Postural Control for Initiation of Lateral Step and Step-up Motions in Young Adults

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Abstract. The purpose of this study was to clarify the postural control of the initiation of lateral step and step-up motions in healthy young adults (24.3 ± 1.8 years: mean ± SD). The tasks involved the lateral step (step lengths, 10 cm and 20 cm) and lateral step-up motions (10-cm high stool; step lengths, 10 cm and 20 cm). The variables for analysis included motion duration (weight-shift phase and swing phase), shifts of the center of pressure (CoP) and the center of gravity (CoG), displacement and inclination of the shoulder and the pelvis, and root mean square electromyographic amplitude of the erector spinae, gluteus medius (GM) and adductor longus (AL) bilaterally. The CoP shift toward the stepping side was larger in the step-up task than in the step task, and it was smaller when both tasks were performed in the long lengths. The CoG shift toward the supporting side and the displacements of the shoulder and pelvis were larger in the step-up task than in the step task. However, the magnitude of GM activity of the supporting leg was larger in the step task than in the step-up task, and increased when both tasks were performed in the long lengths. In the stepping leg, the magnitude of AL activity was larger in the step-up task than in the step task, and decreased when both tasks were performed in the long lengths. These results suggest that GM activity of the supporting leg and AL activity of the stepping leg control the shift of CoG in the frontal plane. Our results indicate that the increase in GM activity does not depend on the height of stepping but on the length of stepping in lateral step and step-up motions.

Key words: Step, Postural control, Motion analysis

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INTRODUCTION

Postural control for the stability and orientation of the body requires complex interactions between musculoskeletal and neural systems1). Since problems in these systems can interfere with postural stability, postural control should be evaluated in patients with functional disorders. Much research has been carried out to quantify human postural control in the standing position using various tasks, including those under resting, perturbed, sensory-conflicted, and walking conditions2, 3).

The examination of the coordination between posture and voluntary movement during gait initiation4–7), single-leg flexion8, 9), and lateral leg raising10–12) has revealed a dynamic component of balance. In each of these prior studies, anticipatory postural adjustments before foot-off were described. One of the phenomena observed in the studies of gait initiation is the shift of the center of pressure (CoP) toward the stepping leg, and this
characteristic CoP shift initiates the center of gravity (CoG) shift towards the supporting leg so that the stepping leg can raise and the equilibrium maintained during the movement. The analysis of the initiation of walking or stepping has provided insights into the mechanism underlying motor control from a double-to a single-limb support for clinicians engaged in therapy for disorders of postural control.

The stepping motion is one of the motor strategies used to recover stability. When in-place strategies such as the ankle and hip strategies are insufficient for recovering balance, a step or a hop (the stepping strategy) is used to realign the support base under the CoG\(^1\) Therefore, the authors adopted a stepping motion as the task in this study.

Sims and Brauer\(^13\)) compared the CoP motion and the activity of the hip abductor muscles in the supporting leg during the forward step-up and forward step tasks, and showed that the forward step-up task required a greater effort to control mediolateral (ML) postural stability and a larger magnitude of gluteus medius activity than the forward step task. The control of ML postural stability deteriorates in the elderly\(^14,\) 15), and it was found to be a predictor of a risk of falling in a prospective study\(^16\). Sims and Brauer\(^13\)) also examined ML postural stability in the step-forward task. However, few studies on the analysis of the detailed motions involved in lateral step and step-up have been performed\(^17\). Does activity of the hip abductor muscles increase more in the lateral step-up than in the lateral step, as in the forward step and step-up tasks?

The purpose of this study was to clarify the postural control of the initiation of the lateral step and the step-up motions in healthy young adults from the viewpoint of the interactions between CoP, CoG and muscle activity.

**SUBJECTS AND METHOD**

**Subjects**

Ten healthy female adults aged 24.3 ± 1.8 (mean ± SD) years, with a height of 160.8 ± 4.0 cm and a body weight of 54.0 ± 7.4 kg participated in this study. Informed consent was obtained from all participants.

