J Rehab Med 2001; 33: 196–199

KNEE MUSCLE ACTIVITY DURING AMBULATION OF TRANS-TIBIAL AMPUTEES

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Gait analysis of trans-tibial amputees brings to light asymmetries of different gait parameters between the amputated and sound legs. The present study investigated the activity of the vastus medialis and biceps femoris muscles during ambulation of trans-tibial amputees. Peak activities of the vastus medialis were reached similarly in both legs (6.06 ± 4.9% and 8.84 ± 3.6% of gait cycle, in the sound and amputated leg, respectively). Biceps femoris peak activities were reached at 92.43 ± 6.6% of gait cycle in the sound leg, and significantly later (at 9.81 ± 4.8% of gait cycle) in the amputated leg ($p < 0.05$). Integrated EMG activity ratios, between swing and stance periods, were similar for the vastus medialis (0.33 in the sound and 0.35 in the amputated leg). However, these ratios differed significantly for the biceps femoris since the amputated leg presented a substantial ($p < 0.05$) smaller ratio (0.22) compared with the sound leg (0.83). The use of prosthesis in trans-tibial amputees requires further activity of the biceps femoris during stance period to improve support of the amputated leg knee joint.

Key words: trans-tibial amputee, gait analysis, electromyography.

J Rehab Med 2001; 33: 196–199

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(Accepted December 5, 2000)

INTRODUCTION

Successful prosthetic rehabilitation is possible in traumatic trans-tibial (TT) amputees. Unlike vascular patients, traumatic amputees are often young and enjoy a good state of health, and therefore an optimal prosthetic fitting enables them to freely ambulate. However, although the quality of ambulation among these amputees is good, gait analysis still shows asymmetries between the amputated and sound legs. These asymmetries, related mainly to the rigid prosthetic ankle–foot component, are manifested as follows. (a) Time-distance and kinematic parameters: in TT amputees during ambulation at a mean speed of 1.34 meters/second, stance period, step length, and double-limb support time are significantly shorter in the sound limb compared with the amputated limb (1, 2). Knees angle of the amputated leg is significantly smaller at loading response phase and significantly greater at toe-off (2). (b) Standing equilibrium: standing balance activity was evaluated in both lower limbs of TT amputees. Results showed that the anterior–posterior foot–ground reactive forces generated by the amputated limb were significantly smaller compared with the sound leg (3, 4). Since loss of the ankle joint and relevant muscles interfere with normal distribution of standing weight bearing, the compensating sound leg muscles become dominant in maintaining standing equilibrium. (c) Decrease in muscle volume and strength: a significant atrophy of the thigh muscles in the amputated TT limb was demonstrated from thigh circumference measurements and from quadriceps muscle biopsies. Quadriceps and hamstrings muscle strength measurements also showed a significant weakness in the amputated leg (5–7).

We assumed that in addition to the loss of proprioception and to muscle weakness in the amputated leg, the rigid prosthetic ankle foot unit interferes with normal gait kinematics and with normal activity of the muscles. Deceleration of the leg during terminal swing and stability of the knee during the task of weight acceptance depend mainly on knee extensors and flexors. Since asymmetry of various gait parameters characterizes TT amputees' gait, we chose to focus our attention on activity of the quadriceps and hamstrings muscles to better comprehend lower limbs course of action during ambulation.

METHODS

Subjects

Eleven males, who met the following criteria, were recruited for the study: traumatic unilateral below-knee amputation, pain-free stump with no skin abrasions, excellent walkers who used their prostheses on a regular basis and conducted an active normal life.

The ages of the subjects ranged from 27 to 58 years (mean age 37.4 ± 8 years); heights ranged from 167 to 184 cm (mean height 177.5 ± 7.2 cm); weights ranged from 65 to 96 kg (mean weight 79.3 ± 7.2 kg). Length of legs was measured before the test to assure equal length of both the amputated and normal leg. Length of lower limbs was measured as follows: each subject stood on both legs with knees fully extended, feet 20 cm apart and his back and heels in contact with the wall. The same researcher (E.I.) measured the distances between the anterior superior iliac spines and a point that divides in two the medial edge of the sole of each shoe.

