Effect of Stance Width on Multidirectional Postural Responses

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Received 16 December 1999; accepted in final form 18 October 2000

Henry, Sharon M., Joyce Fung, and Fay B. Horak. Effect of stance width on multidirectional postural responses. J Neurophysiol 85: 559–570, 2001. The effect of stance width on postural responses to 12 different directions of surface translations was examined. Postural responses were characterized by recording 11 lower limb and trunk muscles, body kinematics, and forces exerted under each foot of 7 healthy subjects while they were subjected to horizontal surface translations in 12 different, randomly presented directions. A quasi-static approach of force analysis was done, examining force integrals in three different epochs (background, passive, and active periods). The latency and amplitude of muscle responses were quantified for each direction, and muscle tuning curves were used to determine the spatial activation patterns for each muscle. The results demonstrate that the horizontal force constraint exerted at the ground was lessened in the wide, compared with narrow, stance for humans, a similar finding to that reported by Macpherson for cats. Despite more trunk displacement in narrow stance, there were no significant changes in body center of mass (CoM) displacement due to large changes in center of pressure (CoP), especially in response to lateral translations. Electromyographic (EMG) magnitude decreased for all directions in wide stance, particularly for the more proximal muscles, whereas latencies remained the same from narrow to wide stance. Equilibrium control in narrow stance was more of an active postural strategy that included regulating the loading/unloading of the limbs and the direction of horizontal force vectors. In wide stance, equilibrium control relied more on an increase in passive stiffness resulting from changes in limb geometry. The selective latency modulation of the proximal muscles with translation direction suggests that the trunk was being actively controlled in all directions. The similar EMG latencies for both narrow and wide stance, with modulation of only the muscle activation magnitude as stance width changed, suggest that the same postural synergy was only slightly modified for a change in stance width. Nevertheless, the magnitude of the trunk displacement, as well as of CoP displacement, was modified based on the degree of passive stiffness in the musculoskeletal system, which increased with stance width. The change from a more passive to an active horizontal force constraint, to larger EMG magnitudes especially in the trunk muscles and larger trunk and CoP excursions in narrow stance are consistent with a more effortful response for equilibrium control in narrow stance to perturbations in all directions.

INTRODUCTION

The purpose of this study was to examine how altering stance width affects postural strategies employed in response to multidirectional support surface translations. Much of what is known about postural control is based on surface translations in the anterior/posterior (A/P) direction in a constant stance width. However, activities of daily living involve adapting postural strategies for a variety of stance configurations to maintain equilibrium in multiple directions. During quiet stance, widening bipedal stance base of support increases stability in the lateral, but not necessarily other directions. How are postural responses to perturbations in different directions modified by altering stance width?

Widening stance width increases lateral stability during quiet standing

Kirby et al. (1987) have shown that with feet together, there is significantly more medial-lateral sway in quiet stance compared with other stance widths. Altering the foot position affects the available base of support, changes the relationship of the body center of mass relative to the limits of stability of the feet, and alters the biomechanical stiffness of the musculoskeletal system. Day et al. (1993) demonstrated that in quiet stance, increasing stance width increases stiffness of legs-pelvis and the hipankle coupling. Winter et al. (1996, 1998) showed that stiffness control is modulated at the ankle plantar flexor muscles for sagittal plane motion and hip abductor/adductor muscles for frontal plane motion. Although the effects of stance width or sway in quiet stance have been investigated, the effects of changing stance width on postural responses in humans have not been studied.

Postural strategies can change when initial conditions change

Postural muscle activation patterns are readily modified by changes in initial support conditions. For example, stabilizing posture by grasping a stable object with the upper extremity suppresses muscle activity in the lower extremities (Cordo and Nashner 1982). Additionally, changing the support surface from a flat surface to a narrow beam necessitated a change in the postural strategy such that hip muscle activity, hip torque, hip and trunk flexion, and shear forces at the ground were added in response to the surface translation (Horak and Nashner 1986). Leaning forward or backward prior to an A/P surface translation also increased hip muscle activation and hip flexion excursion, and delayed and suppressed ankle muscle responses (Moore et al. 1986). Furthermore, changes in initial pelvic tilt resulted in changes in the magnitude of the postural responses without changing the temporal aspects of the muscle synergy (Hirschfeld et al. 1997). The initial velocity, ampli-

