk extensors (ES) and subsequently lower levels of muscle coactivation. The findings of the present study may have some general applicability to understanding the impact altering trunk stiffness has on gait mechanics. These results may be particularly pertinent to older adults who exhibit stiffening of the spine or for addressing how individuals with lower back pain compensate for the increased trunk stiffness that can come with wearing an external brace while walking. Overall, it would appear that the neuromuscular system of young, healthy individuals was able to maintain a consistent pattern of head acceleration, irrespective of the level of external bracing, and that priority was placed over the control of vertical head accelerations during these gait tasks.

DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

Author contributions: S.M. and D.M.R. conception and design of research; S.M., D.M.R., and K.K. performed experiments; S.M., D.M.R., and K.K. analyzed data; S.M., D.M.R., K.K., and M.L.W. interpreted results of experiments; S.M. prepared figures; S.M., D.M.R., and M.L.W. drafted manuscript; S.M., D.M.R., K.K., and M.L.W. edited and revised manuscript; S.M., D.M.R., and M.L.W. approved final version of manuscript.

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