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Research Article

Changes in Muscle Activity and Body Sway Over Time During One-Leg Stance

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Abstract

It is considered that a one-leg stance (OLS) requires greater lower limb muscle strength and balance than a two-leg stance (TLS), and imposes a burden on the lower limb that progresses with time. This study aimed to compare and examine muscle activity and body sway between the OLS and TLS in young men, focusing on changes over time. A total of 15 subjects stood with their eyes open on a stabilometer for 1 min in either OLS or TLS after the surface electrodes were placed over gastrocnemius, soleus, and tibialis anterior. Maximum and mean % root mean square values and path lengths of X and Y axes, as well as their total path lengths, were calculated over 20-s periods (0–20 s, 20–40 s, 40–60 s) as evaluation parameters for muscle activity and body sway. Muscle activity and body sway were significantly larger in OLS than in TLS, and both total and X axis path lengths in the initial 20-s OLS period were significantly longer than those in the other 2 periods. Lower limb muscles showed little changes in activity over time for either stance but a left–right body sway become small during the OLS. In conclusion, lower limb muscle activity and body sway are decreased in the TLS compared to the OLS. Change in muscle activity over time is small in both stances; however, left–right body sway during the OLS becomes stable within 1 min.

Keywords: One-Leg Stance; Balance Ability; Burden on the Lower Limb; Young Man

Introduction

Generally, humans use both lower limbs simultaneously when standing and during locomotion. Therefore, when unable to use 1 lower limb because of injury or disorder, humans may encounter difficulties with movement. A fracture, for example, could necessitate people to stand on 1 lower limb; this lower limb (supporting lower limb) is required to exert larger muscle strength [1-3] and to maintain more balance [4] than that is required to support their body weight. Therefore, individuals with markedly inferior above-stated abilities find it difficult or impossible to stand on 1 lower limb for a long time.

Unlike standing on 2 lower limbs, standing on 1 lower

limb provides only a single, narrow support base, which complicates stable posture. The one-leg stance (OLS) has been used as a static balance test because it can be performed easily without special equipment. For example, the OLS has been used as part of a physical fitness test for the elderly [5], and as a standardized field sobriety test [6]. In contrast, walking is a basic daily life activity and is necessary for independent life. Because individuals with inferior lower limb muscle strength and knee joint function generally have unstable gait, stability during OLS is used as a predictive index for falling and stable gait. In addition to a static balance test, the OLS has been used in the elderly to evaluate the need for long-term care, musculoskeletal ambulation disorder symptom complex, and the ability to walk safely without falling [7].

Compared with the two-leg stance (TLS), in OLS the subject not only has to exert large lower limb muscle strength but also should have the ability to maintain balance. Therefore it has been used various above-mentioned situations. The TLS can extend the supporting base and subsequently increase stability by adjusting the distance between the feet, which is not possible in the OLS. TLS also can reduce the burden imposed on 1 lower limb by shifting the center of gravity (COG) between the lower limbs, which is again not possible in the OLS. In short, people must continuously exert greater force with 1 lower limb during the OLS to support body weight because of instability induced by narrow base of support.

Hence, even for young people with developed lower limb muscle strength, it is difficult to maintain the OLS for a long time as OLS over a long period of time leads to disturbance in balance because of increasing lower limb muscle fatigue. Therefore, we hypothesized that the OLS produces greater postural sway and lower limb muscle electromyographic (EMG) activity than the TLS, and these effects increase with time. This study aimed to compare the mean and maximum EMG levels recorded from lower limb muscles, and postural sway between the OLS and TLS, as well as the changes in these variables over time.

Method

1. Subjects

We explain the aim and procedures of this study to 17 young male students belonging to a university in Japan, and 15 persons among them (mean age: 20.2 ± 1.3 years, mean height: 172.1 ± 7.3 cm, mean weight: 65.4 ± 3.4 kg) who agreed and signed informed consent form participated in this study. It was confirmed that all subjects do not have orthopedic disease and/or nervous system disease. The subjects participated in sports such as baseball, track and field, or swimming 2 or 3 times per week. All subjects were judged as right-leg dominant on the basis of a dominant lower limb survey [8]. This study was approved by the Ethics Committee on Human Experimentation of Faculty of Human Science, Kanazawa University (2012-04).