**Task and protocol**

The subjects were instructed to stand barefoot on a dual force platform system (Advanced Mechanical Technology Inc., AMTI) with one foot on one force platform, with the medial malleoli 10 cm apart, the toes of each foot pointing 15 degrees outward, and the hands behind the back (Fig. 1a). The movement during each trial started upon the presentation of a visual cue using two (right and left) light-emitting diodes set in front of the subject. The illumination of one of the diodes indicated to the subject which leg to step in the lateral direction. To avoid an anticipatory weight shift, the order in which the diodes illuminated was randomized.

In the step task (Fig. 1b), the subject was instructed to step in a short length (between 10 and 20 cm from the fifth metatarsal head of the stepping foot; S10) or a long length (between 20 and 30 cm from the fifth metatarsal head of the stepping foot; S20). In the step-up task (Fig. 1c), the subject was instructed to step with the leg raised onto a 10-cm high stool placed a short length (SU10) or a long length (SU20) on the lateral side. Several trials were performed to familiarize the subjects with the experimental protocol. The subjects were instructed to perform each task as fast as possible, and to maintain the final position for at least three seconds. The order of these tasks was randomized. The means and standard deviations were determined using data from three trials.

**Measurements**

Kinetic, kinematic, and electromyographic (EMG) recordings were performed. The ground reaction force was measured using two force platforms at a sampling frequency of 1,080 Hz. Kinematic measurements were performed using VICON 460 (Oxford Metrics Ltd.) with six cameras operating at a sampling frequency of 120 Hz. Ten light-reflecting markers were bilaterally attached to anatomical landmarks: the acromion, anterior-superior iliac spine, greater trochanter, lateral femoral condyle, and lateral malleolus (Fig. 1).

EMG recordings (MyoSystem 1200, Noraxon USA Inc.) were bilaterally performed on the erector spinae (ES), gluteus medius (GM), and adductor longus (AL) using bipolar, silver-silver chloride disposable surface electrodes (BLUE SENSOR, M-00-S, Medicotest, Denmark). The electrodes were longitudinally placed along the muscle fibers and were spaced 2 cm apart. The ground electrode was affixed to the patella. EMG signals were amplified, band-pass-filtered at 10–500 Hz, sampled at 1,080
Variables for analysis

(1) Motion duration
Two motion durations defined on the basis of kinetic parameters were examined (Fig. 2). 1) The weight-shift phase: the period from the onset of weight shift to foot-off. The onset of weight shift was determined as the first observed change of CoP in the frontal plane. 2) The swing phase: the period from foot-off to foot-on. The onset of CoP change was determined to have occurred when amplitude exceeded three standard deviations from the mean amplitude of the baseline period (1,500 ms period prior to the onset of the visual cue) for more than 50 ms. The foot-off can be identified from the kinetic recording and was defined as the decrease in loading on the force plate under the stepping leg to less than 10 N force. The foot-on was defined as the increase in loading on the force plate to the same extent (10 N).

(2) Kinetic variables
The initial CoP position was calculated from the force-plate recordings during the baseline period. The peak amplitude of CoP shift toward the stepping side (CoP-step) and that toward the supporting side (CoP-support) during the weight-shift phase were calculated from the initial CoP position (Fig. 2). The displacements of CoP were standardized by the size of the individual base of support. The base of support was defined as the length between the left and right fifth metatarsal heads in the initial position.

The shift of the CoG in the frontal plane was calculated from the lateral ground reaction force. The shift of CoG can be obtained using double integration of the acceleration. However, this technique has a disadvantage in that very small errors in force recordings are amplified by the double integration, giving rise to the so-called integration drift. Therefore, to correct the acceleration trace, we adopted the method developed by Lyon and Day. The peak amplitude of CoG shift toward the supporting side (CoG-support) during the weight-shift phase was calculated from the initial CoG position (Fig. 2).

(3) Kinematic variables
The peak amplitudes of the markers placed on the acromion and greater trochanter of the supporting side were measured during the weight-shift phase. The displacements were standardized by the subject’s height.

