Comparison of the Ankle, Knee, Hip, and Trunk Corrective Action Shown During Single-Leg Stance on Firm, Foam, and Multiaxial Surfaces

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ABSTRACT. Riemann BL, Myers JB, Lephart SM. Comparison of the ankle, knee, hip, and trunk corrective action show during single-leg stance on firm, foam, and multiaxial surfaces. Arch Phys Med Rehabil 2003;84:90-5.

Objective: To compare the quantity of ankle, knee, hip, and trunk corrective actions shown during single-leg stance.

Design: Counter-balanced crossover design. Single-leg stance under the conditions of eyes open on firm, foam, and multiaxial surfaces and eyes closed on a firm surface were recorded for 12 seconds in 18 participants.

Setting: A university neuromuscular research laboratory.

Participants: Eighteen healthy and recreationally active college students.

Interventions: Not applicable.

Main Outcome Measure: Average angular displacement magnitude between successive sampling instances for the ankle, knee, hip, and trunk.

Results: A significant condition by joint interaction was revealed. Post hoc comparisons revealed that the ankle dominated as the source of corrective action across each of the testing conditions. As the challenge became greater because of foam surface or eyes closed, more corrective action occurred at proximal joints (hip and/or knee).

Conclusions: The ankle is of primary importance during single-leg stance on firm, foam, and multiaxial surfaces, with proximal joints having an increased role under more challenging conditions. These results provide a scientific basis for clinicians' and researchers' decisions about support surface and visual condition during single-leg postural control testing and training.

Key Words: Biomechanics; Equilibrium; Kinematics; Posture; Rehabilitation.

UNDERLYING ALL MOTOR activities of the body, involving both feed-forward and feed-back mechanisms, are specific strategies taken to ensure the maintenance of postural control. Maintaining postural control is a continuous process. It requires the sensory detection of body and segmental motion and position and the integration and processing of this information within the central nervous system into efferent (motor) commands. Postural control strategies are those sensorimotor solutions used to maintain control over posture; they include muscle synergies, movement patterns, joint torques, and contact forces.[1] The acquisition of effective and efficient postural control strategies is essential. It affects both optimal functional performance and athletic performance. Aside from the neurologic commands, the final outcome of a postural control strategy depends on many environmental, physiologic, and biomechanic factors.[2,3]

During quiet stance on a fixed support surface, the postural control system must contend with gravity as the largest destabilizing force. In this situation, small corrective movements characterize postural control.[4] It is important to recognize that a redundancy exists relative to the different kinematic combinations that can be used to maintain or restore equilibrium.[5] In other words, any major joint in the kinematic chain may be used to produce the needed corrective action. Nashner and McCollum[6] proposed that, despite the redundancy and indefinite number of muscle activation and strategy possibilities, the postural control system uses only a limited set of distinct contractile patterns. This approach reduces the number of motor actions to more manageable numbers. It has also been hypothesized that the postural control system chooses patterns that require a minimal number of muscles.[5,7] Stemming from research on bilateral stance perturbation, several authors[8-10] have proposed and advocated 3 strategies (ankle, hip, stepping) for controlling equilibrium in the sagittal plane. The terms *ankle* and *hip* refer to the joints primarily responsible for corrective action. Nashner and Woollacott[8] also described a suspensatory strategy that involved knee flexion as a means to lower the body's center of mass (COM). Of note, these researchers used bilateral postural perturbation testing conditions to develop their hypotheses. Since the original works, several authors[11-14] have suggested that these strategies coexist during quiet stance.

Although the vast majority of postural stability research has been on double-leg stance,[11,15-22] periods of single-leg stance occur frequently within many activities of daily living, such as putting on a pair of pants. Single-leg postural assessments enable researchers to make bilateral comparisons, an often important application in orthopedic settings.[23] Postural instability increases during single-leg stance,[23-25] most likely as a result of the required reorganization of the center of gravity (COG) over a short and narrow base of support. The increased challenge of maintaining single-leg equilibrium may better elicit postural control differences existing between various populations, such as those associated with different age groups.[24,26] Although sagittal plane control dominates double-leg stance,[27] the frontal plane is hypothetically more important during single-leg stance control.[13] Despite widespread single-leg testing and training, little is known about the motor strat-

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egies humans use to maintain equilibrium during single-leg stance.