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Altering fore-hind paw distances from wide to narrow in cats eliminated the force constraint seen in response to horizontal surface translations, suggesting that the force constraint depends on the initial biomechanical constraints as well as active neural control (Macpherson 1994). The force constraint has been defined as a simplifying strategy in which the direction of the horizontal vectors exerted at the ground are limited to only one of two directions and only vector amplitude needs to be modulated to resist perturbations in many directions (Macpherson 1988). By constraining the force output to two directions, the computation for the correct postural response needed to maintain equilibrium is simplified. Changes in electromyographic (EMG) recordings were also reported with changes in stance width in cats such that the amplitude of lumbar muscle activity was increased when the interpaw distance was closer together in the fore- aft direction (Macpherson and Fung 1998). In contrast, for maintenance of human bipedal stance following a surface translation, the amplitude of muscle activation decreased when the limbs were further apart in the lateral direction (Henry et al. 1995).

Fung et al. (1995) demonstrated a similar force constraint in standing humans in response to multidirectional surface translations. Despite the different musculoskeletal constraints of bipedal and quadrupedal stance, both humans and cats exhibit a bimodal force pattern, suggesting that the force constraint is independent of the particular biomechanics of bipedal or quadrupedal stance and is dependent to some extent on active neural control. To date, the effect of changing stance width on the force constraint has not been explored in humans.

The current study was designed to determine whether changing biomechanical constraints by changing stance width eliminates the force constraint pattern exerted at the feet. Previous studies have shown that the postural responses are characterized by a distal to proximal recruitment of muscles with early addition of a proximal muscle, depending on perturbation direction (Henry et al. 1998b; Maki et al. 1994; Moore et al. 1988). The current study examines how postural strategies, characterized by EMG, kinematic, and force response patterns to multidirectional surface translations, are altered by stance width. The results suggest that decreases in the EMG amplitude and relaxation of the force constraint induced by widening stance width can be explained by increases in passive stability of the musculoskeletal system and a decrease in active neural control needed for maintaining equilibrium in response to horizontal translations.

**METHODS**

**Data collection**

Seven healthy subjects (4 female, 3 males; ages 21–41 yr) stood on a movable platform that was under the control of a hydraulic servomotor. Subjects were instructed to stand in a comfortable position with arms crossed, their head facing forward, and with equal weight on each foot placed on separate force plates. At the beginning of the experiment, subjects were asked to lean forward/backward and laterally as far as possible without stepping or losing their balance, and the voluntary maximum A/P and lateral center of pressure (CoP) excursion was noted. Prior to each trial, subjects were instructed to assume the same initial A/P and lateral weight distribution, as monitored by the experimenter. A ramp-and-hold signal was used to translate the platform 9 cm at a peak velocity of 35 cm/s (peak acceleration of 13.5 m/s^2). Subjects received 5 trials, of 3 s duration, in each of 12 different perturbation directions, presented randomly, as specified in polar coordinates (Fig. 1). A surface translation at 0° was a rightward surface translation and the angle increased in 30° increments counterclockwise such that 90° was a forward translation, 180° was a leftward translation, and 270° was a backward translation. The centers of subjects’ heels were placed 10 or 32 cm apart with 10° of toe out to achieve a comfortable natural stance posture with a relatively small base of support or a relatively larger base of support, respectively.

**FIG. 1.** An example of left leg responses to a rightward translation from one subject show an averaged integrated electromyographic (EMG) trace for the left tensor fascia latae (TFL) muscle, averaged force traces from the left force plates in narrow (gray) and wide (black) stance, and platform positions (X, Y). Zero time represents the onset of platform translation. Three epochs were defined for surface force analysis: 1) quiet stance background (50–150 ms before translation); 2) the passive period (50–100 ms after translation onset), indicated by the dashed vertical lines; and 3) the active period (100–300 ms after translation onset) represented by the gray shaded area. Force responses for the right force plates are not shown; however, during the background period, the vertical (Fz) and longitudinal shear (Fx) forces are similar for both the right and left force plates, and the lateral shear (Fy) forces from the right force plate are of equal magnitude but opposite polarity compared with the left force plate (see Henry et al. 1998a, Fig. 1 for a representative example).
The order of stance width was randomly assigned for each subject. The same experiment was repeated on five subjects who returned for a second day of testing. Since there were no differences in the muscle latencies ($0.102 < P < 0.945$) or in maximum displacement of CoP ($0.153 < P < 0.999$) between testing days, data were combined for a total of 10 trials in each of the perturbation directions for each of these 5 subjects.

