RESEARCH ARTICLES

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The influence of artificially increased hip and trunk stiffness on balance control in man

Abstract Lightweight corsets were used to produce mid-body stiffening, rendering the hip and trunk joints practically inflexible. To examine the effect of this artificially increased stiffness on balance control, we perturbed the upright stance of young subjects (20–34 years of age) while they wore one of two types of corset or no corset at all. One type, the "half-corset", only increased hip stiffness, and the other, the "full-corset", increased stiffness of the hips and trunk. The perturbations consisted of combined roll and pitch rotations of the support surface (7.5 deg, 60 deg/s) in one of six different directions. Outcome measures were biomechanical responses of the legs, trunk, arms and head, and electromyographic (EMG) responses from leg, trunk, and upper arm muscles. With the full-corset, a decrease in forward stabilising trunk pitch rotation compared to the no-corset condition occurred for backward pitch tilts of the support surface. In contrast, the half-corset condition yielded increased forward trunk motion. Trunk backward pitch motion after forwards support-surface perturbations was the same for all corset conditions. Ankle torques and lower leg angle changes in the pitch direction were decreased for both corset conditions for forward pitch tilts of the support-surface but unaltered for backward tilts. Changes in trunk roll motion with increased stiffness were profound. After onset of a roll support-surface perturbation, the trunk rolled in the opposite direction to the support-surface tilt for the no-corset and half-corset conditions, but in the same direction as the tilt for the full-corset condition. Initial head roll angular accelerations (at 100 ms) were larger for the full-corset condition but in the same direction (opposite platform tilt) for all conditions. Arm roll movements were initially in the same direction as trunk movements, and were followed by large compensatory arm movements only for the full-corset condition. Leg muscle (soleus, peroneus longus, but not tibialis anterior) balance-correcting responses were reduced for roll and pitch tilts under both corset conditions. Responses in paraspinals were also reduced. These results indicate that young healthy normals cannot rapidly modify movement strategies sufficiently to account for changes in link flexibility following increases in hip and trunk stiffness. The changes in leg and trunk muscle responses failed to achieve a normal roll or pitch trunk end position at 700 ms (except for forward tilt rotations), even though head accelerations and trunk joint proprioception seemed to provide information on changed trunk movement profiles over the first 300 ms following the perturbation. The major adaptation to stiffness involved increased use of arm movements to regain stability. The major differences in trunk motion for the no-corset, half-corset and full-corset conditions support the concept of a multi-link pendulum with different control dynamics in the pitch and roll planes as a model of human stance. Stiffening of the hip and trunk increases the likelihood of a loss of balance laterally and/or backwards. Thus, these results may have implications for the elderly and others, with and without disease states, who stiffen for a variety of reasons.

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Introduction

The strategies humans use to regain upright posture after a perturbation to quiet stance have been described extensively. Studies on quiet stance and pitch-plane dynamic posturography fuelled theories about human balance control as an inverted pendulum about the ankle joint when body motion is small (Nashner and McCollum 1985; Fitzpatrick et al. 1992, 1994; Winter et al. 1998; Gatev et al. 1999; Jacobs 1997; Lauk et al. 1999; Johansson and Magnusson 1991), or like a two-joint system with motion restricted to the hip and ankle joints (the "hip strategy") when body motion is larger (Horak and Nashner 1986; Kuo and Zajac 1993; Henry et al. 1998a). For these models, motion at the knees or the lumbro-sacral joints is assumed to be minimal. The inverted pendulum model may be valid for a restricted number of movements in a single plane, including those induced by small and slow horizontal support surface translations. Recently, however, this concept has been challenged for motion occurring during quiet stance (Aramaki et al. 2001; Loram and Lakie 2002). Indeed, as body motion increases in amplitude and direction from that of quiet standing (Fitzpatrick et al. 1992, 1994; Winter et al. 1996, 1998; Gatev et al. 1999; Accorneo et al. 1997), to that induced by a support surface perturbation in the pitch plane alone (Cordo and Nashner 1982; Allum et al. 1993; Horak et al. 1997), and finally to that induced by combined roll and pitch plane perturbations (Moore et al. 1988; Maki et al. 1994a, 1994b; Henry et al. 1998b; Carpenter et al. 1999; Allum et al. 2002), the multi-link nature of human postural corrections becomes increasingly prominent. This multi-link strategy involves hinging at the knees, hips and lower vertebral column, in addition to ankle joint motion (Allum et al. 2003). With the use of perturbations applied to the trunk and pelvis, these multi-link aspects of trunk motion become even more prominent (Gilles et al. 1999; Rietdyk et al. 1999).

The common aim of these studies—and the models that ensued—was to provide evidence about the processing of sensory information in the central nervous system (CNS), and to better understand how motor commands are generated to withstand external balance perturbations in different directions. Presumably the use of multiple joints would require use of feedback information from all these joints, as well as increased interlink coordination, to maintain stability compared to the requirements to control motion about the ankle joint alone. Furthermore, adequate balance corrections in the pitch and roll planes must be processed—and perhaps generated—sequentially as the trunk moves more rapidly in roll than pitch when compensating for multidirectional falls (Carpenter et al. 1999).

The inverted pendulum or multi-link concepts of human postural control can be tested directly by splinting various joints, but such experiments are rare and were restricted to the pitch plane (Loram and Lakie et al. 2002; Peterka 2002). Here we tested the hypothesis that normal balance control is highly dependent on a multi-link mode of movement by stiffening the hips and trunk (but not the ankles and knees) with two different rigid corsets. Our assumption was that stiffening the hips and trunk would change the response dynamics of the body detrimentally as it would then resemble an inverted pendulum and be more unstable. If, however, the CNS controlled the body predominantly as an inverted pendulum by naturally stiffening the hips and trunk, then little difference in movement strategies and muscle strategies should be observed.

Our second goal was to clarify the pathophysiology of trunk stiffness which we suggested helps cause impaired balance control with ageing and in neurological disorders, including Parkinson's disease and severe proprioceptive loss (Carpenter et al. 2004; Allum et al. 2002; Bloem et al. 2002). In all these groups, changes in trunk roll motion were noted as early as 50 ms after a postural perturbation, suggesting that active stiffening prior to the perturbation—rather than inappropriate active postural reactions—was mainly responsible for the loss of trunk flexibility. If this were true, then artificial stiffening with corsets should largely reproduce this previously observed pattern of trunk stiffness. The corsets should also induce marked instability for another reason. Prior studies suggested that the difference in trunk flexibility in the roll and pitch directions is crucial to maintaining balance following multidirectional perturbations (Carpenter et al. 1999; Allum et al. 2002). Thus eliminating or changing this difference with corsets might lead to difficulties in CNS processing of balance corrections.

Our third goal was to study the nature of any changes in muscle synergies following artificial stiffening. We were particularly interested to see if any alterations in muscle response amplitudes might mimic the changes observed previously with ageing. Elderly subjects have increased trunk stiffness, particularly in the roll plane, and this occurred in parallel with delayed muscle onset times, reduced amplitudes of balance correcting muscle responses and, in the leg muscles, increased amplitudes of later occurring balance corrections (Allum et al. 2002). It remained unclear whether these changes all resulted from age-related degeneration, or whether some alterations were in fact compensatory strategies for the loss of intrasegmental flexibility of the trunk. We predicted that any compensatory changes should be reproduced in young persons with artificial stiffening of hips and trunk. Alternatively, arm movements might be used as a compensatory mechanism if balance corrections in the trunk and legs were insufficient to accommodate the effects of stiffening.

Methods

Five healthy subjects (three men, two women, age 22–34 years) were examined under three conditions. The first condition was without any corset ("no-corsets"). The second condition involved wearing a custom-fitted "half-corset" extending from above the knees up to approximately level Th 10 (see upper part, Fig. 1). This corset reduced hip joint motion and relative motion between the hip joints, for example when one knee flexes and the other remains

extended to less than 1 deg (Table 1). Such differential knee flexion was limited by a bar fixed across the legs of the corsets. The third test condition involved wearing a custom-fitted "full-corset", similar in construction to the half-corset but extending all the way up to the shoulders and thereby also reducing trunk motion (lower part Fig. 1). This full-corset also restricted motion of the upper trunk (shoulders and sternum) with respect to the pelvis to less than 1 deg in pitch and 2 degs in roll (Table 1). Both corsets caused no restrictions in arm and lower-leg movements. Scotchcast and Softcast (3M™) material was used to construct the corsets. Testing was done in the same order for all subjects over 2 days. The first test was without a corset and the second 4 h later with the half-corset. Tests with the full-corset were performed a day later. All subjects were tested for a lack of orthopaedic or balance problems using techniques described in Allum and Adkin (2003) and then gave witnessed informed and written consent to participate in the experiments according to the Declaration of Helsinki. The Institutional Review Board of the University Hospital in Basel approved the study.