Hoogvliet et al[13] described 2 frontal plane strategies for fixed surface, single-leg stance. The first, the foot-tilt strategy, refers to the tilting movements of the foot resulting from movements of the subtalar joint. The second strategy, the hip strategy, coincides with the previously described hip strategy,[10] with the exception that it occurs in the frontal plane. Similar observations of hip strategy during single-leg stance, under identical support surface conditions, were made by Tropp and Odenrick[12] in their comparison of normal subjects and patients with functionally unstable ankles. Whether the suspensatory strategy[8] of increasing knee flexion, and possibly trunk flexion, during moments when single-leg stance equilibrium becomes challenged was not investigated.

No published literature to date describes single-leg stance on unstable surfaces. Despite their widespread use, especially in orthopedic rehabilitation and postural control testing, no data could be found describing the kinematic patterns exhibited during single-leg stance on foam and multiaxial surfaces. Therefore, the purpose of the present investigation was to compare the quantity of ankle, knee, hip, and trunk corrective actions shown during single-leg stance on firm, foam, and multiaxial surfaces.

METHODS

Participants

Eighteen healthy and recreationally active subjects (9 men, 9 women; age range, 18–25y; mean weight ± standard deviation [SD], 70.9±9.4kg; mean height, 171.1±10.0cm) participated. *Recreationally active* was operationally defined as individuals who participated in some form of physical activity, 20 minutes in duration, at least 3 times weekly. Subjects were excluded if they presented with a history of balance deficits, lower-extremity neurologic disorders, or lower-extremity musculoskeletal injury. Because postural deficits exist in individuals who sustain even mild head injury,[28] individuals were excluded if they had previously sustained a head injury. Before participation, subjects were asked to sign a university institutional review board approved informed consent form.

Procedures

Each subject completed a battery of single-leg stance tests under 4 different conditions: firm surface–eyes open (FIEO), firm surface–eyes closed (FIEC), foam surface–eyes open (FOEO), and multiaxial surface–eyes open (MAEO). The order of tests was counterbalanced between subjects, with each trial lasting 12 seconds. Subjects completing each task were instructed to stand as motionless as possible, while maintaining their hands on their iliac crests with their head positioned straight ahead. Additionally, participants were instructed to maintain the contralateral limb (nonstance limb) in 30° of knee and hip flexion. All testing was conducted with the subjects barefoot using the dominant limb, defined as the preferred leg to kick a ball.

During the firm surface condition, subjects stood directly on a firm, fixed surface, whereas in the foam surface condition, subjects stood on a 10-cm thick piece of medium density foam. The Biodex Stability System[a] was used to provide the multiaxial support surface. The device is an unstable multiaxial platform that permits up to 20° of surface deflection in any direction. The relative stability of the platform is adjustable through 8 levels by an electronically controlled stiffness pot, with level 1 being the most stable. During testing under both visual conditions, the multiaxial stability level was maintained at level 6 and foot placement on the platform also was standardized. In the anteroposterior direction, foot placement was made with the navicular tubercle 2.5cm posterior to the central pivot. The posterior aspect of the calcaneus was placed midway over the mediolateral midline, with the second ray pointing 5° lateral to the same line.

Before the 3 trials in which kinematic data were recorded, subjects were given 1 practice trial under each support surface and visual condition. Progressive directions were given to the participants to first take the required stance position, and then place their hands on the iliac crests (followed by eye closure during eyes-closed trials). Data collection for 12 seconds began on a prearranged signal given by the subject. Participants were instructed that on losing their balance, they were to make the necessary adjustments to regain balance (ie, hands off iliac crests, touch down) and return to the standardized testing position as quickly as possible. If a compensatory event occurred, the principal investigator activated an electronic switch for the duration of time the participant remained out of the standardized testing position. Compensatory events were defined according to the Balance Error Scoring System.[23] For each incomplete trial, defined as those in which more than 3 compensatory events occurred, subjects were given 1 retest trial. Thus, the maximum number of trials a subject could attempt at a given condition was 6.