**Forces**

Two force plates, each with four vertical and two horizontal strain gauge transducers, were mounted within the movable platform. The three components of the ground reaction force ($F_x$, $F_y$, and $F_z$) under each foot were recorded for each trial (Fig. 1). The force signals were amplified and sampled at 120 Hz. The vertical component ($F_z$) was derived by summing forces from four force transducers mounted at the four corners of each force plate. The individual and resultant CoP displacements in both the A/P (CoP$_x$) and lateral (CoP$_y$) directions were calculated as previously described (Henry et al. 1998a).

The longitudinal ($F_x$) and lateral ($F_y$) force vectors were summed to produce the resultant horizontal force vector under each foot. A quasi-static approach (Henry et al. 1998a; Macpherson 1988) was used to calculate the $F_z$, $F_x$, and $F_y$ force integrals in three epochs: 1) during the background quiet stance (50–150 ms before translation onset), 2) during the passive force response period (50–100 ms after translation onset), and 3) during the active force response period (100–300 ms after translation onset; Fig. 1). The window for the passive period (50–100 ms after translation) did not include the force impulse of the platform movement. Considering electromechanical delay, the passive period did not include forces due to active muscle contraction either, since the earliest muscle activation was observed at 90 ms following platform movement. The defined active period included not only force responses due to active muscle contraction, but also forces due to passive viscoelastic properties of the musculoskeletal system.

**Kinematics**

A Hi-Res Motion Analysis System (Santa Rosa, CA) provided three-dimensional spatial coordinate information for calculation of the joint and body segment kinematics. Reflective markers were placed on the left side of the body near the center of joint rotation for the fifth metatarsophalangeal, the ankle, the knee, the hip, and the shoulder joints. Two high-resolution video cameras, sampling at 60 Hz, recorded the displacement of the left side of the body. Symmetry of motions about the body's midline was confirmed by placing markers bilaterally on the five subjects who returned for a second day of testing. Thus in all subsequent kinematic analysis, movement of the right side of the body was extrapolated from the left-sided kinematics.

Segment angles were defined with respect to horizontal in the following manner: the trunk segment was defined by connecting the shoulder and hip marker, the thigh segment by connecting the hip and knee marker, and the shank segment by connecting the knee and ankle marker. A reflective marker was also placed on the moving platform, and the displacement of this marker was subtracted from all other anatomical markers. Moreover, the average joint or segment displacement angles during the background period were subtracted from each respective kinematic joint or segment trace. Thus the kinematic measurements were all referenced to the moving platform and to the standing position during the quiet period.

Twenty-seven anthropometric measures including length and girth...
of limb and trunk segments were taken for each individual subject. These anthropometric data were used together with the kinematic data to calculate the total body center of mass (CoM) position for the shank, the thigh, and the head-arms-trunk segments in the A/P (CoM<sub>x</sub>) and lateral directions (CoM<sub>y</sub>) (Vaughan et al. 1982; Yeadon and Morlock 1989). The body CoM positions (<i>X</i>, <i>Y</i>, and <i>Z</i>) were calculated as weighted sum of all the segments’ CoM positions, using a three link, body segment model (Vaughan et al. 1991).

**EMG**

Bipolar, silver-silver chloride, electrodes with a 20-mm interelectrode distance were placed over the following 11 left-sided muscles for the EMG recordings: tibialis anterior (TIB), peroneus longus (PER), medial gastrocnemius (MGS), soleus (SOL), vastus medialis (VSM), rectus femoris (RFM), adductor longus (ADL), semimembranosus (SEM), tensor fascia latae (TFL), rectus abdominis (RAB), and erector spinae (ESP). A ground electrode was placed over the left medial, proximal tibial plateau. The EMG signals were amplified (×5,000–10,000), band-pass filtered (75–2,000 Hz), full-wave rectified, low-pass filtered at a cutoff frequency of 200 Hz, and then sampled at 480 Hz. Subjects were asked to perform isometric contractions on each of the 11 muscles before the experiment, and the EMG associated with the voluntary muscle activation was monitored on an oscilloscope to ensure that cross talk between the muscles did not exist.