Outcome measures

EMG and biomechanical measurements were obtained using previously described techniques (Carpenter et al. 1999; Allum et al. 2002). To record EMG activity, pairs of silver-silver chloride electrodes were placed approximately 3 cm apart along the muscle bellies of left tibialis anterior, left soleus, left peroneus longus, left medial gastrocnemius and bilaterally on paraspinals at the L1-L2 level of the spine, and on the left medial deltoid (pars acromialis) muscles. Small apertures were cut in the back of the full-corsets to permit application of the paraspinal electrodes. EMG amplifier gains were kept constant throughout the experiments and pairs of electrodes and lead lengths assigned to individual muscles were not changed between and within subjects. The individual locations of the electrodes were marked and reused for each test condition.

Support-surface reaction forces of the left foot were measured from strain gauges embedded within the rotating support-surface. The strain gauges were located under the corners of the plate supporting the left foot. From these forces, the left anterior-posterior (AP) ankle torque was calculated. To measure lower leg angle in the pitch plane a lightweight metal rod was fixed with an adjustable strap to the lateral aspect of the left tibia, about 4 cm below the level of the lateral condyle. The rod was connected to a potentiometer located on the pitch axis of the platform. Trunk angular velocity in the pitch and roll planes was collected using Watson Industries transducers (±300 deg/s range) mounted onto a metal plate (total weight 700 g) that hung at the level of the sternum from shoulder straps that wrapped around the shoulders, back and chest. Two smaller Systron-Donner angular velocity transducers (Inglewood, CA, USA) measured movements of the left lower arm just below the elbow in the pitch and roll directions. These transducers had a range of ±200 deg/s. The transducers were attached to a 10-cm-long metal plate curved to the radius of the arm. The plate was strapped to the lateral aspect of the left lower arm using an elasticised bandage, and the total weight mounted in this fashion was 200 g. Head roll angular accelerations were computed from the outputs of two dual axis linear accelerometers (Entran), with ranges of ±5 g, mounted at 180 deg separation in the coronal plane just above the ears on a lightweight adjustable head band.

Procedure

The subject's feet were lightly strapped into heel guides fixed to the top surface of the dual-axis platform which rotated about the pitch (anterior-posterior) and roll (medial-lateral) planes. The heel guides were adjusted in the AP direction to ensure that the ankle joint axes were aligned with the pitch axis of the rotating platform. The roll axis had the same height as the pitch axis and passed between the feet. Both the corsets and the foot straps prevented stepping

Fig. 1 Front and side views of the two types of corsets. The *upper photos* are of the half-corset, which allows movement of the trunk with respect to the pelvis. The *lower photos* are of the full-corset. The Velcro straps (*black* for the full-corset, *white* for the half-corset) helped ensure a snug fit of the corsets across both the pelvis and trunk

reactions when stance perturbations occurred. Just prior to the experiment, subjects were asked to assume their 'preferred' standing posture with the arms hanging comfortably at their sides. At each individual's 'preferred-stance' position, we sampled the low pass filtered (5 Hz) sum of the AP torques from the two strain gauge systems embedded in the surface of the rotating platform under each foot. This measurement sample was then treated as the reference value for 'preferred-stance' for the remainder of the experiment.

Under each corset condition subjects were presented with two series of 44 perturbations each. The order of the two series among the three conditions was counterbalanced across subjects. The first trial of each series was excluded from data analysis to reduce habituation effects entering the data (Keshner et al. 1987). The perturbations comprised six different perturbation directions presented with one velocity (60 deg/s), and with a constant amplitude

Table 1 Hip and trunk maximum flexibility (deg). Angles were measured based on 18 markers placed on the body of one subject and tracked with a motion analysis system, Optotrak (for examples of the resultant stick figures, see Allum et al. 2003). The subject was asked to bend forward, backward or laterally as far as possible 8 times for each direction and within 2 s. An average of the body segment angles was then taken for each direction. The three markers defining the (upper) trunk were placed on both shoulders and the upper level of the sternum

Segment angle[a]	Direction	Forward			Backward			Lateral flexion		
		At 0.8 s[b] Normal	At 2 s Half	At 2 s Full	At 2 s Normal	At 2 s Half	At 2 s Full	At 2 s Normal	At 2 s Half	At 2 s Full
Right upper leg abduction		*	*	*	#	*	*	5.4	#	*
Right upper leg flexion		−4.8	*	#	5.4	*	*	2.9	*	*
Left upper leg abduction		*	*	*	*	*	*	4.2	#	*
Left upper leg flexion		−8.7	*	#	5.2	*	*	0.8	*	*
Trunk to pelvis relative angle	Pitch[c]	−48.7	−18.7	#	10.8	10.4	#	6.3	4.0	#
Trunk to pelvis relative angle	Roll	4.5	#	*	*	*	*	26.9	17.6	2.1
Trunk to pelvis relative angle	Yaw	4.4	−1.5	#	5.2	2.0	*	28.8	13.6	2.8
Absolute trunk angle	Pitch	−36.9	−18.0	*	19.1	10.0	3.0	9.0	2.8	*
Absolute trunk angle	Roll	1.0	1.2	*	3.2	1.2	*	41.4	22.5	3.9
Absolute trunk angle	Yaw	*	*	*	4.7	1.6	#	5.9	#	#

[a]Average of left and right angle values taken for lateral flexion
[b]After 800 ms the shoulder markers were hidden from the cameras due to the body bending forward
[c]Forward pitch defined as negative
*0.4 deg or less
#less than 1 deg, greater than 0.4 deg

of 7.5 deg. The six perturbation directions included two that were purely in the pitch plane (forward or 0 deg; and backward or 180 deg in our notation). For the four additional perturbation directions, pitch stimuli were combined with leftward and rightward roll components to form 'forward right' (45 deg), 'backward right' (135 deg), 'backward left' (225 deg) or 'forward left' (315 deg) perturbations. Each of the 6 different combinations of perturbation direction was randomly presented 14 or 15 times throughout the two series for a total of 88 perturbations.

Each perturbation was preceded by a random 5–20 s delay. During this period, subjects were asked to monitor on an oscilloscope the low-pass-filtered AP torque signal, described above. The oscilloscope was located at eye level, approximately 1 m in front of the subject. Using this visual feedback, subjects were required to maintain AP ankle torque within a range of ±4 Nm of their 'preferred-stance' reference value prior to stimulus onset. The 5–20 s interstimulus delay was initiated automatically once the platform had returned to its original level pre-stimulus position and the subject had regained and maintained his preferred vertical position as monitored by AP ankle torque reading. Once the platform moved, the monitoring signal on the oscilloscope was blanked out and subjects were instructed to recover their balance as quickly as possible with in place reactions. Three handrails (generally located 80 cm above foot level but adjustable to the hand height of each subject) were located at a distance of 40 cm to the sides and to the front of the platform centre point. Subjects were informed they were allowed to grasp the handrails if needed. Two assistants (one behind and one to the side of the subjects) were present to lend support in case of a fall (generally only necessary for the full-corset condition).

All EMG and biomechanical recordings were initiated 100 ms prior to rotation onset and had a sampling duration of 1 s. EMG recordings were band-pass analog filtered between 60–600 Hz, full wave rectified, and low pass filtered at 100 Hz prior to sampling at 1 kHz. All biomechanical data were sampled at 500 Hz after passing through anti-aliasing filters (second-order low-pass filters with a cut-off at 53 Hz) and then low-pass filtered off-line at 25 Hz using a zero phase-shift 10th-order Butterworth digital filter.