Kinematic Data Collection

Kinematic data were collected by an electromagnetic tracking system with the MotionMonitor™, a commercially available acquisition and analysis software.[b] At the core of the system is a transmitter with 3 orthogonal coils that are used to create an electromagnetic field. Sensors in the magnetic field record the magnetic flux and convey the signals to a base computer through long cables. The MotionMonitor software calculates sensor position and orientation from data conveyed by the sensors. Hardware consisted of a standard range direct-current transmitter and 4 receivers[c] with all settings in the default mode. Sensor data were sampled by the computer at a frequency of 100Hz. The electronic hand switch used to mark compensatory events was also sampled at 100Hz.

During subject setup, sensors were firmly attached to the shank, thigh, sacrum, and seventh cervical vertebra. The ankle, knee, and hip joint centers were calculated with respect to the secured shank and thigh sensors by taking the midpoint between 2 points digitized on contralateral aspects of the joint. Subjects' height and weight were used for the appropriate anthropometric calculations required for locating each segment's COM using the Dempster codes as reported by Winter.[29]

We established a global coordinate system by mounting the transmitter on a custom tripod. During data collection, the transmitter, and therefore the global coordinate system, was aligned with the subject's 3 cardinal planes. The default software global coordinate system was used, which included the positive z axis pointed inferior, the positive x axis pointed anterior, and the positive y axis pointed to the right of the subject. The local coordinates for each segment were established with the positive segmental z axis pointed laterally, the positive segmental y axis pointed superiorly, and the positive segmental x axis pointed anteriorly, matching the International Society of Biomechanics standardization recommendations.[30] A software-driven boresight procedure was used to align sensor axes with the global coordinate system.

The MotionMonitor software was used to calculate Euler angles ($Zy'x''$) between adjacent segments to determine the

Table 1: Relative (ICC) and Absolute (SEM) Reliability for the Dependent Variables

	FIEO		FIEC		FOEO		MAEO	
	ICC	SEM	ICC	SEM	ICC	SEM	ICC	SEM
Ankle	.92	.00687	.90	.01531	.50	.02896	.81	.01243
Knee	.85	.00237	.89	.00738	.76	.00866	.64	.01371
Hip	.91	.00472	.85	.01512	.71	.01824	.87	.01016
Trunk	.91	.00204	.97	.00456	.73	.00700	.82	.00982

Abbreviation: SEM, standard error of measurement.

relative joint angles. Based on the local coordinates and the Euler sequence, flexion-extension was determined first as movement occurring around the z axis, internal-external (left-right rotation for trunk) rotation was determined second as movement occurring around the y axis, abduction-adduction (lateral flexion for trunk) was determined third as movement occurring around the x axis. The specific joint movements we considered were knee flexion, abduction, and rotation; hip flexion, abduction, and rotation; and trunk flexion, lateral flexion, and rotation.

One disadvantage of electromagnetic tracking systems is their sensitivity to metallic objects located within the field created by the transmitter. Because of the metallic construction of the multiaxial surface, we were not able to make direct ankle measurements. Ankle kinematics were determined indirectly based on the motion of the shank segment. By using the boresighted shank sensor axes, the lower-leg orientation with respect to the global coordinate system was determined about the 3 axes by using procedures similar to those described previously. We considered lower-leg orientation with respect to the sagittal plane to represent ankle flexion–plantarflexion, lower-leg orientation with respect to the frontal plane to represent inversion-eversion, and lower-leg orientation with respect to the transverse plane to represent foot abduction-adduction.