The latency of each muscle burst was identified as the first burst that was greater than 2 SDs above baseline using an interactive software program (Axograph, Axon Instruments, Foster City, CA). The mean baseline was calculated between 50 and 150 ms prior to platform onset. The first point above the mean plus 2 SDs was noted. From this point, the EMG burst was followed back to the mean baseline, and this latency was recorded as the onset of the muscle burst. A muscle had to reach a recruitment probability of 60% (active in at least 6 of 10 trials; or 3 of 5 trials for the 2 subjects who had only 5 trials for each translation direction during 1 day of testing) to be considered physiologically significant in contributing to the postural response.

The mean amplitude of each muscle response was determined by integrating the area under the EMG response during a fixed 200-ms epoch from 70 to 270 ms after the onset of platform motion (Fig. 8A). This epoch precedes and overlaps the active force epoch (100–300 ms after platform onset) to account for the average 30-ms electromechanical delay between muscle onset and force production. The mean background level of EMG activity for the 100 ms prior to platform movement was subtracted. For each subject, the integrals from each muscle were averaged for each set of 5 trials in each of the 12 directions. For each muscle, the mean EMG amplitudes were normalized to the maximum response of the 12 directions and 2 stance widths. The normalized data were then plotted against the direction of translation as muscle tuning curves in polar coordinates to compare EMG modulation across directions, stance width, and across muscles (Fig. 8B).

**Statistical analysis**

A two-way ANOVA (12 translation directions × 2 stance widths), with the <i>P</i> value set at 0.05, was used to examine the effect of stance width across translation directions on the following dependent variables: 1) maximum displacement for the trunk, thigh, and shank
segments; 2) maximum displacement of the total body CoM in the lateral (CoM_L) and anterior/posterior (CoM_A) directions; 3) maximum displacement of the CoP in the lateral (CoP_L) and anterior/posterior (CoP_A) directions; and 4) EMG latencies for translation directions and stance widths in which the muscle was activated for 60% or more of the trials. If there were significant main effects or interactions, pairwise post hoc comparisons (Tukey’s test) were done to determine in which translation directions and/or stance widths the dependent variables were significantly different. To determine whether a force constraint was utilized, as reflected by significant clustering of the horizontal vectors, circular statistics were used to test for the uniformity of distribution of these vectors (Holmquist and Sandberg 1991).

The above experimental protocol was approved by the Legacy Good Samaritan Hospital and Medical Center Institutional Review Board, and all subjects signed a consent form prior to testing.

RESULTS

Changing the stance width had significant effects on the direction and magnitude of the horizontal force vectors during the active period, on the lateral displacement of the CoP, trunk segment displacement in response to lateral translations, and on the EMG magnitudes for all translation directions. The force results will be presented first, followed by the kinematic and EMG data.

Forces

During the active response period (100–300 ms post platform onset), the horizontal force vectors at the ground were clustered in one of two directions in narrow stance (Fig. 2A), regardless of translation direction, similar to the “force constraint” previously described for quadrupeds (Macpherson 1988). In wide stance, however, the force constraint was relaxed such that the horizontal ground reactions force vectors were dispersed in more directions. The polar plot in Fig. 2A shows the pattern of clustering of horizontal force vectors for one individual, and the cartesian plot just below the polar plot reflects the group mean, demonstrating a similar pattern of clustering for all subjects in the narrow stance.

The major change in horizontal vector direction from narrow to wide stance was in response to right- and leftward translations, whereas the vector direction remained similar in response to backward or forward translations. The direction of horizontal force vectors exerted in response to backward and forward translations during wide stance remained the same compared with the direction of vectors exerted in narrow stance (indicated by the letters B and F, respectively in Fig. 2B). In contrast, the horizontal force vectors exerted in response to a right- and leftward translation (indicated by the letters R and L, respectively, in Fig. 2B) during wide stance shifted by one quadrant compared with the direction of horizontal vectors in narrow stance. For both feet, the shift in direction of the horizontal vector in response to a

### TABLE 1. The average horizontal force vector magnitudes during the background period (50–150 ms before platform onset) for each subject in narrow and wide stance

