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Muscle Activity and Postural Control during Standing of Healthy Adults Wearing a Simulated Trans-Femoral Prosthesis

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Abstract. [Purpose] It is unknown what compensatory strategies are employed by amputees during standing at the initial stage of prosthesis wearing. This study examined the muscle activity during standing with different positions of center of gravity (COG) of healthy subjects wearing a simulated trans-femoral prosthesis on the right leg for the first time. [Subjects] Eight healthy subjects participated in this study. [Methods] Electromyograms of the left lower limb muscles were recorded during standing with different COG conditions. Center of pressure was also measured using a force platform. [Results] There were significant differences among the standing conditions for all the muscles. The tibialis anterior and gastrocnemius were mainly activated in weight shift backward and forward, respectively, which suggests that stiffness of ankle muscles, an ankle strategy, is related to maintaining balance. Significant activity of the semitendinosus, vastus medialis and lateralis also occurred, which indicates that an additional hip strategy was also employed for the sound leg during standing. In addition, there were significant differences of total sway path between shift backward and the other standing conditions. [Conclusion] Fear of falling and lack of sensory feedback may affect standing balance control, particularly in shift backward. Subjects will employ/develop a hip strategy as a compensatory mechanism due to peripheral dysfunction and fear of falling at the first experience of prosthetic standing.
Key words: Compensatory Strategies, Amputees, Simulated trans-femoral prosthesis

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INTRODUCTION

Amputation is a significant health problem and impairs balance ability. Lack of balance control is possibly related to the incidence of falls among amputees. According to Miller et al.[1]) approximately 50% of people with lower limb amputation experienced falls, and the same percentage also reported fear of falling. Fear of falling may lead to avoidance of daily activities causing a decrease in mobility and independent living ability. Optimal balance control is therefore a major concern for amputees. To prevent falling and deterioration of balance ability, physical therapy interventions should aim to provide specific exercises based on the appropriate assessment of balance.

Trans-tibial or trans-femoral lower limb amputation causes significant balance problems because of biomechanical changes such as muscle dysfunction and loss of joints. Postural stability as a key function of balance control is usually evaluated by postural sway. Several studies have revealed that the postural sway of lower limb amputees is increased relative to that of healthy subjects[2–4]. These results suggest such patients have reduced balance ability in static standing positions. Upright standing is a static posture, however, and it takes less muscular effort to maintain postural stability because the ligaments support and maintain the integrity of the joints. Electromyographic studies during normal standing found none of the intrinsic foot musculature was active, and the posterior calf muscles such as the gastrocnemius and soleus were more active than the anterior muscles[5]. Many of the balance problems experienced by amputees result from inadequate muscle control over maintenance of dynamic control of postural stability. The evaluation of static balance is insufficient for assessing the ability of dynamic movements such as walking. Therefore, the assessment of balance control for amputees should include muscle activity in various positions of the center of gravity.

Optimal postural control ability is an essential component of smooth and energy-efficient walking[6]. Walking is a continuous control movement, which includes a transition from bipedal to monopodal stance. Many studies have described the control of balance during standing and walking. Using kinematic and electromyographic data, previous studies have provided descriptions of the coordination between posture and movement during biomechanical tasks, such as lateral leg raising or single leg flexion[7,8]. In addition, recent studies have revealed the existence of compensatory strategies for maintaining standing balance employed by patients with lower leg impairment[9,10]. Isakov[10] showed that compensatory sound leg muscles became dominant in maintaining standing equilibrium because loss of the ankle joint and relevant muscles interfered with the normal distribution of weight bearing in standing.

Table 1. Characteristics of Subjects

Age (year)	Height (cm)	Weight (kg)
29.5 ± 6.5	172.0 ± 7.1	63.38 ± 8.92

N=8 (Mean ± SD)

Those studies have investigated various perspectives of balance including postural control in dynamic movement by amputees. However, most subjects included in those previous studies had already used the prosthesis for a period of time prior to their participation, and had previously experienced standing or walking with the prosthesis. Little information is available to reveal how lower limb amputees improve balance control or achieve motor control when they wear prostheses for the first time. It is well known that the best balance training results for people with lower limb amputations can be achieved in the short time period after the amputation and first prosthesis fitting[11]. Therefore, it is important to examine postural control and to know what compensatory strategies are employed at the first time of prosthesis wearing. In the present study, we examined lower limb muscle activity and postural sway during standing with different positions of center of gravity of healthy subjects wearing a simulated trans-femoral prosthesis for the first time. We aimed to investigate postural control during the first experience of prosthetic standing by requesting these subjects to perform standing tasks as if they were recent right trans-femoral amputees.