All 3-dimensional angular data calculated by the MotionMonitor software, as well as the hand switch data, were exported into text files. Custom software was written to conduct additional postprocessing. This processing included smoothing the angular data (4th-order zero phase lag Butterworth, 10-Hz cutoff) and truncating compensatory event intervals as indicated by the electronic switch signal. Although we recorded 12 seconds of data for each trial, only the middle 10 seconds were used to calculate the variables of interest. First, for each of the joints (ankle, knee, hip, trunk), the vector sum of the 3 separate angular position vectors (flexion-extension, abduction-adduction, rotation) was calculated at every sampling instance. Next, the difference between 2 successive vector sums was calculated and averaged across the entire trial. The average difference at a particular joint across a trial represented the average angular displacement (ie, corrective action) between sampling instances independent of mean joint position. The 4 average angular displacements from the ankle, knee, hip, and trunk were used as the dependent variables entered into the statistical analyses.

The absolute and relative reliability of the these methods were determined by testing 14 individuals who did not participate in the main part of this investigation. Intraclass correlation coefficients ($ICC_{2,1}$) and standard error of measurement of the dependent variable (average angular displacement) for each condition and joint are in table 1.

Data Analysis

The average of the dependent variables across each complete test trial was calculated and entered into statistical analysis. A 2-factor (condition by joint) repeated-measures analysis of variance (ANOVA) was used to statistically analyze the data using SPSS for Windows, version 9.0.[d] The α level was set a priori at P less than .05. Simple main effect post hoc analyses were conducted on the 2-way interaction determined to be significant by the omnibus F test by using Dunn-Bonferroni procedures. Interaction pairwise comparisons were only considered within a joint across each condition and within a condition across each joint.

RESULTS

One subject, who could not complete at least 1 successful trial under the FIEC condition, was removed from the statistical analyses. Means and SDs of the dependent variables are in table 2. Results of the 2-factor ANOVA revealed a significant condition by joint interaction ($F_{9,144}=20.27$, $P<.001$) and significant main effects for condition ($F_{3,48}=49.61$, $P<.001$) and joint ($F_{3,48}=26.57$, $P<.001$). Between joint-within condition comparisons revealed significant differences for FIEC (ankle>hip>knee>trunk), FOEO (ankle>knee and hip>trunk), and MAEO (ankle>knee) conditions. Within joint-between condition comparisons revealed significant differences for the ankle (FIEC>FOEO>MAEO>FIEO), knee (FIEC>FOEO> FIEO and MAEO), hip (FIEC>FOEO>FIEO and MAEO), and trunk (FIEC>FOEO and MAEO>FIEO) joints.

DISCUSSION

Common across all 4 joints, the FIEC condition promoted the greatest quantity of corrective action. Comparison between

Table 2: Means for the Dependent Variables

	FIEO	FIEC	FOEO	MAEO
Ankle	.01652±.0066	.08026±.0339	.05457±.0207	.02335±.0103
Knee	.01296±.0054	.04720±.0213	.03006±.0094	.01661±.0071
Hip	.01449±.0047	.06606±.0324	.03254±.0131	.01861±.0073
Trunk	.01281±.0033	.04021±.0157	.02282±.0071	.01857±.0103

NOTE. Values are mean degrees ± SD.

the eyes open conditions identified the FIEO condition as generally requiring the least amount of corrective action (ankle and trunk), whereas the FOEO condition generally required the most (ankle, knee, hip). This information provides assistance to researchers and clinicians deciding which types of support surface and visual conditions are needed to target specific joints during single-leg stance testing and rehabilitation procedures.