<table>
<thead>
<tr>
<th>Subject</th>
<th>Narrow</th>
<th>Wide</th>
</tr>
</thead>
<tbody>
<tr>
<td>1) GS</td>
<td>18.2</td>
<td>40.1</td>
</tr>
<tr>
<td>2) YY</td>
<td>4.8</td>
<td>30.9</td>
</tr>
<tr>
<td>3) KM</td>
<td>12.2</td>
<td>33.3</td>
</tr>
<tr>
<td>4) PK</td>
<td>20.8</td>
<td>43.0</td>
</tr>
<tr>
<td>5) CB</td>
<td>34.2</td>
<td>34.4</td>
</tr>
<tr>
<td>6) KB</td>
<td>14.1</td>
<td>15.4</td>
</tr>
<tr>
<td>7) TN</td>
<td>10.6</td>
<td>13.3</td>
</tr>
<tr>
<td>Grand average</td>
<td>16.4</td>
<td>30.1</td>
</tr>
</tbody>
</table>

The individual averages from the right and left feet are calculated from 10 trials each for subjects 1–5 and from 5 trials each for subjects 6 and 7.
rightward translation was counterclockwise, whereas in response to a leftward translation, the shift was clockwise when the stance width changed from narrow to wide. The “force constraint” did not disappear altogether in wide stance because the dispersion pattern of the horizontal vectors in the wide stance was not as uniformly opposite to the direction of platform perturbation as compared with the pattern observed during the passive period (compare vector directions in Fig. 2B to those in Fig. 4B).

The magnitude of the horizontal force vectors varied with translation direction and stance width. The largest vectors were in response to lateral translations in wide stance (15.4 N) and in response to A/P translations in narrow stance (14.2 N). The smallest vectors were in response to diagonal translations for both narrow and wide stance (6.4 for wide and 4.6 N for narrow stance). Although force AM varied for wide and narrow stance, the range of horizontal force magnitudes was similar for both feet. For narrow stance, horizontal vector magnitude ranged from 4.6 to 14.2 N for the left foot and 6.7 to 14.1 N for the right foot. In wide stance, the vector magnitude ranged from 6.4 to 15.4 N for the left foot and 9.1 to 15.4 N for the right foot.

The resultant horizontal force vectors combined from both the left and right feet together were directed opposite to each platform translation direction (Fig. 3A). This pattern was similar to the one seen during the passive period (compare vector direction in Fig. 3A to those in Fig. 4B). The results showed that, although the force responses were different for the individual foot during the active period, the summed right and left horizontal force vectors were effective in resisting the perturbation induced by the platform translation. Interestingly, the right- (0°) and leftward translations (180°) were transition directions in which subjects exerted horizontal force vectors either in the 270° direction (3 subjects) or in the 60° direction (4 subjects) in narrow stance. However, the large standard error in horizontal force vector responses observed with a leftward (0°) translation during wide stance (Fig. 3B) was due to random inter-subject variability; the responses were clustered at either 0 or 360°.

The magnitude of the resultant horizontal vectors was modulated with translation direction in narrow stance, ranging from 5.9 N for lateral translations (0 and 180°) to 22.0 N for backward translations (270°; Fig. 3B). In contrast to narrow stance, in wide stance, the resultant horizontal vector magni-
tudes were larger (19.7–25.2 N) and were not modulated with translation direction. The pattern of vector magnitude modulation in narrow stance was due to the significant decrease in the magnitude of the resultant horizontal vectors exerted in response to rightward (0°) and leftward (180°) translations. This modulation was primarily due, in turn, to the decrease in magnitude of $F_y$ in response to lateral translations (refer to $F_y$ in Fig. 2, A and B).

In contrast to horizontal forces, the vertical forces under each foot were not significantly different from narrow to wide stance ($P < 0.01$). The vertical forces were modulated in a similar pattern in narrow and wide stance such that the change in vertical forces was the greatest in response to lateral translations where one leg was loaded while the other leg was unloaded (data not shown). The change in vertical force under each foot was the smallest in response to A/P translations since there was no loading/unloading response of the right/left legs, although there was a loading/unloading of the forefoot and hindfoot.

During the background period (quiet stance 50–150 ms prior to onset of platform translation), the horizontal force vectors were
directed laterally and were significantly larger in magnitude by an average difference of 13.7 ± 11.6 N (SE) in the wide stance compared with narrow stance (P < 0.05; Fig. 4A; Table 1). There were three subjects (CB, KB, and TN) who did not change the amount of horizontal force exerted during the background period when changing stance width. During the background period, CB exerted a much higher horizontal force in narrow stance, whereas KB and TN exerted a much lower horizontal force in wide stance compared with the other subjects.