2. Experimental procedure

Subjects stood in either the OLS or TLS on the stabilometer with eyes open after pasting electrode on gastrocnemius, soleus and tibialis anterior muscles of the right leg. In the OLS, subjects stood on the right lower limb (supporting lower limb) with their arms relaxed at their sides, and while flexing the left knee approximately 90 degrees. In the TLS, subjects stood with their arms relaxed at their sides. After confirming that the body's way has reached a stable state, subjects stood in either the OLS or TLS for a single 60-s trial. Subjects performed 3 trials for both stances with 1-min rest periods interleaved be-

tween trials. After subjects performed each stance, we recorded EMG activity at maximum voluntarily isometric contraction (MVIC) during plantarflexion and dorsal extension. MVICs for each movement were performed in three 5-s trials with 1-min rest intervals between trials.

3. Surface EMG

Surface EMG was measured in all tests using a multi-channel telemetry system (NIHON KOHDEN, Tokyo, Japan). The codeless active electrodes (ZB-150H, Nihon Kohden, Tokyo, Japan) were attached target muscles, and obtained EMG data were sent the host computer. Active electrodes can measure noiseless EMG data, because they prevent mixing artifact. ICC of EMG data in this study are 0.61~0.98. Landis et al. (1977) reported that sufficient reliability is insured if ICC is over 0.61. Hence, reliability about EMG data was judged to be sufficient in this study. EMG signals were band-pass filtered (20–500 Hz) and sampled at 1000 Hz. Various lower limb muscles contribute to maintain the standing position. Especially in young people, ankle strategy is mainly used to maintain stable standing condition [9], and during that time leg muscle groups related to plantar flexion and dorsal extension work together positively. Therefore, we recorded EMG activity in the right leg from gastrocnemius, soleus, and tibialis anterior. The raw EMG signal was translated by calculating the root mean square (RMS) of each 1-s unit every 0.1s.

4. Body sway

Body sway was measured by a stabilometer (gravicorder GP-5000, Anima). The stabilometer contains 3 vertical load cells that measured the center of pressure (COP) for vertical loads, which was streamed to a personal computer using an analog/digital converter. We recorded the COP path in all tests with a sampling rate of 20 Hz.

5. Evaluation variables

In this study, maximum %RMS and mean %RMS were used as variables to evaluate muscle activity. Maximum and mean %RMS were calculated every 20 s during the OLS and TLS on the basis of maximum and mean RMSs collected every 20 s during the OLS and TLS, and maximum RMS during MVIC. Maximum RMS values during MVIC were calculated for gastrocnemius and soleus when the subject performed plantarflexion and tibialis anterior when they performed dorsal extension. The mean of these values over 3 trials were used in the analysis. In this study, the path length of X and Y axes and their total path length were used to quantify variables affecting body sway [10]. The path lengths are typical measures of body sway, whereas total path length is the total component for X and Y axes path length. These variables were calculated every 20 s for OLS and TLS; a mean of 3 trials was used for the

analysis.

6. Statistical analysis

A two-way repeated measures analysis of variance (ANOVA) was used to examine differences among means. EMG activity and body sway were the dependent variables, while the stances (OLS and TLS) and time periods (0–20 s, 20–40 s, and 40–60 s) were the independent variables. Tukey’s HSD test was used for multiple comparisons if a significant interaction or main effect was found. The significance level was set to $p < 0.05$.

Results

Tables 1 and 2 list the basic statistics and statistical comparisons of maximum and mean %RMS for OLS and TLS along with results from two-way ANOVA. In both parameters, the interaction and main effect of the time period factor were statistically insignificant, whereas the main effect of the standing posture factor was statistically significant. Means of both parameters were larger in OLS than in TLS during all time periods.

Table 1. Comparisons of maximum %RMS for OLS and TLS in each muscle. (n=15)

Maximum %RMS (%)		Time unit						F-value	Post-hoc Tukey's HSD	
		0-20s		20-40s		40-60s				
		Mean	SD	Mean	SD	Mean	SD			
Tibialis anterior muscle	TLS	2.87	1.07	3.13	1.27	3.50	1.47	F1	40.198*	OLS > TLS
	OLS	20.55	8.06	22.06	9.53	24.24	12.48	F2	1.163	
								F3	0.591	
Gastrocnemius muscle	TLS	9.94	2.60	11.59	6.09	10.74	4.47	F1	30.964*	OLS > TLS
	OLS	35.79	12.46	36.40	12.09	39.91	16.92	F2	2.620	
								F3	1.689	
Soleus muscle	TLS	21.32	9.88	20.08	9.46	20.77	11.46	F1	17.695*	OLS > TLS
	OLS	38.36	6.80	36.10	8.12	39.99	10.90	F2	1.271	
								F3	0.589	

Note. *: $p < .05$. F1: standing posture. F2: time unit. F3: interaction. TLS: two-leg stance. OLS: one-leg stance

Table 2. Comparisons of mean %RMS for OLS and TLS in each muscle. (n=15)