SUBJECTS AND METHODS

Eight healthy subjects (six males and two females) volunteered to participate in the study (Table 1). The subjects had no history of neurological, vestibular or musculoskeletal impairments. All subjects read and signed an informed consent revealing all the details of the study protocol, which had been approved by the ethics committee of Tohoku University Graduate School of Medicine.

A simulated trans-femoral prosthesis was produced by Sasaki Prosthetic and Orthotic Service, Inc, Sendai, Japan. The prosthesis had a quadrilateral socket, single-axis knee joint (3R15 OttoBock) and SACH foot (J-foot: M1170 Imasen

Engineering Corp.Japan). The length of the prosthesis was adjusted by the pylon (30 mm) for each subject.

The electromyographic (EMG) activity of the left leg was recorded as the simulated sound side. Disposable self-adhesive electrodes (Blue sensor) were used to record the surface EMG activity of five leg muscles: the vastus lateralis (VL), vastus medialis (VM), semitendinosus (ST), tibialis anterior (TA) and gastrocnemius (GA). The EMG signals were amplified (to 1.0 mV) and filtered (band pass 20–1,000 Hz) using a NEC: Synact MT-11 (NEC Corp., Tokyo, Japan). The signals were digitized using Power Lab 16s (AD Instruments Ltd, Hasting, UK) at a sampling rate of 1,000 Hz and stored on a personal computer, Mac OS J1-8.6, (Apple Computers, Cupertino, CA, USA) for off-line analysis. The digitized data were analyzed using Chart 36.8 software. The EMG recorder was synchronized with a force platform, (Anima System Gravicorder: G-5500, Anima Corp., Tokyo, Japan) that recorded ground reaction forces. Three force transducers on each corner of the plate recorded the vertical ground reaction forces. Force plate data were sampled at 100 Hz, and passed through a digital low-pass 20 Hz filter to eliminate erroneous reads due to noise.

The subjects were asked to stand in an upright position (UP) with both feet on a force plate for 10 seconds. They then wore the simulated prosthesis on the right leg and were asked to stand under three different conditions on the force platform : prosthetic standing (PS), weight shift to forward (SF), and backward (SB). EMG activity was recorded for ten seconds during each standing condition. To obtain the standard values of EMG activity, EMG and muscle strength were recorded during maximal isometric contractions in the standard positions of the manual muscle test as described by Kendall et al.[12] for all of the target muscles for five seconds before the start of each trial. When EMG activity during standing was being recorded, the vertical ground reaction forces at the three transducers were also recorded using the force plate to calculate the movement of the center of pressure (COP). Recording was continuously performed from start to end of a trial.

All EMG data were smoothed by the root- mean-square (RMS) method. RMS in each standing condition was normalized by the maximum isometric contraction for five seconds as 100% (%RMSEMG). The total length of sway path (TSP) was produced by measuring COP movement for ten seconds.

The acquired data were analyzed using SPSS statistical software. Repeated measures analysis of variance (ANOVA) was used to test for significant differences between the four conditions (UP, PS, SF and SB) in each muscle. Bonferroni's multiple comparisons test was then used to compare the muscle activities between pairs of conditions. In addition, ANOVA was used to investigate sway path. The level of statistical significance was determined as $p<0.05$.

RESULTS

ANOVA showed a significant difference of %RMSEMG among the four different standing conditions for each muscle examined (TA: $F = 13.734$, $p<0.01$, GA: $F = 21.002$, $p<0.01$, VL: $F = 4.796$, $p<0.01$, VM: $F = 12.256$, $p<0.01$, ST: $F = 8.828$, $p<0.01$) (Table 2). Bonferroni's multiple comparisons test revealed significant differences between some conditions for each muscle. In TA, there were significant differences found as follows: SB > UP ($p<0.01$), SB > PS ($p<0.01$), SB > SF ($p<0.05$). In GA, there were significant differences found as follows: SF > UP ($p<0.01$), SF > PS ($p<0.01$), SF > SB ($p<0.01$). In VL, there was a significant difference between SB and UP (SB > UP, $p<0.01$). In VM, there were significant differences found as follows: SB > UP ($p<0.01$, SB > SF ($p<0.01$). In ST, there was a significant difference between SF and UP (SF > UP, $p<0.01$) (Table 2).