During the passive period (50–100 ms post onset of platform translation), the horizontal vectors, due primarily to the passive viscoelastic properties of the musculoskeletal system, were directed opposite to the platform translation direction and similar in magnitude for both narrow and wide stance (Fig. 4B). The grand average for the passive horizontal vector magnitude across all subjects in narrow stance was 13.9 ± 2.9 N and in wide stance was 14.7 ± 1.8 N, which was not significantly different (P > 0.05). The passive horizontal vector magnitude and direction did not differ for the right and left foot.

**CoM and CoP**

The maximum displacement of the body CoM in the lateral (CoM_y) and anterior/posterior (CoM_x) directions was not significantly different for any translational direction or for narrow versus wide stance (5.1 ± 0.10 cm for CoM_x and 4.5 ± 0.07 cm for CoM_y; Fig. 5). In contrast, the maximum displacement of the CoP in the lateral (CoP_y) direction was significantly greater in narrow, compared with wide, stance (P < 0.001–0.0083) for all translations except the following directions: forward (90°), forward/leftward (120 and 150°), and backward (270°). For example, in response to a lateral translation (180°), the average maximum CoP_y displacement was 13.7 ± 0.2 cm in narrow stance, whereas it was only 8.2 ± 0.3 cm in wide stance. The maximum displacement of CoP in the anterior/posterior (CoP_x) direction was not significantly different for any translational direction for both stance widths. For example, in response to an anterior translation (90°), the average maximum CoP_x displacement was 7.2 ± 0.2 cm in narrow stance, and it was 6.4 ± 0.2 cm in wide stance.

**Kinematics**

Corresponding to the large CoP_y displacements in narrow stance, the kinematic response to lateral translations revealed significantly larger frontal plane trunk displacements (8–10° in narrow vs. 2° in wide for lateral translations, for example; P < 0.001; Fig. 6A). There were no significant differences in maximum sagittal displacement of the trunk (Fig. 7) or in frontal or sagittal plane displacement of the shank or thigh segments (data not shown in Fig. 7) in response to any translational direction or between stance widths. As previously reported for

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**Table 2. Muscle latencies in narrow and wide stance for each left-sided muscle**

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Narrow</th>
<th>Wide</th>
</tr>
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<tbody>
<tr>
<td>LTIB</td>
<td>129 ± 2</td>
<td>130 ± 2</td>
</tr>
<tr>
<td>LMGS</td>
<td>118 ± 2</td>
<td>122 ± 3</td>
</tr>
<tr>
<td>LSOL</td>
<td>115 ± 3</td>
<td>124 ± 4</td>
</tr>
<tr>
<td>LPER</td>
<td>131 ± 5</td>
<td>136 ± 4</td>
</tr>
<tr>
<td>LVSM</td>
<td>140 ± 4</td>
<td>147 ± 5</td>
</tr>
<tr>
<td>LSTM</td>
<td>154 ± 7</td>
<td>156 ± 7</td>
</tr>
<tr>
<td>LRFM</td>
<td>143 ± 8</td>
<td>133 ± 7</td>
</tr>
<tr>
<td>LADL</td>
<td>145 ± 5</td>
<td>151 ± 4</td>
</tr>
<tr>
<td>LTFL</td>
<td>105 ± 2</td>
<td>108 ± 2</td>
</tr>
<tr>
<td>LRAB</td>
<td>127 ± 7</td>
<td>128 ± 9</td>
</tr>
<tr>
<td>LESP</td>
<td>124 ± 5</td>
<td>134 ± 7</td>
</tr>
</tbody>
</table>

Values are means ± SE. The latency reported is for the direction of maximal activity for each muscle (indicated in bold). The direction of maximal activity and the other directions listed are those directions in which the muscle was active in at least 60% of the trials for each subject, and these respective latencies were entered into the analysis of variance that revealed no significant difference between narrow and wide stance (P > 0.28–0.68). LTIB, tibialis anterior; LMGS, medial gastrocnemius; LSOL, soleus; LPER, peroneus longus; LVSM, vastus medialis; LSTM, semimembranosus; LADL, adductor longus; LTFL, tensor fascia latae; LRFM, rectus femoris; LRAB, rectus abdominis; LESP, erector spinae.
narrow stance (Henry et al. 1998a), the kinematic response is characterized by a distal-to-proximal progression of body segment movement, such that the trunk was the last to return to a vertical orientation. An example of the body kinematic response in narrow and wide stance to a rightward (D0) translation can be seen in the stick figures in Fig. 6B. The same distal to proximal pattern of segment displacement was observed when subjects stood in the wide stance, although the magnitude of frontal plane trunk displacement was less.