Data analysis

Following analogue to digital conversion of the data, all biomechanical and EMG signals were averaged offline across each perturbation direction. Zero latency was defined as the first inflexion of ankle rotation velocity after stimulus command onset and did not vary with direction or subject. Subject averages were pooled to produce population averages for a single direction (as shown in Figs. 2, 3, 7). Trunk angular velocity was calculated as the average over the intervals between 90–130 and 180–220 ms for roll and pitch, respectively, that is at the time when these velocities are known to peak, on average, from previous studies (Carpenter et al. 1999). All angular velocity traces (two each for the trunk and arm) were integrated off-line using trapezoid integration to yield angular displacement. The differences between the angle value at 0 ms and 200 ms for roll and 0 ms and 300 ms for pitch, and between 0 ms and 700 ms for both roll and pitch, were employed as measures of the body link angular changes caused by the support-surface rotations. At 200 ms, and 300 ms for pitch, angles normally peaked or did not change further (see Figs. 2, 3). Angular displacements of the arm were calculated relative to the trunk by subtracting the arm position from the trunk position. The roll angle of the arm was measured at the same time as that of the trunk, at 200 ms. Average head roll angular acceleration was calculated over the period 90–110 ms when it first peaked (see Figs. 2, 3). Ankle torque changes were calculated between 160–260 ms, that is shifted 40 ms with respect to EMG measurement periods for balance corrections (see below).

EMG areas were calculated using trapezoid integration within pre-determined time intervals associated with stretch reflex (40–100 ms from stimulus onset), and balance correcting responses (120–220 ms). EMG areas were corrected for background muscle activity (BGA) by subtracting the area due to BGA (measured as the average activity level over the 100 ms period prior to perturbation onset) from the overall EMG response prior to integration.

Our primary statistical analyses concerned between-condition comparisons for stiffness effects due to the corsets. To examine differences between different perturbation directions and between corset conditions, we used an ANOVA model for repeated
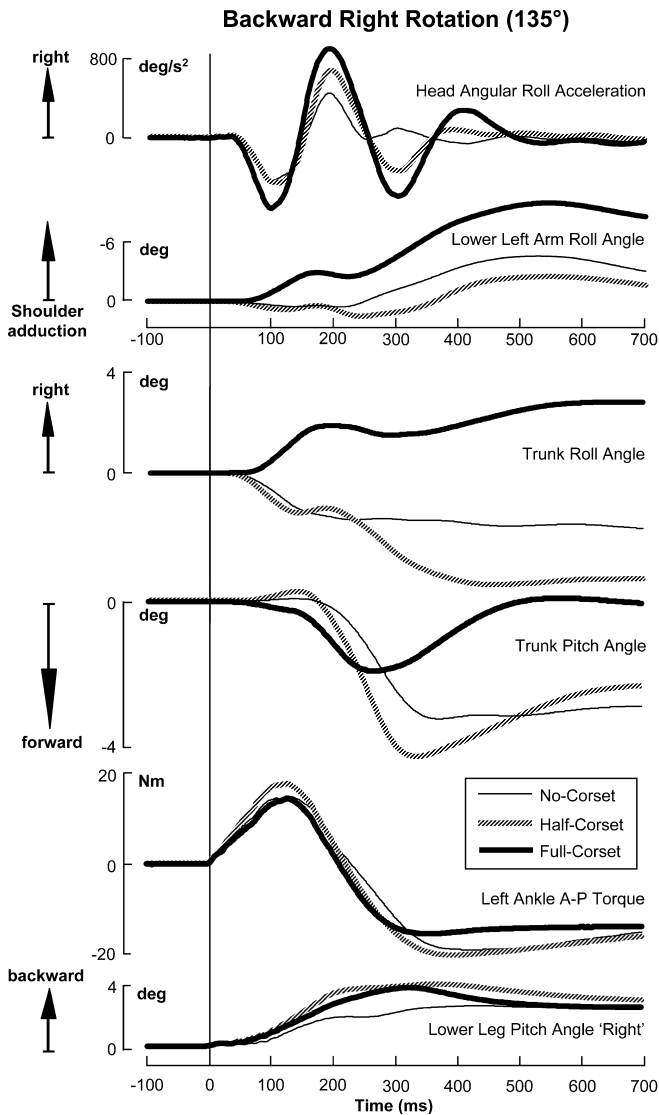
Fig. 2 Biomechanical responses to a backward-right (135 deg) rotation of the support surface. Each of the traces is the average of 5 subjects with 14 responses per subject for a total of 70 responses per condition. The traces corresponding to the three conditions, no corset (normal), half-corset and full-corset, are shown in the *insert*. The onset of the stimulus is shown by the *vertical line* at 0 ms and is aligned with the first inflexion of stimulus velocity. Stimulus termination was at 125 ms, where the initial increase in ankle dorsiflexing torque imposed by the support-surface rotation levelled off. The different directions of initial trunk roll movements but similar directions of initial head roll accelerations are indicated by the *arrows* to the left of the traces

measurements of the subject averages after checking with the Kolmogorov-Smirnov test that the biomechanical and EMG measurements of subject averages were normally distributed. Significant main corset and interaction effects were further explored using post hoc comparisons using *t*-tests with a Bonferroni correction to account for the effect of comparing three conditions at once. ANOVA and Bonferroni test results with $P<0.05$ were considered significant.
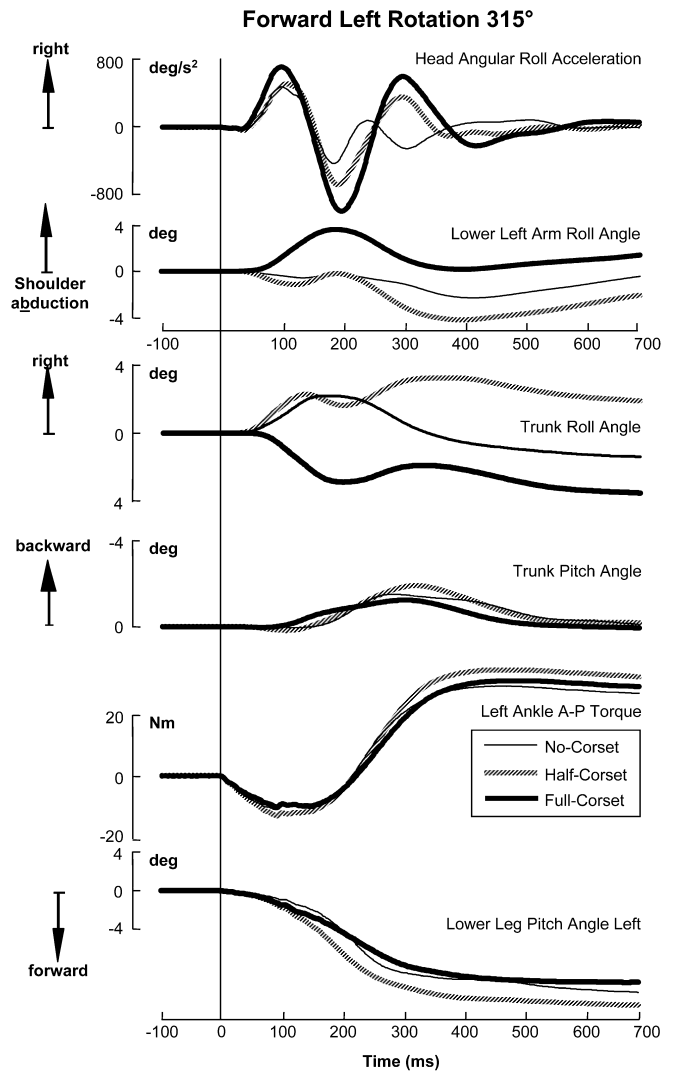

Fig. 3 Biomechanical responses to a forward-left (315 deg) rotation of the support surface. Details are provided in the legend to Fig. 2

Results

The effects on trunk roll and pitch motion caused by mid-body stiffening were different for the half- and full-corset. The influence of stiffening was more profound for perturbations which included a roll component than for pure pitch plane perturbations. Trunk stiffening affected trunk muscles responses more than those of the lower leg. These observations provide the focus of our description of biomechanical and muscle response differences seen with stiffening. Major changes in responses with stimulus direction for the no-corset condition have been reported in detail elsewhere (Carpenter et al. 1999; Allum et al. 2002).