Although the assumption that the body sways as an inverted pendulum about the ankle during quiet double-leg stance has been widely used,[31] multiple studies[21,31-33] have shown that the body does not move as a rigid segment but rather as a multilink structure. For example, Day et al[21] revealed that, in the sagittal plane, the most angular movement occurred between the trunk and upper leg, independent of stance width. Likewise, in the frontal plane, the most angular movement occurred between the trunk and upper leg for stance widths greater than 8cm. The results of the current investigation support a similar concept during single-leg stance on firm, foam, and multiaxial surfaces. Although ankle corrective action dominated, especially during the FIEC and FOEO conditions, common to all test conditions was also varying quantities of knee, hip, and trunk motion. Although it may be expected that having the kinetic chain function in multiple segments rather than 1 segment (ie, inverse pendulum) would decrease stability and increase control complexity, this does not appear to be the case.[34] Having the body function as a multilink structure increases stability through several mechanisms. First, it decreases the large inertia that would be associated with 1 large segment.[34] Furthermore, at each segment, passive dampening can occur, thereby decreasing the need for sustained muscle activation.

During the FIEO condition, an equal amount of corrective action occurred at each of the 4 joints. A similar pattern was revealed for the MAEO condition, with the exception of the ankle joint. Its corrective action was significantly greater than that of the knee. The importance of the ankle joint also extended to the FOEO and FIEC conditions. During these 2 conditions, ankle joint corrective action was significantly greater than any of the other joints. As the task became more challenging (foam surface or removal of vision), an increased reliance on proximal joints was revealed. Specifically, the hip joint became the second greatest source of corrective action during FIEC, whereas both the hip and knee contributed during FOEO. The trunk, defined as the orientation of the seventh cervical vertebra with respect to the sacrum, appeared to be the least important source of corrective action. During the more challenging conditions (FOEO, FIEC), significantly more corrective action occurred between the pelvis and thigh than between the pelvis and trunk. It can be speculated that the higher inertia associated with the trunk may preclude it from contributing to the quick adjustments necessary for single-leg stance equilibrium.

The FIEC condition required the greatest quantity of corrective action across the 4 joints. In contrast, the FIEO condition required the least amount of corrective action, especially with respect to the ankle and trunk joints. No differences existed between the quantity of hip and knee joint corrective action between the FIEO and MAEO conditions. The FOEO condition required the second greatest amount of corrective action from the ankle, knee, and hip joints. From these results, the conditions we studied can be hierarchically ranked according to relative quantity of corrective action required to remain in equilibrium: FIEO$<$MAEO$<$FOEO$<$FIEC. Clinically, this finding provides a rationale for progressing the challenge of single-leg postural control exercise.

Previous studies[18,21,35] of bilateral stances have revealed that vision is a nonessential, but potent influence on postural stabilization. In contrast, the present investigation of single-leg stance revealed that the absence of visual inputs had profound effects on the quantity of corrective action. One commonality between the present investigation and previous bilateral stance studies was that eye closed resulted in increased corrective action originating at the hip joint.[11]

The potent influence of vision shown in the present investigation compared with the previous studies may be attributed to the decreased inherent stability (smaller base of support) associated with single-leg stance. The small margin of sway permitted to remain in equilibrium (vertical projection of COG over base of support) requires accurate and prompt sensory information. Furthermore, single-leg stance reduces the quantity of useful and accurate somatosensory information (both proprioceptive and plantar cutaneous) available to the postural control system. Although both visual and somatosensory inputs have been regarded as being sensitive and important for bilateral quiet stance, the combination of these factors may produce a different situation for single-leg stance. In other words, the decrease in somatosensory information, coupled with decreased inherent stability, may result in a higher reliance on visual inputs. Future research should examine this hypothesis, as well as consider the effects of inaccurate visual and somatosensory inputs on single-leg postural control.

Fixed, firm support surfaces are the most common surfaces used for postural control assessment and training. Unstable surfaces, such as the foam and multiaxial surfaces in the present investigation, are attributed with requiring faster stabilization mechanisms that originate from proprioception.[36,37] Pliant (ie, foam) and freely moving (ie, multiaxial) support surfaces are often used to alter the somatosensory input arising from the plantar cutaneous and ankle joint mechanoreceptors.[9,11,23,38-40] Shumway-Cook and Horak[41] suggested that stance on pliant surfaces caused an increased reliance on a hip strategy. The results of the present investigation support the application of this idea to single-leg stance on a foam surface but not the multiaxial surface.