The kinematic responses to backward translations were similar to lateral translations in that the trunk remained relatively erect initially as the shank and thigh were displaced with the platform (Henry et al. 1998a). As the shank and thigh began their correction, the trunk began to flex opposite to platform movement. Kinematic responses to anterior translations resulted in slightly smaller displacements of the trunk and larger displacements of the thigh, but the pattern was similar for narrow and wide stance.

**EMG**

Although the forces at the ground and the trunk kinematic responses were somewhat different from narrow to wide stance, the temporal and spatial muscle activation patterns remained very similar between the two stance widths.

The muscle latency and sequencing in narrow stance has previously been reported in detail (Henry et al. 1998b). Interestingly, changing stance width from narrow to wide did not significantly change the latencies or recruitment of muscles ($P > 0.28–0.68$; Table 2). Previous studies showed that the latencies of three proximal muscles (TFL, RAB, and ESP) were modulated depending on translational direction, whereas the latencies of the shank and thigh muscles were not (Henry et al. 1998b). Increasing stance width did not alter the latency modulation of the three proximal muscles (TFL, RAB, and ESP) or the fixed latencies of the shank and thigh muscles.

Although the latencies and directions of muscle activation did not vary with stance width, all muscles had a larger magnitude of activation in narrow stance compared with wide stance for all directions (Fig. 8B). On average, the magnitude of muscle activation was 12–64% more in narrow, compared with wide, stance; and the mean difference in magnitude of activation was greater for the more proximal muscles compared with the distal muscles (Table 3). In addition, most muscles had one direction of maximal activity, and it was the same direction for narrow and wide stance. The muscles were generally maximally active in response to diagonal translations, with the exception of TFL and RFM, which were maximally active in response to diagonal translation, with the direction of maximal activation being the same for narrow and wide stance.
Values are means ± SE. The grand mean percent difference in the electromyographic (EMG) integral from narrow to wide stance was derived by 1st calculating the difference between the mean integrals in narrow and wide stance for the direction of maximal activity for each subject and each left-sided (L) muscle. The difference was then divided by the mean integral in narrow stance and multiplied by 100 to obtain a percent difference for each subject and muscle. The grand mean percent difference in the EMG integral is the average of 7 subjects. See Table 2 for abbreviations.

**Table 3. Difference in EMG integral from narrow to wide stance**

<table>
<thead>
<tr>
<th>Anatomical Location</th>
<th>Muscle</th>
<th>Grand Mean %Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distal, 30 ± 7.6%</td>
<td>LTIB</td>
<td>22 ± 27</td>
</tr>
<tr>
<td></td>
<td>LMGs</td>
<td>35 ± 16</td>
</tr>
<tr>
<td></td>
<td>LSOL</td>
<td>40 ± 22</td>
</tr>
<tr>
<td></td>
<td>LPER</td>
<td>24 ± 27</td>
</tr>
<tr>
<td>Intermediate, 33 ± 16.3%</td>
<td>LVSM</td>
<td>12 ± 34</td>
</tr>
<tr>
<td></td>
<td>LSTM</td>
<td>30 ± 43</td>
</tr>
<tr>
<td></td>
<td>LRPF</td>
<td>58 ± 37</td>
</tr>
<tr>
<td></td>
<td>LADL</td>
<td>32 ± 51</td>
</tr>
<tr>
<td>Proximal, 53 ± 12.6%</td>
<td>LTLFL</td>
<td>65 ± 20</td>
</tr>
<tr>
<td></td>
<td>LRAB</td>
<td>35 ± 55</td>
</tr>
<tr>
<td></td>
<td>LESP</td>
<td>58 ± 37</td>
</tr>
</tbody>
</table>

The change in the pattern of the horizontal vectors in wide stance in bipedal humans occurred due to the change in direction in which the horizontal vectors were exerted in response to lateral, but not A/P, translations (refer to Fig. 2, A and B). The shift in vector direction in response to lateral translation for wide stance may be due to the increase in lateral stability that accompanies a wider stance (Winter et al. 1998). Since the base of support in the A/P direction was not altered when separating the feet in the frontal plane, it is not surprising that the horizontal force vector direction did not change in response to A/P translations.