Biomechanical responses

Both corsets altered biomechanical responses, however in a different manner. Figures 2 and 3 show the population

traces for several biomechanical variables measured in the pitch and roll planes. The traces in these figures were elicited in response to a backward-right rotation (direction of 135 deg—Fig. 2) and a forward-left rotation (315 deg—Fig. 3) of the support surface. One major difference with the corsets was noted for the forward pitch of the trunk (Fig. 2). The mean population differences in trunk pitch for all rotations of the support surface with or without a roll component are shown in Fig. 4. For the no-corset condition, the upper body hinges forward about the hip and lumbro-sacral joints following backwards support-surface rotations (Allum et al. 2003). Our measurements at 300 ms revealed a forward pitch of the trunk with the half-corset that was approximately twice that for the other conditions (Figs. 2, 4; $F_{(2,8)}$=17.1,P<0.001). For all backwards tilt directions, forward displacement of the trunk at 300 ms was larger (P<0.05) with the half-corset (Fig. 4). However, by 700 ms the trunk pitch angle with the half-corset was corrected to the inclination recorded with no-corset. In contrast, at 700 ms, the trunk inclination for the full-corset condition was negligible for backward support-surface rotations with respect to the pre-stimulus lean (Figs. 2, 4). Thus, across test conditions, trunk pitch angle at 700 ms varied ($F_{(2,8)}$=11.6,P<0.004). Specifically, for all backwards perturbations, the trunk forward pitch with the full corset was less at 700 ms than for the no- and half-corset conditions (P<0.01, see Fig. 4). No differences in backwards trunk motion were documented across the three test conditions for forward support-surface rotation, with and without a roll component (see Figs. 3, 4). Forward rotations yielded a comparable backward rotation of the trunk at 300 ms and trunk straightening of the trunk by 700 ms (Figs. 3, 4). For these perturbation directions,

most pitch rotation of the trunk is about the knees (Allum et al. 2003).

A second major difference with the corsets occurred for trunk roll motion. Figures 2, 3 and 4 show that trunk roll motion at 200 ms was reversed in direction for the full-corset condition with respect to both other corset conditions. Instead of the normal tilt of the trunk to the left when the support-surface is tilted to the right, the trunk tilted rightwards too with the full corset. Onset of trunk roll motion was also delayed some 25 ms for the full-corset condition (see trunk roll angle traces in Figs. 2, 3). This change in trunk roll direction was also documented using the average trunk velocity between 90 and 130 ms ($F_{(2,8)}$=12.5,P=0.003). This effect occurred for all support-surface rotations with a roll component (see polar plots in Fig. 5) and produced a significant difference in full-corset trunk roll angles at 200 ms with respect to other test conditions (lower part, Fig. 4). The initial reversal of trunk roll motion for the full-corset persisted. Roll amplitude at 700 ms (Fig. 4) remained significantly different from the other conditions (P<0.05).

Despite these significant differences in trunk roll motion, the head was accelerated in the same direction for all three corset conditions (to the left for rightward surface rotations and vice versa; see Figs. 2, 3, 6). However, differences in initial head acceleration (average at ca. 100 ms) were found across test conditions ($F_{(2,8)}$=28.6,P=0.001). Head acceleration was greater with the full corset for all perturbation directions involving a roll component (P<0.01; see Fig. 6).

Lower leg rotations occurred mostly in the first 350 ms (see Figs. 2, 3) after onset of the platform movement (backwards with perturbation rotation backwards and vice versa). We found relatively small differences in pitch

Fig. 4 Mean forward and backward pitch of the trunk at 300 and 700 ms and mean left and right trunk roll angle at 200 and 700 ms after stimulus onset. The *columns* indicate the mean and standard error of mean for different combinations of pitch and roll perturbations as indicated along the *plot abscissas*. Differences between the means are indicated in the *insert* (*wrt* with respect to). Note the highly significant effect of the full-corset condition on trunk angles with respect to the no-corset condition
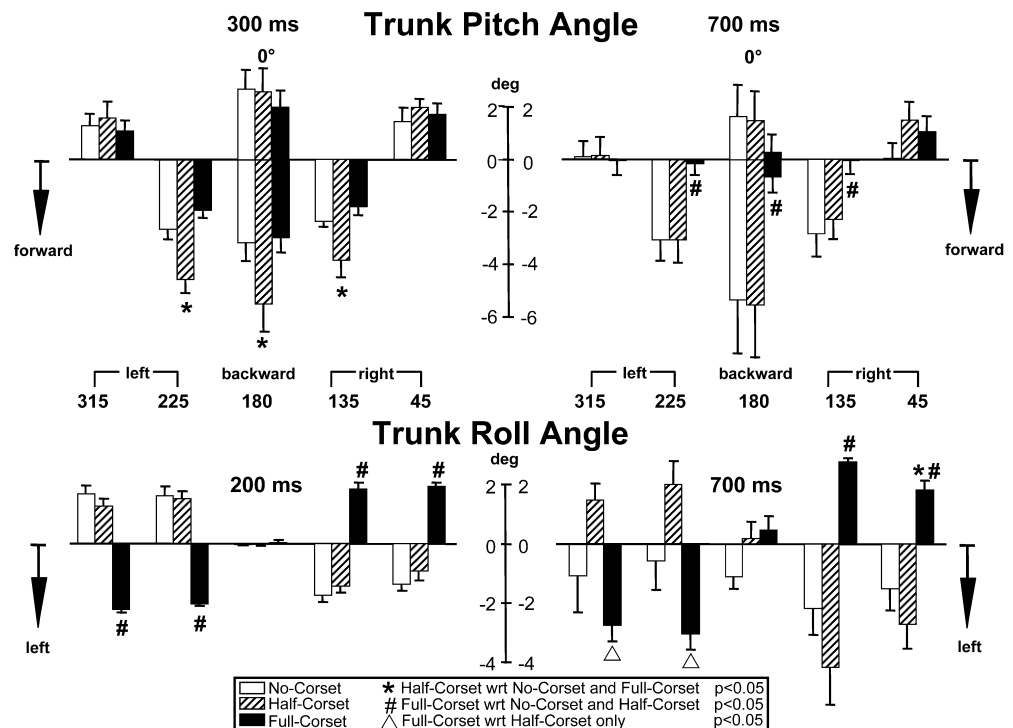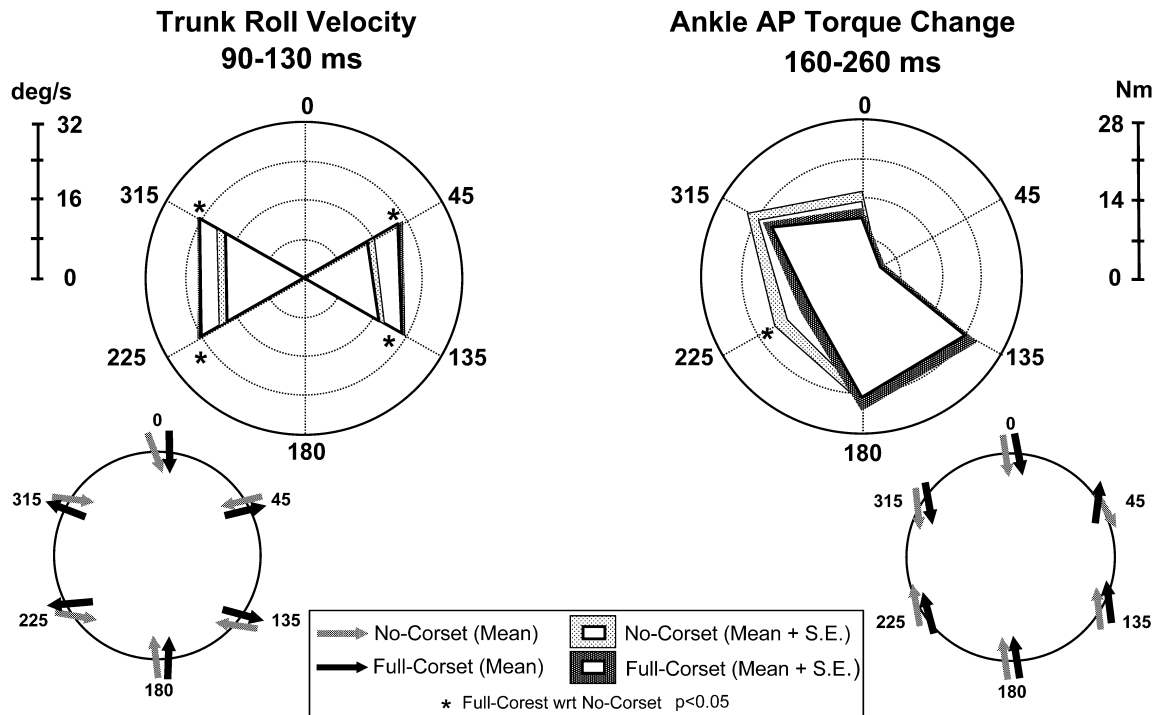
Fig. 5 Polar plots of initial trunk roll velocity and ankle torque AP changes. The mean population amplitude of the responses is plotted as the value along a medial spoke of the plot according to the scales between the plots. The standard error of each mean is shown added to the mean. The directions of the platform perturbations are equivalent to those of the spokes. Amplitudes that are significantly larger for the no-corset compared to the full-corset condition ($P<0.05$) are indicated by *asterisks*. To the left of the polar plot and right of the AP torque trunk roll velocity polar plot, the vector directions of ankle torque and trunk velocity are shown. Note the lack of change in ankle torque direction (except for 45 deg when AP torque is practically zero) but reversal in direction of trunk movement for the full-corset condition for all combined roll and pitch perturbations. Significant differences in amplitude are marked with an *asterisk* on the polar plots

rotations of the lower leg (measured as the change between 100 and 300 ms) across stiffness conditions ($F_{(2,8)}=20.0,P<0.01$). Post hoc comparisons showed a significant reduction ($P<0.05$) with the full corset for the forward left direction (Fig. 3), which also produced the largest left lower leg rotation with no corset.