The following two angles on the frontal plane were computed the shoulder angle between the shoulder line (joining the bilateral acromion markers) and the horizontal plane; and the pelvic angle between the pelvic line (joining the bilateral anterior superior iliac spine markers) and the horizontal plane. The variations in the two angles during the weight-shift phase were calculated from the initial position. Because the motions were performed in the frontal plane, only this plane was considered in the analysis.

(4) EMG
EMG signals were analyzed by using MyoResearch software (Noraxon USA Inc.) The root mean square (RMS) amplitude of EMG signals
was computed for the period from the onset of weight-shift toward the supporting side to foot-off. The onset of weight-shift toward the supporting side was defined as the time at which the shift of CoP toward the supporting side was initiated. RMS amplitude was standardized to the percentage of the maximal voluntary isometric contraction measured by an examiner in the following positions: trunk extension (prone position), hip abduction and adduction (supine position, the hip was in the neutral position).

(5) Statistical analysis
The two-way repeated measures ANOVA (step height (step vs. step-up) × step length (10 cm vs. 20 cm)) was used in the statistical analysis of all variables. Correlation between CoP-step and the RMS amplitude of EMG in the four tasks was determined using Pearson’s correlation coefficients. A probability of p<0.05 was considered statistically significant.

RESULTS
No interaction effects were detected between step height and step length in all variables.

In the motion duration, the swing phase was longer in the step-up task than in the step task (p<0.01, Table 1), and it was smaller when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.01, Table 1). No difference was found in CoP-support (Table 1). The CoG-support was larger in the step-up task than in the step task, and it was smaller when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.05, Table 1). We also found that subjects moved the CoG directly toward the stepping side in S10 and S20 (Table 1).

The kinetic analysis revealed that CoP-step was larger in the step-up task than in the step task (p<0.01, Table 1), and it was smaller when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.01, Table 1). No difference was found in CoP-support (Table 1). The CoG-support was larger in the step-up task than in the step task, and it was smaller when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.05, Table 1). We also found that subjects moved the CoG directly toward the stepping side in S10 and S20 (Table 1).

The displacements of the acromion and greater trochanter to the supporting side during the weight-shift phase were larger in the step-up task (p<0.01, Table 1), and the greater trochanter tended to move toward the stepping side in the step task. The displacement of the greater trochanter toward the supporting side was smaller when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.01, Table 1). Although the shoulder angle was larger when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.01, Table 1), the shoulder and the pelvic angles were not significantly different between the two tasks (Table 1).

The magnitude of GM activity of the supporting leg was smaller in the step-up task than in the step task, and was larger when the step and step-up tasks were performed for the long lengths (S20 and SU20,
p<0.01, Table 1). There was no difference in the AL activity of the supporting leg (Table 1). In the stepping leg, the magnitude of AL activity was larger in the step-up task than in the step task, and was smaller when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.05, Table 1). Also, the magnitude of GM activity was larger when the step and step-up tasks were performed for the long lengths (S20 and SU20, p<0.01, Table 1).

The magnitude of GM activity of the supporting leg negatively correlated with CoP-step (r=–0.63, p<0.01, Fig. 3). The magnitude of AL activity of the stepping leg weakly correlated with CoP-step (r=0.36, p<0.05).

**DISCUSSION**

It was reported that GM activity of the supporting leg is larger in the forward step-up task than in the forward step task. In this study, the GM activity of the supporting leg was smaller in the step-up task than in the step task; therefore, we consider that the lateral step and step-up motions are kinesiologically

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**Table 1** Variables during lateral step and step-up tasks (mean ± SD)