The force constraint observed in narrow stance may reflect a means for simplifying the control needed to maintain equilibrium in response to perturbations (Fung et al. 1995; Macpherson 1988). It has been suggested that the utilization of the force constraint in cats may be useful in controlling the trunk, thus ensuring a linear translation of the body CoM in recovering balance (Macpherson 1994). In humans during narrow stance, the horizontal force constraint may be useful in countering rotary movements of the trunk that accompany lateral trunk flexion (Henry et al. 1998a). Narrow stance perturbations result in larger lateral flexion of the trunk that is accompanied by trunk rotation and flexion, and may be related to coupling of rotation in active, purposeful trunk motion (Pearcy and Tibrewal 1984). In wide stance, due to the increased stability and passive stiffness of the musculoskeletal system, there is less trunk motion to control and thus a relaxation of the horizontal force constraint. Thus equilibrium in wide stance can primarily be restored with hip abduction/adduction mechanisms (Winter et al. 1998).

**CoP control of CoM displacement**

Although there were changes in the lower limb biomechanics resulting in changes in the horizontal force constraint between narrow and wide stance, the body CoM was displaced a similar amount in both stance configurations for all translation directions. This strict limit of maximum CoM displacement regardless of biomechanical constraints associated with displacement direction or stance width suggests that body CoM is a highly regulated CNS variable (Massion 1992, 1994).

In contrast to CoM, there were large changes in CoP, particularly in response to lateral translations. This result suggests that active force (effort) was larger in narrow stance width to limit CoM displacement to keep it within the base of support. Equilibrium control in narrow stance required more of an active postural strategy that included regulating left/right foot loading/unloading and the direction of horizontal force vectors that effectively regulated CoP position. Similar results have also been reported for CoM displacements in zero gravity under water and in 1 × G conditions (Massion et al. 1995) during voluntary trunk bending, supporting the hypothesis that CoM is highly regulated.

Large left/right foot loading/unloading responses occurred with lateral translations in narrow stance creating a large shift in the CoP, which, in turn, created the large corrective moment about the body CoM needed to restore equilibrium (Fung et al.
suggested a shift from more active control of the trunk in narrow stance to more passive control in wide stance.

It is interesting to note that the EMG magnitude decreased in wide stance for all translation directions, including the A/P direction, indicating less need for active neural control for sagittal plane stability. It is unclear how increasing frontal plane stability by widening stance laterally would affect sagittal plane stability, but perhaps there is an interaction between planes such that once the body is more stable in the frontal plane, control in the sagittal plane motion is more effective (Day et al. 1993).

Postural control strategies

We propose that a similar postural control strategy is being utilized to maintain bipedal stance under the two different stance conditions. The evidence for this lies in the similar EMG latencies for both narrow and wide stance, with modulation of only the muscle magnitude as stance width changed. However, the magnitude of the trunk displacement, as well as the CoP displacement, was modified based on the degree of passive stiffness in the musculoskeletal system, which increased with stance width and which has been shown to contribute to equilibrium control (Rietdyk et al. 1999). The end result was effective regulation of equilibrium or body CoM such that the CoM was not displaced to any greater degree in narrow than wide stance. CoM control in narrow stance required more of an active postural strategy that included regulating the loading/unloading of the limbs and the direction of horizontal force vectors. In contrast, in wide stance, increase in passive stiffness as a result of limb geometry allowed for CoM control with a combination of smaller active and larger passive mechanisms.

Given that we tested only two stance widths in the current study, we cannot determine the boundaries for the horizontal force constraint, for active trunk control or for changes in muscle synergies. Future studies will examine multiple velocities, accelerations, and stance widths, including tandem stance, in multidirectional horizontal translations to search for thresholds at which there might be a distinct postural strategy change.

The authors thank M. Richard for technical assistance in data analysis and J. Macpherson, Ph.D. for contributions to this manuscript through numerous scientific discussions.

This research project was supported by National Institutes of Health (NIH)/National Aeronautics and Space Administration (NASA) Grant P60DC-02072 and NIH Grant AG-06457 to F. Horak. S. Henry was supported by postdoctoral fellowships from NIH (1F32AG/HD-05738) and the American Physical Therapy Association, and by NIH/National Center for Medical Rehabilitation Research Grant KO1HD-01194. J. Fung was supported by a fellowship from the Rick Hansen Foundation and NIH/NASA.

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