Consistent with the decrease in lower leg pitch ankle rotation for forward perturbations, and the slight off-pitch directional sensitivity of ankle torque (see polar plot lower right, Fig. 5), left AP ankle torque was smaller with the corsets for forward and left perturbations (Fig. 5). As measured by the torque change between 160 and 260 ms, there was a significant interaction effect between corset condition and the direction of platform rotation ($F_{(10,40)}=3.0,P=0.004$).

Recordings of left lower arm movements indicated major differences across test conditions. The early arm movements with the full-corset, like the trunk movements, were in the opposite direction to the arm movements of the other corset conditions (Figs. 2, 3, 6). The amplitudes of roll arm movements measured at 200 ms, when the initial arm movements plateaued before moving again, were different across test conditions ($F_{(2,8)}=5.3,P=0.04$). Post hoc comparisons showed increased amplitudes for the full-corset following all support-surface rotations with a roll component ($P<0.001$, see Fig. 6).

EMG activity

The stiffening corset-conditions led to reduced amplitudes of balance correcting activity in some muscles. Figure 7 shows an example of the population EMG traces for a backward-right rotation of the support surface. These traces reveal a major effect on trunk muscle activity and a lesser effect on the ankle muscles.

The corsets caused a significant reduction of pre-stimulus background activity (BGA) in tibialis anterior ($F_{(2,8)}=6.9,P=0.02$), soleus ($F_{(2,8)}=6.9,P=0.02$), gastrocnemius medialis ($F_{(2,8)}=5.5,P<0.03$) and peroneus longus ($F_{(2,8)}=4.5,P=0.05$) muscles (see also Fig. 7). BGA in paraspinals was not changed (Fig. 8). The BGA was lowest for the full corset condition in soleus, peroneus longus, and gastrocnemius ($P<0.05$; Fig. 8). In tibialis anterior, the half-corset condition yielded the lowest BGA ($P<0.01$). The tendency for stretch reflex responses to decrease for the full-corset condition (see Fig. 7) in soleus and gastrocnemius muscles was not significant once these response amplitudes were corrected for BGA. The effect of the corsets on balance-correcting responses in ankle muscles (120–220 ms measurement period) varied (Fig. 9). The general trend was a response decrease, when subjects wore the corsets, in ankle plantar-flexor muscles following forward rotations of the support surface. However, in the ankle dorsi-flexors, tibialis
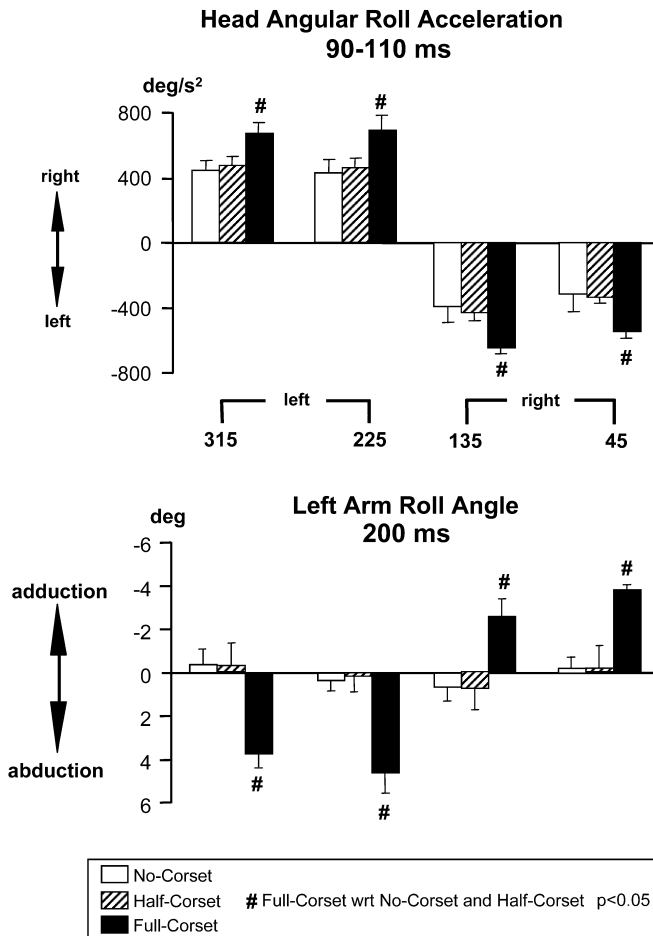
Fig. 6 Mean amplitude of head roll angular acceleration between 90 and 110 ms (*upper row of columns*). Mean amplitudes of arm angles at 200 ms (*lower row of columns*). The *column heights* represent mean population values and the *error bars* the standard error of the mean. A # *symbol* next to a column indicates a significant difference in the full-corset mean with respect to other corset conditions (see *insert*). The directions of the combined roll and pitch perturbations are indicated along the *plot abscissas*. Note the highly significant effect of the full-corset condition on head roll acceleration and arm roll angles

Fig. 7 Leg, trunk and arm muscle responses to a backward right rotation of the support-surface. Details are provided in the legend to Fig. 2

anterior, responses did not change for any perturbation direction (Fig. 9). Balance correcting responses in soleus were decreased for the corset conditions ($F_{(2,8)}=8,4$, $P<0.01$) in particular for forward rotations ($P<0.05$, see Fig. 9). Similar amplitude reductions were noted in gastrocnemius ($F_{(2,8)}=2.3$, $P=0.03$; not illustrated), and in peroneus longus ($F_{(2,8)}=7.8$, $P=0.013$) across almost all directions, except for pure forward perturbations (Fig. 9). The directional sensitivity of ankle muscles, which is normally aligned along the pitch plane (Carpenter et al. 1999), was not changed by the corset conditions (Fig. 9).

Paraspinal balance correcting activity decreased for corset conditions ($F_{(2,8)}=6.7$, $P=0.02$), in particular for all backward tilt directions ($P<0.05$; Fig. 7). Major changes in left-to-right activation ratios of paraspinals would be expected for the half-corset conditions if trunk sway was corrected to that occurring with no corset. For this reason, we examined the ratio between left- and right-sided
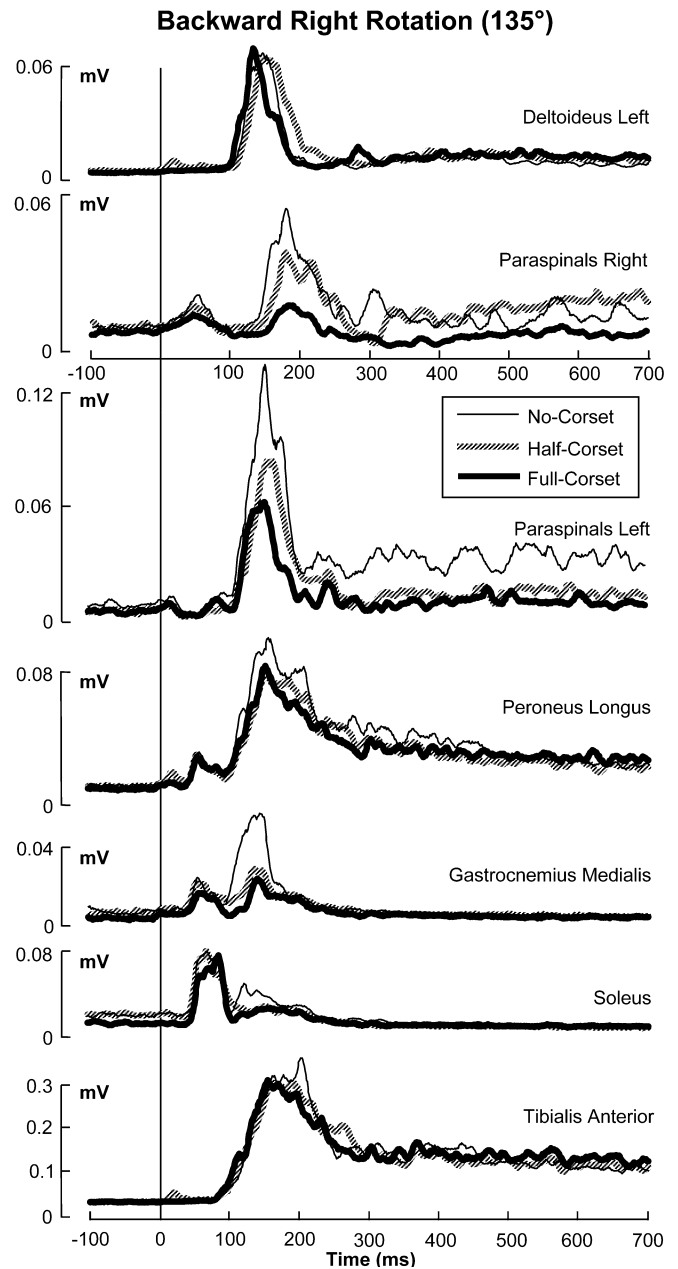
paraspinal balance correcting activity and found that this ratio was preserved with the half-corset (1.5 and 3.2 for backward left and right rotations, respectively) compared to the no-corset condition (1.3 and 2.9, respectively). Thus, despite the reduced paraspinal activity but maintained activation ratio, the trunk still moved more uphill with the half-corset than for the no-corset condition (Fig. 4). Paraspinal activity has presumably little effect on trunk roll during the full-corset condition; hence it is not surprising that responses were reduced for this condition.