Mean %RMS (%)		Time unit						F-value	Post-hoc Tukey's HSD	
		0-20s		20-40s		40-60s				
		Mean	SD	Mean	SD	Mean	SD			
Tibialis anterior muscle	TLS	2.58	0.90	2.61	0.91	2.72	0.98	F1	21.757*	OLS > TLS
	OLS	10.21	4.40	9.48	4.80	9.80	5.29	F2	0.375	
								F3	0.515	
Gastrocnemius muscle	TLS	7.14	0.86	7.20	1.28	7.46	1.26	F1	31.312*	OLS > TLS
	OLS	24.37	8.73	24.89	8.55	25.19	9.60	F2	2.636	
								F3	1.702	
Soleus muscle	TLS	16.84	7.18	15.66	5.75	16.17	7.61	F1	17.639*	OLS > TLS
	OLS	27.68	7.52	26.85	8.15	28.31	8.87	F2	2.470	
								F3	1.417	

Note. *: $p < .05$. F1: standing posture. F2: time unit. F3: interaction. TLS: two-leg stance. OLS: one-leg stance.

Figures 1–3 present the results of the two-way ANOVA. Significant interactions were found in the total and X axis path lengths. In the post hoc analysis, the means were longer in the OLS than in the TLS in all time periods. In the OLS, the mean path lengths were longer in the 0–20-s period than in the 2 other periods (20–40 s and 40–60 s). The Y axis path length was significantly longer in the OLS than in the TLS during all time periods.

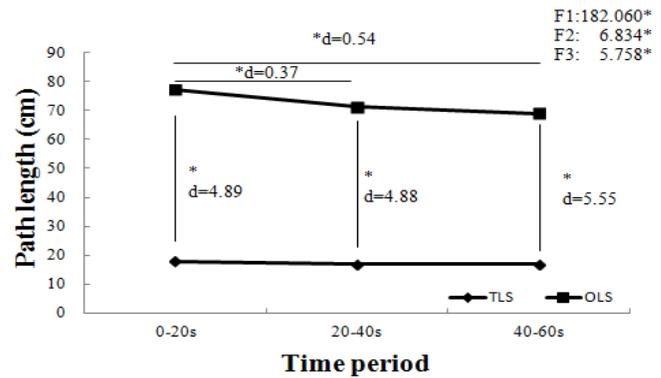


Figure 1. Mean changes among each 20 second in TLS and OLS for total path length and results of two way ANOVA (n=15). *: $p < 0.05$. d = Cohen’s d. F1: Standing posture. F2: Time period. F3: Interaction. TLS: Two legs stance. OLS: One leg stance.

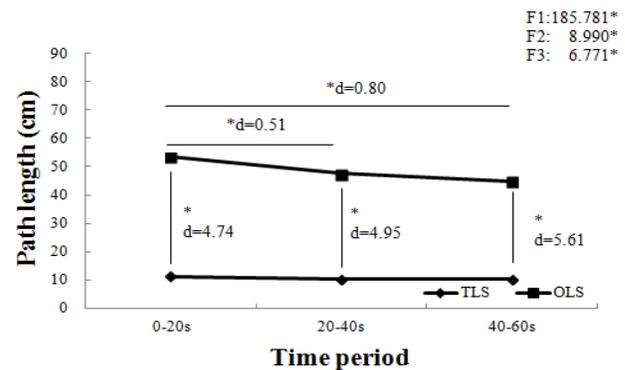


Figure 2. Mean changes among each 20 second in TLS and OLS for X axis path length and results of two way ANOVA (n=15). *: $p < 0.05$. d = Cohen’s d. F1: Standing posture. F2: Time period. F3: Interaction. TLS: Two legs stance. OLS: One leg stance.

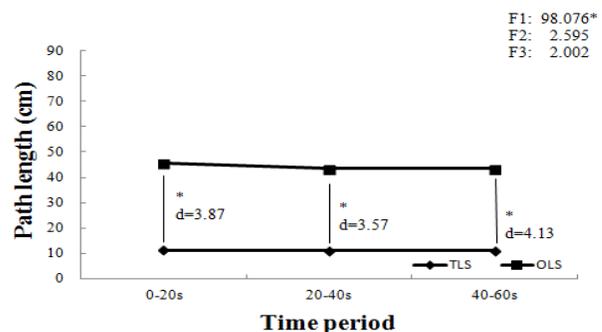


Figure 3. Mean changes among each 20 second in TLS and OLS for Y axis path length and results of two way ANOVA (n=15). *: $p < 0.05$. d =

Cohen's d. F1: Standing posture. F2: Time period. F3: Interaction. TLS: Two legs stance. OLS: One leg stance.