Although single-leg postural control testing is desirable from the perspective of making bilateral comparisons, the reduction in a person's base of support makes compensatory actions and falls frequent occurrences even in the absence of further experimental manipulation or pathology. Researchers have managed this situation by shortening testing trials,[42] retesting,[4] or grading those trials as incomplete.[26] Unfortunately, these methods may reduce the ability to make an accurate assessment or detect slight alterations. At the heart of the issue over compensatory events is how each type of event (ie, touch down vs eyes opening) influences the measurement variables.[42] The approach chosen in the present investigation was to include trials with compensatory events by truncating the measurement data during the event based on the definitions of the Balance Error Scoring System.[23]

The original design of the current investigation also included trials with eyes closed on the foam and multiaxial surfaces. Unfortunately, the challenge to the postural control system yielded numerous compensatory events and incomplete trials. An analysis of the relative and absolute reliability of the dependent variables under those conditions yielded poor results. As shown by the data in table 1, the relative and absolute reliability for the dependent variables for the conditions we used in the present investigation were within acceptable standards. This suggests that subjects were consistent in the amount of corrective action occurring at a particular joint. Interestingly, the reliability across the dependent variables and conditions in the current investigation are slightly higher than those reported for forceplate measures of single-leg postural steadiness.[42]

Forceplates, which are sensitive to the forces exerted by the body during various tasks, are most often used in postural control assessments. Although they are readily available to many researchers and clinicians, it has been speculated that forceplates are largely influenced by ankle activity[43,44] and may fail to reveal alterations in postural control patterns.[11] Combining forceplate measures with kinematic measures provides an enhanced perspective, possibly more pathognostic, concerning the strategies by which the postural control system maintains equilibrium. In the orthopedic literature, Tropp and Odenrick[12] tested this hypothesis by considering a 2-dimensional analysis of the frontal plane postural control strategies adopted by persons with functional ankle instability during single-leg stance. In their investigation, they revealed that persons with functional ankle instability displayed an increased tendency to use more postural movements at the hip, rather than the ankle, than did the normal subjects. Similarly, Kuo et al[11] provided further evidence supporting the advantage of combining forceplate and kinematic measures for assessing postural stability under varying sensory conditions in normal individuals. The results of the present investigation advance the reports of Tropp and Kuo and provide a basis for future researchers of single-leg stance testing to consider incorporating kinematic measures in addition to the traditional support-surface variables.

A major assumption, and therefore a limitation, in the present investigation resided in the inference of ankle motion based on tibial kinematics. This assumption was necessary because of the interference that the metallic multiaxial surface presented to the electromagnetic tracking system. Basing our understanding on the biomechanical descriptions of ankle joint function by Inman and Mann,[45] we consider this assumption to be valid during stance on fixed support surfaces. Tropp and Odenrick[12] made a similar assumption during a kinematic analysis of single-leg stance on a fixed support surface. A recent direct comparison[46] that used the tibia as a marker for 3-dimensional ankle motion during single-leg stance yielded strong cross correlations between lower-leg orientation in the transverse plane and inversion and eversion (range, −.91 to −.97) and lower-leg orientation in the sagittal plane and plantarflexion and dorsiflexion (range, −.75 to −.76). However, to date, the degree to which this assumption holds true during stance on unstable surfaces is a matter of speculation. We recommend that future research consider the relation between tibial kinematics and ankle motion during single-leg stance on unstable surfaces.

CONCLUSIONS

The results of the present investigation show the importance of the ankle joint for single-leg stabilization on firm, foam, and multiaxial surfaces. The relative contributions of the ankle, knee, hip, and trunk to corrective actions under eyes open and closed conditions and various surfaces showed that proximal joints have a greater role under more challenging conditions. These results provide a scientific basis on which clinicians and researchers can base decisions about support surface and visual conditions for single-leg postural control testing and training.

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Suppliers

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