<table>
<thead>
<tr>
<th></th>
<th>Step 10 cm</th>
<th>Step 20 cm</th>
<th>Step-up 10 cm</th>
<th>Step-up 20 cm</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Weight-shift phase (ms)</td>
<td>253.3 ± 75.3</td>
<td>256.5 ± 76.4</td>
<td>308.8 ± 53.2</td>
<td>266.2 ± 75.5</td>
<td>–</td>
</tr>
<tr>
<td>Swing phase (ms)</td>
<td>248.2 ± 37.1</td>
<td>313.0 ± 49.5</td>
<td>369.9 ± 51.2</td>
<td>378.2 ± 51.1</td>
<td>–</td>
</tr>
<tr>
<td>CoP-step (%)</td>
<td>5.4 ± 3.9</td>
<td>4.8 ± 3.3</td>
<td>15.4 ± 3.1</td>
<td>12.1 ± 5.7</td>
<td>**</td>
</tr>
<tr>
<td>CoP-support (%)</td>
<td>25.1 ± 2.4</td>
<td>24.5 ± 1.5</td>
<td>23.5 ± 2.6</td>
<td>24.5 ± 2.0</td>
<td>**</td>
</tr>
<tr>
<td>CoG-support (%)</td>
<td>0.6 ± 1.6</td>
<td>-0.1 ± 1.4</td>
<td>3.3 ± 1.3</td>
<td>2.5 ± 1.8</td>
<td>**</td>
</tr>
<tr>
<td>Acromion (%)</td>
<td>0.5 ± 0.4</td>
<td>0.6 ± 0.4</td>
<td>1.1 ± 0.3</td>
<td>1.1 ± 0.4</td>
<td>**</td>
</tr>
<tr>
<td>Greater trochanter (%)</td>
<td>-0.1 ± 0.3</td>
<td>-0.3 ± 0.3</td>
<td>0.5 ± 0.3</td>
<td>0.3 ± 0.5</td>
<td>**</td>
</tr>
<tr>
<td>Shoulder angle (degrees)</td>
<td>0.5 ± 0.4</td>
<td>0.9 ± 0.7</td>
<td>0.7 ± 0.6</td>
<td>1.0 ± 0.7</td>
<td>**</td>
</tr>
<tr>
<td>Pelvic angle (degrees)</td>
<td>1.9 ± 0.8</td>
<td>2.1 ± 0.9</td>
<td>1.8 ± 0.8</td>
<td>2.1 ± 0.8</td>
<td>–</td>
</tr>
<tr>
<td>ES-support (%)</td>
<td>4.6 ± 3.4</td>
<td>5.4 ± 3.0</td>
<td>6.0 ± 3.4</td>
<td>5.5 ± 3.1</td>
<td>–</td>
</tr>
<tr>
<td>GMD-support (%)</td>
<td>49.7 ± 9.3</td>
<td>55.9 ± 11.2</td>
<td>41.6 ± 8.4</td>
<td>44.6 ± 10.3</td>
<td>**</td>
</tr>
<tr>
<td>ADD-support (%)</td>
<td>6.9 ± 8.4</td>
<td>7.8 ± 8.5</td>
<td>6.8 ± 7.4</td>
<td>7.9 ± 10.4</td>
<td>–</td>
</tr>
<tr>
<td>ES-step (%)</td>
<td>13.4 ± 8.6</td>
<td>13.2 ± 7.9</td>
<td>16.1 ± 8.8</td>
<td>14.9 ± 8.1</td>
<td>–</td>
</tr>
<tr>
<td>GMD-step (%)</td>
<td>12.7 ± 13.4</td>
<td>13.4 ± 12.8</td>
<td>5.9 ± 3.8</td>
<td>7.5 ± 4.5</td>
<td>– **</td>
</tr>
<tr>
<td>ADD-step (%)</td>
<td>6.1 ± 10.2</td>
<td>3.7 ± 3.5</td>
<td>26.1 ± 33.9</td>
<td>20.7 ± 29.6</td>
<td>**</td>
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</tbody>
</table>

a: % base of support, b: % height, c: % maximal voluntary contraction, *: p<0.05, **: p<0.01.

**Fig. 3.** Relationship between center of pressure shift toward stepping side and gluteus medius activity of the supporting leg.
very different from the forward motions. According to prior studies\(^6, 7\), the CoP shift represents an accurate index of CoG acceleration. In the step-up task in this study, CoP shifted more significantly toward the stepping side to translate CoG to the supporting side during the weight-shift phase, and the swing phase was longer than in the step task. The longer swing phase provided greater opportunity for CoG to shift laterally toward the stepping side during the single-limb support. Therefore, the CoP shift should propel CoG toward the supporting leg and effectively decrease the extent of CoG movement toward the stepping leg during the step-up task.