ANOVA showed a significant difference in TSP among the four different standing conditions ($F = 6.264$, $p<0.01$) (Table 3). Bonferroni's multiple comparisons test revealed significant differences between the conditions as follows: SB > UP ($p<0.05$), SB > PS ($p<0.01$), SB > SF ($p<0.01$).

DISCUSSION

Falling is a major concern for lower limb amputees. Balance control is one of the abilities that need to be acquired for standing or walking without falling. The assessment of muscle activities and COP during upright standing is usually performed to evaluate postural control ability. In the present study, to assess the postural control

Table 2. Average of % RMSEMG during Standing

Type of Muscles	Standing Positions			
	UP	PS	SF	SB
TA[a]	6.69 ± 6.03[#]	16.21 ± 16.65[µ]	22.68 ± 18.94[†]	40.09 ± 17.6[#µ†]
GA[a]	10.32 ± 5.96[¶]	32.35 ± 17.60[§]	109.19 ± 54.64[¶§†]	36.67 ± 20.86[†]
VL[a]	6.15 ± 2.53[#]	23.86 ± 30.39	21.25 ± 21.44	47.96 ± 20.54[#]
VM[a]	6.91 ± 4.01[#]	13.36 ± 13.59[µ]	10.88 ± 8.92[†]	36.49 ± 25.40[#µ†]
ST[a]	6.05 ± 4.60[¶]	20.83 ± 20.39	33.71 ± 19.92[¶]	20.33 ± 9.44

N=8 (mean ± SD)

Average of %RMSEMG in tested muscles during standing positions.
Abbreviations; TA, tibialis anterior; GA, gastrocnemius; VL, vastus lateralis; VM, vastus medialis; ST, semitendinosus; UP, upright position with both feet; PS, prosthetic standing; SF, shift to forward; SB, shift to backward.
[a]: Significant main effect for standing positions ($p<0.01$). [#]: Significant difference between UP and SB.
[¶]: Sig nificant difference between UP and SF. [µ]: Significant difference between PS and SB.
[§]: Significant difference between PS and SF. [†]: Significant difference between SF and SB.

Table 3. Average of Total Length of Sway Path (TSP) during Standing

	Standing Positions			
	UP	PS	SF	SB
TSP (mm)[a]	1025.10 ± 188.81[#]	1010.66 ± 195.61[µ]	1005.23 ± 188.62[†]	1121.20 ± 116.72[#µ†]

N=8 (mean ± SD)

Average of TSP during tested standing positions.
Abbreviations: TSP, total length of sway path; UP,upright position with both feet; PS, prosthetic standing; SF, shift to forward; SB, shift to backward.
[a]: Significant main effect for standing positions ($p<0.01$). [#]: Significant difference between UP and SB.
[µ]: Significant difference between PS and SB. [†]: Significant difference between SF and SB.

during prosthetic standing, healthy subjects wore a simulated prosthesis on their right thigh as if they were recent right trans-femoral amputees. It was the first time the subjects wore a simulated prosthesis, so postural stability would have been directly affected by this first experience of prosthetic standing. Practice was not permitted prior to the recorded trials, in order to investigate what compensatory strategies are employed during a first prosthetic standing experience.

During the first task of UP, the subjects placed their weight evenly on both feet. This UP condition therefore took less muscular effort to maintain postural stability. There was no significant difference between UP and PS for all the recorded muscle activities. Even in PS, when the subjects maintained upright standing with a simulated trans-femoral prosthesis, all of the muscles required relatively low effort to maintain postural stability because the line of gravity passed directly through the center of each joint. According to Tokuno et al.[13]), the line of gravity falls approximately through the knee joint axis and the moment arm of the line of gravity is zero in the standing position with the knee in full extension. No muscular force is needed to maintain equilibrium in this condition. In the present study, therefore, the least stress would have been placed on the muscles and ligaments during static standing conditions like UP and PS. In addition, the finding that TSP showed no significant difference between UP and PS also indicates the line of gravity of the entire body remained within the base of support to maintain stability during static standing.