Discussion

During OLS, body weight is supported with only 1 lower limb and COG is adjusted only within a narrow support base. Therefore, in OLS, exertion of greater muscle strength and ability to balance is required. From the present study, it can be inferred that greater lower limb muscle strength and balance were required in the OLS than in the TLS because body sway and muscle activity were greater in OLS.

People can reduce the burden imposed on 1 lower limb by shifting the COG between lower limbs, which cannot be performed in the OLS. We hypothesized that standing in the OLS imposes a greater burden on the supporting lower limb and that postural sway and muscle activity increases with time. However, we observed that body sway did not increase, rather it decreased with time. Therefore, this hypothesis was rejected. In daily life, the lower limb muscles are continuously used to maintain posture and balance, because basically humans are usually active in standing posture. Leg muscles contain a larger proportion of slow-twitch muscle fibers [11], which are superior in muscle endurance compared with other fibers. In addition, the burden imposed on leg muscles during the OLS was small considering the mean EMG levels were approximately 10% of the MVIC in tibialis anterior and approximately 20% in gastrocnemius and soleus. Kahn and Monod (1989) reported that when MVIC is below 15%, muscle fatigue is rare, which allows muscles to work for a very long time [12]. The present results failed to show an increase in body sway with time. Considering subjects were young with developed lower limb muscle strength, lower limb fatigue hardly occurred during OLS over 1-min periods. Furthermore, body sway did not increase in the latter phase. In future studies, it will be necessary to further examine these variables in elderly subjects with inferior lower limb muscle strength.

Body sway can be unstable for a few seconds after lifting 1 lower limb in the OLS [13]. In a body sway test of static standing posture, measurement generally begins when body sway has stabilized [10]. In the present study, to determine a unified measurement condition, both tests (OLS or TLS) were started after confirming body sway was stable. Therefore, it can be inferred that an initial body sway had little effect on the present results. Body sway in the left–right direction and the total in the OLS decreased with time, but sway in the front–back direction did not. The stable range of COG in the front–back direction was within 30%–60% of foot length from the heel to toe, but when the COG drifts beyond this range towards the edge of the supporting base limits, body sway increases [14]. In addition, in the left–right direction, body sway increases at the edge of the supporting base [15]. It is considered that be-

cause the range of the supporting base is very narrow in the left–right direction as compared with front–back direction, instability in the left–right direction is very high during OLS. It is inferred that an initial phase (first 20s) in the OLS contains greater body sway in the left–right direction to adjust the COG, but after the fine adjustment, body sway is stabilized. It is considered that a change of total path length is mainly attributed to a decrease in body sway in the left–right direction. However, these observations were in young subjects with developed lower limb muscle strength. Subjects with inferior lower limb muscle strength, specifically the elderly, may yield different results.

Large body sway means active shifts in the COG, and it is considered that muscle activity level increases to adjust for such changes [16]. We hypothesized that muscle activity level increases with increasing body sway in the latter phases of OLS. However, body sway decreased with no increase in muscle activity. In addition, it was hypothesized that muscle activity would increase in the initial phase of the OLS because of body sway in the left–right direction. However, EMG activity was not increased in the target muscles i.e., gastrocnemius, soleus, and tibialis anterior during the initial OLS phase. Therefore, this hypothesis was rejected.

Adjustment in the COG for the left–right direction during standing is mediated by hip and trunk muscles, whereas ankle muscles provide little contribution to such adjustments. Gastrocnemius, soleus, and tibialis anterior contribute to plantarflexion and dorsal extension, respectively. Therefore, these muscles are less likely to be active during the initial body sway in the left–right direction during the OLS [17].

Change in body sway and lower limb muscle activity with time during one-leg and two-leg stances has little been studied. We obtained findings about the above change with time by comparing them during one-leg and two-leg stances. The findings in this study will be available when the elderly with inferior leg strength used one-leg stance to evaluate balance ability and/or to enhance leg strength.

This study has some limitations: a large problem was that we used young males as subjects and used 1 min as measurement time of one-leg stance. Because young males with superior leg strength and balance ability can perform one leg stance easily, they may have been a little also burden imposing on the supporting leg for one minute. Hence, muscle fatigue of the supporting leg may have little occurred found during one-leg stance. Different results may be found when used the elderly with inferior leg strength and balance ability as subjects. Further studies will be needed to examine the above problem using the elderly as subjects and using one-leg stance with longer time for the young.

In conclusion, lower limb muscle activity and body sway are lower in the TLS than in the OLS. Change in muscle activity over time is minimal in both postures, but body sway in the left-right direction in the OLS is stable within 1 min.

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