According to reports on ML postural control, the control of balance occurs primarily at the hips rather than at the ankles\(^1, 2\), and the hip abductors of the supporting leg as well as the hip adductors of the stepping leg contributes to the checking of the CoG shift to the supporting side\(^9\). In addition, it is presumed that a correlation exists between the CoP shift toward the swing side and the magnitude of GM activity of the supporting leg\(^13\). In the present study, a negative correlation was found between CoP-step and the magnitude of GM activity of the supporting leg\(^13\). In the present study, a negative correlation was found between CoP-step and the magnitude of GM activity of the supporting leg, and a weak correlation was found between CoP-step and the magnitude of AL activity of the stepping leg. In the rapid step and step-up motions in the lateral direction examined in this study, the shift of CoG toward the supporting side might be disadvantageous, because it moved diametrically opposite in the stepping direction. Therefore, it might be that the GM activity of the supporting leg checked the shift of CoG to the supporting side, and produced the momentum of CoG to the stepping side. This phenomenon was more notable when the step length was extended. A previous study demonstrated that the phase of anticipatory postural adjustments is absent or shortened when subjects react rapidly to the imposed instability in a perturbation task\(^19\). The phenomena observed in the step tasks in our present study should also be considered as responses to the required rapid motions. On the other hand, in the step-up task, it seemed that CoG shifted toward the supporting side to allow the stepping leg to rise onto the stool. Consequently, the GM activity of the supporting leg decreased in the step-up tasks. Moreover, we consider that when a larger shift of CoG to the supporting side was required, the AL activity of the stepping leg was recruited to control CoG. It was reported that the limb extensor moment (hip extension, knee extension, and ankle plantar flexion, the so-called “support moment”) plays a dominant role in controlling CoG in the vertical direction during gait ascent and descent\(^21, 22\). Although EMGs of the gluteus maximus, quadriceps and triceps surae were not examined in this study, it might be that activities in these muscles are higher in the step-up task than in the step task.

Regarding the kinematical results in the weight-shift phase, the characteristic movement of the greater trochanter was confirmed. The greater trochanter moved toward the supporting side in the step-up task, whereas it tended to move toward the stepping side in the step task. These movements of the greater trochanter were similar to those of CoG. On the other hand, the acromion moved toward the supporting side in both the step and step-up tasks. From the analysis of these kinematical variables, it might be that the movement of the pelvis is prioritized in strategies for rapid lateral motion. A study of lower-trunk coordination and energy transfer during walking showed that young adults use a gait in which the pelvis leads the trunk (pelvis-leading strategy), and that elderly more likely use a gait in which the trunk leads the pelvis (trunk-leading strategy); the pelvis-leading strategy decreases the mechanical energy expenditure required\(^20\). Our results suggest that young adults perform the task efficiently in terms of mechanical energy expenditure. Although the shoulder angle increased more when the step and step-up tasks were performed for the long lengths, it would be appropriate to interpret this observation as a reaction accompanying the placement of the lower limb more distantly, rather than the disruption of postural balance. The finding that the shoulder angle was not significantly different between the step and step-up tasks, in spite of the motion of the CoG changing dynamically, might support our interpretation.

In conclusion, the kinesiological analysis of the normal subjects indicates that, in contrast to the results of forward step and step-up motions, CoG shifted further toward the supporting side in the step-up task than in the step task, and that the GM activity of the supporting leg was larger in the step task than in the step-up task. Also, the AL activity of the stepping leg was larger in the step-up task than in the step task. These findings suggest that the
GM activity of the supporting leg and the AL activity of the stepping leg might control the shift of CoG in the frontal plane. Our findings indicate that the increase in GM activity does not depend on the height of stepping but on the length of stepping during lateral step and step-up motions. This information provides a baseline for the kinesiological norms of lateral step and step-up motion tasks performed by healthy young adults. The application of this protocol to the elderly population and patients with hip disorders or balance problems may enhance our understanding of the mechanisms underlying dysfunctions in postural control.

REFERENCES