However, if one segment moves forward, another must move backward to maintain stability. According to Esquezazi and DiGiacomo[11]), the use of a prosthesis by trans-tibial amputees requires further activity of the biceps femoris during the stance period to improve the support of the amputated leg knee joint. Gatev et al.[14] also showed that the activity of the lateral gastrocnemius muscle anticipated the antero-posterior movement of COP during quiet standing. In the present study, when subjects shifted their weight forward, the activities of posterior muscles such as ST and GA increased. There were significant differences between SF and the other task positions, especially in the activity of GA. These results suggest the posterior muscles of the lower extremity have an important role in maintaining the stability of forward movement. On the other hand, when the subjects shifted their weight backward, the activity of anterior muscles such as TA increased gradually. The stabilization of balance during quiet standing was achieved by the stiffness of ankle muscles[15]. This "ankle strategy" is an important mechanism for controlling balance and it occurs in quiet standing and during small perturbations. The results for GA and TA in the present study indicate that the stiffness of ankle muscles strongly is related to maintaining quiet standing. Lower extremity amputation, however, eliminates the work of the unilateral leg ankle strategy because the calf and other leg muscles are absent. In the present study, it was assumed that the subjects had to utilize an ankle strategy only on the left sound leg and utilized different strategies to maintain the standing posture. When the ankle muscle activity is insufficient, a hip strategy will flex the hip to compensate for the anterior movement of the center of mass, or to extend the hip to prevent excessive posterior movement. The resultant significant activity of ST in SF and that of VL and VM in SB suggest such a hip strategy was employed by the subjects in this study. Generally, VM works mainly to maintain knee extension. In the present study, however, VL may have been more active than VM during standing with the prosthesis in some tasks. It is probable that the center of gravity always shifted laterally toward the sound side, in all tasks of prosthetic standing. This weight bearing shift possibly induces more activity in VL than VM. The significant difference of VL activity only seen between SB and UP may suggest VL also worked in PS and SF to some extent. Conversely, the activity of VM was needed only in SB because there were significant differences between SB and the other standing positions. Therefore, these results indicate VL works mainly to maintain standing in all conditions of prosthetic standing and this will be one of compensatory hip strategies employed by subjects experiencing of prosthetic standing for the first time.

An alternative interpretation has been proposed to explain the compensatory strategy. Tokuno et al.[13] explained the lack of posterior displacement as a compensatory action limiting the body's disequilibrium. By limiting displacement, persons with below-knee amputation would experience less disequilibrium and, hence, diminish the chance of falling or stumbling during gait initiation. In the present study, the task of backward weight shift may have been difficult for the subjects because they wore the simulated prosthesis for the first time, and therefore were unfamiliar with controlling an artificial knee joint to maintain extension. Fear of falling occurring when shifting the center of mass backward would affect their movements. If subjects had bent the left knee to shift weight backward, excessive bending of the prosthetic knee joint would easily cause a fall. Consequently, they would have had to extend both knee joints to maintain a standing position. The findings that the knee extensors contracted more during SB than during UP and PS is in agreement with this supposition. Furthermore, the significantly large value of mean TSP in SB also indicates SB was the most difficult task of the standing conditions.

To maintain upright standing, humans usually use multiple sensory systems, including the vestibular system, visual system and somatosensory system. These major sensory systems are related to the control of balance and posture. Postural control via sensory systems includes both open-loop and closed-loop mechanisms. According to Laughton[16] , closed-loop control operates with sensory feedback, while open-loop control needs no sensory feedback. Proprioceptive information from the plantar sole plays an important role in the control of posture and gait[17,18]. In the present study, it is evident that all of the subjects could not use this specific form of sensory feedback from the leg with a simulated prosthesis. Furthermore, most of the subjects showed longer TSP in SB. Feedback information not only from the proprioceptive system but also from the visual system may not be utilized sufficiently when performing the task of SB. Longer TSP in SB would also have resulted from the lack of closed-loop control.

The results of the present study suggest that an additional hip strategy was employed by subjects for the sound leg during the first experience of prosthetic standing. We suggest it was originated by dysfunction of peripheral systems, fear of falling, lack of feedback information from sensory systems and other factors. Amputees need to adapt toimpaired physical and psychological conditions and to acquire coordinated muscle activity as a new strategy to improve balance control. Therefore, physical therapy intervention should focus on integrating muscle activity and training balance strategies. Further study is necessary to determine the change and learning process of compensation for balance control in amputees during the rehabilitation period.

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