Although our recordings were limited to only one arm muscle, deltoid, the slight shift of the peak activity in the deltoid muscle to earlier occurrences for the full-corset
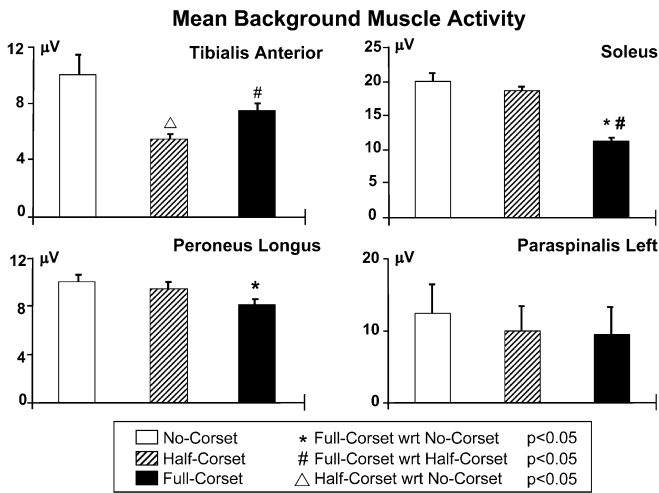
Mean Background Muscle Activity

Fig. 8 Mean background activity in leg and trunk muscles for different corset conditions. The height of each of the columns represents the average background activity measured in the 100 ms period prior to stimulus onset. Error bars of the means are also shown. The general tendency is for the background activity to decrease in the order no-corset, half-corset, full-corset (except for tibialis anterior)

condition is consistent with the larger left arm movements observed for this condition (Fig. 7). Neither the onset of deltoid activity nor its area between 120–220 ms was changed by the corset conditions.

Discussion

We used two different types of stiffening corsets to evaluate postural control and to study how the CNS might compensate for upper body stiffness by adapting muscle synergies and movement strategies. Our main finding was that both types of corset caused profound changes in the profile of trunk movements, indicating that stiffening the hip and trunk of itself increases instability and therefore the likelihood of a loss of balance. The effect was different in the pitch and roll planes, but in both planes little adaptation of muscle synergies to improve stability occurred. Changes in trunk pitch motion induced by backward directed tilts of the support surface (causing backwards perturbations of the centre of mass, COM) were noted even with the half-corset. No changes were noted when the COM was perturbed forwards, suggesting that movements about the hips and knees, and therefore the effect of hip stiffness, play a different role in maintaining balance for backwards and forwards perturbations. The observed changes in trunk movements and resultant instability were greater for the full-corset, compared to the half-corset, particularly in the roll plane, suggesting that flexibility of the entire trunk normally plays a vital role in maintaining balance. The observed reversal of trunk roll motion with the full-corset (but not the half-corset) replicated earlier findings in a total leg proprioceptive loss patient (Bloem et al. 2002) and, to a lesser extent, in healthy elderly subjects (Allum et al.
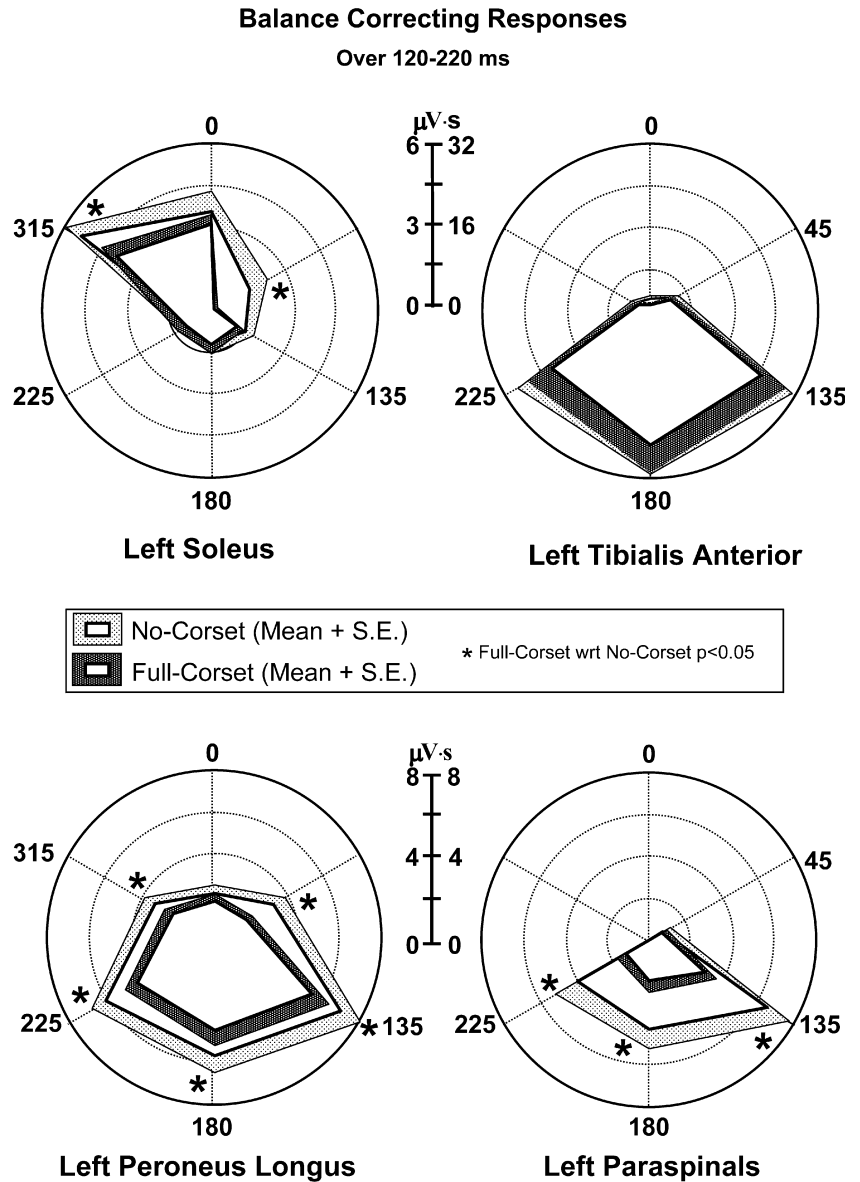
2002) and patients with Parkinson's disease (Carpenter et al. 2004), thereby suggesting that axial stiffness of the entire trunk and pelvis plays a key role in causing instability in these subjects, and perhaps in those normal subjects that stiffen with a fear of falling. The instability that resulted from wearing the full-corset was associated with reversed and larger arm movements in roll, presumably reflecting compensation. These findings will be discussed in more detail below.

Postural control in the pitch plane

For pitch perturbations backwards that rotated the COM backwards too, but the trunk forwards, the amplitude of trunk motion was changed by the corsets, but not its direction. Generally, the full-corset condition caused the greatest instability in our healthy young subjects. However, even the half-corset conditions led to changes in postural corrections. For example, during the half-corset condition, we observed an increased early forward pitch of the trunk following perturbations that induced backward directed instability of the COM. This latter finding can be explained as an increased hinging movement of the trunk about the lumbro-sacral joints due to stiffening of the more caudal hip joints and therefore a more rapid effect of abdominal balance-correcting muscle responses on upper trunk motion. This finding suggests that corrective movements about the hips and lumbro-sacral joints are both an integral part of normal balance-correcting movement strategies, at least for postural corrections in the pitch plane. Normally both early passive hinging and active muscle action underlie forward motion of the trunk. The most rapid forward motion occurs when abdominal muscles are active (Allum et al. 1993). Thus the even more rapid action with the half-corset is most parsimoniously explained by this muscle action not adapting sufficiently to the changed hip stiffness. The hinging movement of the trunk was not seen during the full-corset condition, because now the entire trunk was immobilised. For the full-corset condition, we observed a decreased late forward pitch of the trunk following backward perturbations, and this led to considerable instability. We noted no changes in AP directed ankle torques for backwards perturbations. Taken together, these results indicate that important postural corrections occur at both the hip and lumbro-sacral joints when the COM is perturbed backwards.

The stiffening corsets hardly affected trunk pitch movements following rotational perturbations that caused forward movement of the COM. Indeed, forward perturbations yielded with a comparable backward trunk rotation at 300 ms, and a comparable trunk straightening at 700 ms, no differences for all three test conditions. This can be explained by the fact that such forward falls are mostly absorbed and corrected by flexing movements about the knees and ankles (Allum et al. 2003) and that same ankle torque compensation may have occurred for this direction (Fig. 5). These different findings for

Fig. 9 Polar plots of amplitudes of balance-correcting responses in leg and trunk muscles. Note the lack of changes in tibialis anterior but changed trunk responses for the half- and full-corset condition. The plots are drawn in the same way as those in Fig. 5

Balance Correcting Responses

Over 120-220 ms

μV·s

Left Soleus

Left Tibialis Anterior

No-Corset (Mean + S.E.)

Full-Corset (Mean + S.E.)

∗ Full-Corset wrt No-Corset $p < 0.05$

μV·s

Left Peroneus Longus

Left Paraspinals

backwards and forwards perturbations of the COM may also have some clinical relevance by indicating that trunk stiffness that occurs with, for example, aging (Allum et al. 2002) will mostly lead to backward falls. Such backward falls are a leading cause of wrist fractures in the elderly (Nevitt and Cummings 1993).

Postural control in the roll plane

When a roll component was added to the perturbation, we found not only changes in amplitude of trunk motion, but also directional changes. Similar to the pitch plane perturbations, we again observed an interesting discrepancy between the half- and full-corset conditions. Here, the most salient finding was a reversal of trunk roll motion that occurred during the full-corset condition, but not during the half-corset condition. Thus, with only the pelvis stiffened (half-corset condition), the trunk moved faster in

roll compared to the no-corset condition but the direction of trunk motion was unaltered. The trunk still moved in the opposite direction to that of support surface tilt. With both the pelvis and the trunk stiffened (full-corset condition), the trunk roll was in the same direction as support-surface tilt. This observation has several implications concerning the effect of "tensing up" muscles, be it as the result of a disease state or a fear of falling. The reversal of trunk roll motion during the full-corset condition sheds an interesting new light on our observations of increased background muscle activity both in totally proprioceptive loss patients (Bloem et al. 2002), those with bilateral vestibular loss (Carpenter et al. 2001) and those with Parkinson's disease (Carpenter et al. 2004). We wondered whether this excessive background muscle activity caused increased stiffness of the hips and trunk leading to poor trunk control in the roll planes. This explanation seemed plausible because the reversal in trunk movements occurred very early (some 50 ms) after onset

of the perturbation, and therefore too early to be reflexive in nature (Bloem et al. 2002). Our current findings corroborate this suggestion, as passively induced stiffness during the full-corset condition led to a very comparable and early reversal of trunk movements in the roll plane. It is conceivable that either an increase of passive stiffness or an increase in the background activity or both underlies the reduction in compensatory trunk roll movements that occurs with normal aging (Allum et al. 2002). In some of the elderly, age-related degeneration of joints and ligaments leading to stiffness may play less of a role because a fear of falling also leads to active stiffening due to increases in background muscle activity. In others without a fear of falling, joint stiffness, as in our young subjects, fitted with the corsets, may be the dominant factor. It is also possible the greater reliance of the non-fearful elderly on joint stiffness to control posture leads to lower background muscle activity. We observed a similar phenomenon in these experiments (Fig. 8). These observations may have some clinical relevance. The reversed trunk roll motion, as induced by a stiffening full-corset, forced the COM to be displaced into the same direction as the platform rotation, i.e. into the direction of the impending fall. It is therefore possible that loss of hip and trunk flexibility predisposes subjects to lateral falls, which are commonly associated with hip fractures (Nevitt and Cummings 1993; Greenspan et al. 1998).

Compensatory strategies using the arms

One of our goals was to examine how young healthy subjects would compensate for a sudden loss of intersegmental flexibility. We were particularly interested to examine adaptations in corrective arm movements, as these are an important defence strategy in case of imminent falls (McIlroy and Maki 1995). For the full-corset condition, where instability was greatest, we observed that arm movements in the roll plane were increased in amplitude and reversed in direction. That is, the arms were stretched out "downhill" into the direction of the imposed lateral instability during the full-corset condition, whereas the arms moved "uphill" (away from the imposed lateral instability) during the normal, no-corset, condition. This latter finding replicates our earlier observations in the elderly (Allum et al. 2002) and suggests that young healthy subjects, when they can, use their arms to grasp for "uphill" support and thereby prevent loss of stability. In addition, moving the arms opposite to the perturbation direction helps to keep the COM away from the impending fall direction (Grin 2003). Interestingly, the reversal of arm movements seen with the full-corset condition resembles the pattern seen in healthy elderly persons (Allum et al. 2002). There are at least two possible explanations. First, stiffening the hip joint and forcing the body to move more like an inverted pendulum may create uncontrollable instability unless large compensatory arm movements are used. According to this concept, young but artificially stiffened healthy persons, as

well as elderly persons with age-related loss of trunk flexibility, may sense their lateral instability (caused by abnormally directed trunk movements) and try to minimise the impact of a possible fall by directing their arms into the direction of the presumed impact. Second, arm movements are obviously linked to those of the trunk, and perhaps initially are always similarly directed as the trunk. Indeed, in all our previous studies, the arms always moved in the same direction as the trunk (Allum et al. 2002; Carpenter et al. 2004). It would be interesting to determine if the coupling between movements of these body segments can be dissociated under different experimental conditions.

Lack of leg and trunk compensatory muscle synergies

We also recorded EMG activity to evaluate if some compensation for stiffness occurred in the form of changes in muscle activation patterns. A consistent observation was that balance correcting activity *decreased* during the corset conditions. This pattern differs from the *increase* in late balance correcting activity that we observed previously in the leg muscles of elderly subjects (Allum et al. 2002) and those with vestibular loss (Carpenter et al. 2001). We interpreted this increased balance correcting activity as a compensatory strategy, but why then was this not seen in young-and-artificially-stiffened persons? One possible explanation is that the robust adaptations in corrective arm movements provided sufficient compensation for our current subjects, thereby obviating the need for additional late muscular compensation. Another possible explanation is that short-term exposure to artificial stiffness is not enough to elicit the full repertoire of compensatory strategies that can be seen in persons with "chronic" deficits that have gradually developed over months or years (Allum and Honegger 1998; Bloem et al. 2002). A third explanation is that the proprioceptive receptors at the level of the trunk, which have been assigned a central role in the triggering of postural responses (Bloem et al. 2000, 2002; Carpenter et al. 2001), were hardly stimulated because of the artificial reduction in trunk flexibility. However, one would expect that other sensory systems—such as the vestibular system—might also contribute to generating compensatory muscle synergies. Finally, the single order of testing we used, where full-corset condition was employed last, may have masked some of the adaptation. The lack of a counterbalanced order of testing is in fact a limitation of this study.

When young normals without corsets are suddenly tilted to one side by a support-surface rotation having a roll component, the legs, acting like pistons, push the pelvis in the same direction. In contrast, the trunk hinges at the waist and rotates in the opposite direction to that of the legs and pelvis (Allum et al. 2003). Because the head also moves in the same direction as the trunk, both the head and trunk segments are rotated "uphill", thereby helping to prevent, early on, a possible fall downhill. Because these rotations commence within 30 ms of tilt support surface

(see Fig. 2 and Carpenter et al. 1999) and just prior to the onset of stretch reflexes in hip and trunk muscles (Allum et al. 2002, 2003), we assume these movements represent the mainly passive biomechanical response of the young body to the tilt. The appropriate balance correcting response (to ensure stability is maintained) is to keep the trunk more or less in the same position achieved biomechanically by activating the uphill and unloaded paraspinal muscles more than the downhill and stretched contralateral paraspinal and abdominal muscles. The sensory signals that contribute to triggering and modulating this response synergy are presumably those muscle receptors whose afferents are excited or inhibited by the early stretch and unloading reflexes in paraspinals and gluteus medius (Allum et al. 2003), signals from joint receptors in the spinal column, and vertical semicircular canal afferents responding to the over 400 deg/s^2 angular accelerations of the head (Figs. 2, 3, 6).

When the pelvis was stiffened with the half-corset, and initially moved slightly faster "uphill" than normal, the reduced activation of paraspinal muscles still caused the trunk to move once more uphill after 200 ms (Figs. 2, 3, 4), leading to a large overshoot of the trunk roll profile. It is possible that further reductions of paraspinal response amplitudes might have yielded an identical response to that with no corset. However, such a reduction appeared to be beyond the short-term adaptation capabilities of the CNS.

When the trunk was stiffened in addition to the pelvis with the full-corset, the trunk moved in the downhill direction (Figs. 2, 3, 4, 5). An appropriate response would have been to decrease the activity of the uphill paraspinals even more because such activity with an incompliant trunk would have no or little effect on moving the trunk uphill again. The responses observed in paraspinal responses were decreased even more than with the half-corset (see Figs. 7, 9). Under these circumstances, if the body can be controlled about the ankle joints as an inverted pendulum, leg muscles alone should be used to move the body more uphill to a stable position. In fact, the trunk motion downhill was not compensated by larger laterally acting leg muscle activity (see Fig. 9). Presumably the large inertion of the body acting around the ankle joints cannot be controlled adequately by the small ankle torques available.

Given these limitations of the CNS to generate appropriate balance-correcting synergies, it is instructive to look at the problems that might occur with the integration of sensory signals useful for generating appropriate balance corrections when the trunk is stiffer. The pelvis is driven by the legs to move relative to the trunk. The trunk then hinges about the pelvis and the trunk falls in the opposite direction to pelvis rotation (and support surface rotation). As the trunk gets stiffer, particularly in the thoracic segments, less hinging of the trunk occurs. This effect leads to stretch and unloading responses in paraspinal muscles that are slightly reduced in the elderly (Allum et al. 2002). Depending on the stiffness of the trunk, a cessation in hinging motion occurs and the

trunk reverses in direction. This was observed in the current experiments for the full-corset condition, and also occurs in patients with pathologically stiff trunk motion (Bloem et al. 2002), and in the elderly, for whom the trunk hardly moves at all (Allum et al. 2002). Presumably, flexibility appears to permit more relevant sensory information about trunk motion to reach the CNS. The head, though, continues to move in the same direction it would have had the trunk hinged (Figs. 4, 6) just as it did in the elderly (Allum et al. 2002) albeit with a faster acceleration (Fig. 6) due to a presumed whiplash effect. Thus, two of the sensory signals (muscle proprioception and vestibular afferents) presumably used to generate appropriate balance corrections appear to provide different, even if sufficient, information on the actual trunk movement. It is perhaps the integration of these differing signals that caused problems for young subjects with artificially increased stiffness (and also for the elderly with age-related stiffness) to generate an appropriate roll-stabilising muscle synergy.

Arm movements and trunk motion

When stance is perturbed, arm reactions could provide at least three types of compensatory reactions (McIlroy and Maki 1995; Maki and McIlroy 1997). First to create a righting reaction on the trunk, the downhill arm must be rapidly abducted out in the same direction as the trunk is falling. Second to maintain equilibrium and prevent falling, the uphill arm should be abducted to provide a counterweight to the downhill lean of the legs and pelvis. Third, if equilibrium cannot be regained, the uphill arm will be adducted and downhill arm abducted to grasp a handrail or to buffer the impact of a fall. The first and third functions are complementary as the downhill arm moves in the same direction.

The current results provide insight into the role of arm movements in creating righting torques on the trunk and preparing to cushion a fall. The records of lower-arm angles in Figs. 2 and 3 confirm earlier results seen with normal populations of different ages based on upper arm recordings (Allum et al. 2002). As with the upper arm, the initial lower arm movements appear to be biomechanically driven as these parallel the trunk roll movements. That is, trunk roll right with the full-corset leads to left adducting arm movements following a right tilt and small oppositely directed left arm movements with no-corset or the half-corset (Figs. 2, 3, 6). However, under all corset conditions arm adducting movements followed these initial responses and tended to push the trunk slightly uphill (Figs. 2, 3). Because of the initial arm movement and the large subsequent movements, the arms ended up further downhill and were statically destabilising for the full-corset condition.

The recognition that adequate control over arm and trunk motion is vitally important for maintenance of stable posture may help ameliorate some of the clinical manifestations of balance disorders. For example, pro-

grams to help the elderly develop a more adequate protection against falls by raising the uphill arm rather than moving it downhill as the young did with the full-corset may help to reduce falling tendencies of the elderly. Reduction of trunk stiffness by physical therapy thereby improving trunk flexibility would be another means to achieve adequate balance control in the elderly.

Relationship of biomechanical recordings to models of ankle and hip movement strategies

The control of human posture during quiet standing and when stance is perturbed has often been modelled in a number of studies as an ankle and/or a hip strategy. The ankle strategy assumes that the body's COM is repositioned by moving the whole body as a single-segment, inverted pendulum, about the ankle joints, either by controlling ankle muscle stiffness or by increasing ankle muscle activity and changing the "muscle-spring" offset position (Nashner and McCollum 1985; Horak et al. 1990; Winter et al. 1996, 1998; Morasso and Sanguinetti 2002). The hip strategy involves antiphasic motion about both the ankle and hip joints (Nashner and McCollum 1985; Horak et al. 1990; Winter et al. 1996, 1998; Runge et al. 1999). Extensions of these strategies include a mixed hip-ankle strategy (Kuo and Zajak 1993; Kuo 1995).

Neither the ankle nor the hip strategy for the pitch plane takes knee rotations into account. Indeed, both strategies assume that the knee is locked (Nashner and McCollum 1985). However, knee locking only occurs for backward rotations (Carpenter et al. 1999) but not for forward rotations (Allum et al. 2003) or translations (Allum and Honegger 1998) of the support surface. The effect of ignoring knee flexion changes the mode of control and underestimates the CNS processing used to maintain balance. Thus the fact that we did not find major differences in movement strategies for forward support-surface rotations which induced rearward trunk motion is to be expected because flexions of the knees and ankles are the distinguishing characteristics of the response strategy for this perturbation (Carpenter et al. 1999; Allum et al. 2003). In hindsight we regret not having recorded knee angles in the present study, but we were limited in the number of recording channels available.

In models of hip motion, used to describe the response strategy in response to pitch and roll plane perturbations, the hip-pelvis-trunk complex is reduced to a single hip joint only. Evidence for splitting these into at least two joint complexes emerges from recent studies of body motion after lateral perturbations to the trunk (Winter et al. 1998; Rietdyk et al. 1999). Evidence of antiphasic motion of the pelvis and trunk for roll plane perturbations was postulated based on the gluteus medius and paraspinal stretch reflexes following roll tilt of the support surface (Carpenter et al. 1999; Allum et al. 2002). For example, in response to a tilt of the support surface to the left, stretch reflexes in the right gluteus medius and the left paraspinals were observed. Short latency, unloading responses were observed for both the left gluteus medius and the right paraspinal for the same tilting direction. This reflex activation pattern was a strong argument that the upper-leg, pelvis and trunk are linked by at least two joints and are not as one single joint as in many models of hip motion. Furthermore, our recent recordings of pelvis and trunk motion confirm this antiphasic mode of motion for roll perturbations (Allum et al. 2003). Our present study showed that stiffening only the hip joint with the half-corset caused the trunk to move faster forward about the lumbro-sacral joint following backwards support-surface rotations, or faster laterally following roll support-surface rotations. Thus rotation of the trunk relative to the pelvis is a crucial element of the movement strategy for roll and backward tilts of the support-surface.

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