The time lapse between the date of amputation and the time of testing ranged from 5 to 44 years, with a mean time lapse of

 ISSN 1650–1977

Table I. *Ratios of activity (iEMG) of vastus medialis and biceps femoris between different gait cycle periods*

	Swing 1st half / Stance 2nd half		Swing 2nd half / Stance 1st half		Swing / Stance	
	Sound	Amp.	Sound	Amp.	Sound	Amp.
Vastus medialis	0.30	0.41	0.34	0.32	0.33	0.35
Biceps femoris	0.57	0.15*	1.04	0.28*	0.83	0.22*

* $p < 0.05$.

12.8 ± 11.1 years. All amputees prostheses were patellar-tendon-bearing (PTB) with a solid-ankle-cushion-heel (SACH) foot.

METHODS

Before testing, all subjects were assessed by a technician to ensure optimal fit and function of the prosthesis. Subjects were instructed to ambulate at their most comfortable speed along a 10 meters long × 1 meter wide non-slip conductive rubber walkway with 10 Ω/meter. Ambulation speed and time-distance parameters were calculated from signals obtained from the electric contact system walkway.

Electromyographic activity of the vastus medialis and the medial hamstrings (biceps femoris) was recorded bilaterally using Beckman surface electrodes. Each pair of electrodes was positioned over the muscle belly, 2 cm apart. Shaving and abrasion of the skin reduced the skin-electrode interface impedance to values lower than 10 Ω in all cases. The EMG signals were transmitted through an overhead cable system to a processing and recording unit (Myosystem 2000, Noraxon Inc., Scottsdale AZ, USA). Subsequently, the raw EMG signals were full-wave rectified to allow quantification, and filtered by low-pass filter (5 Hz) to provide the linear envelope representative of the EMG signal. This processing of the signal rendered an envelope that represented a profile of the activity of the muscle in time.

The EMG activity of the vastus medialis and biceps femoris in the amputated and sound legs was simultaneously recorded with the time-distance parameters. Timing of the peak activities of the relevant muscles, expressed in percentage of gait cycle (GC) time, was determined from the obtained muscles EMG linear envelope. Ratios of the integrated EMG of the monitored muscles were calculated for each leg as follows: (1) Between total periods of swing and stance. (2) Between the first half of swing and second half of stance period. (3) Between the second half of swing and first half of stance period. Foot switches allowed simultaneous recording of the foot–floor contact timing with the EMG.

The significance level of the differences was determined by using Student's *t*-test paired. Results were deemed to be statistically significant at $p < 0.05$.

RESULTS

The average number of steps analyzed in each subject was 21.2 ± 3.3, obtained during an uninterrupted walk along the laboratory walkway at a mean speed of 0.95 meter/second (range: 0.82–1.02 meter/second). Data of peak activity of the vastus medialis and biceps femoris reflect means of results obtained from the average of the steps analyzed. The vastus medialis muscle reached peak at 6.06 ± 4.9% of GC in the sound leg and at 8.84 ± 3.6% of GC in the amputated leg. The biceps femoris reached peak activity in the sound leg at the end of swing period and somewhat before heel strike, at 92.43 ± 6.6% of GC. Peak activity of the same muscle in the amputated leg was reached only following HS, at 9.81 ± 4.8% of GC. This difference was significant ($p < 0.05$).

Table I includes the calculated ratios of iEMG of the vastus medialis and biceps femoris. Ratios of the vastus medialis activity during the first half of swing and the second half of the stance period in the sound and amputated leg (0.30 and 0.41, relatively) were not significantly different. These ratios, 0.57 in the sound leg and 0.15 in the amputated leg, were found to be significantly different for the biceps femoris. Ratios of the vastus medialis activity during the second half of swing and the first half of stance period in the sound and amputated leg (0.34 and 0.32, relatively) were almost similar for the vastus medialis. These ratios, 1.04 in the sound leg and 0.28 in the amputated leg, were found to be significantly different for the biceps femoris. Ratios of iEMG activity during swing and stance periods, obtained for the vastus medialis, were almost similar (0.33 in the sound and 0.35 in the amputated leg). Hence, these inter-legs ratios differ significantly for the biceps femoris since the sound leg presented a substantial higher ratio (0.83) compared with the amputated leg (0.22).

DISCUSSION

Normal gait cycle is characterized by repetitious sequence of limb motion with well-defined timing of onset and ending of known gait periods, tasks, and phases. It was found that symmetry of some time–distance parameters between normal subjects' legs ranged between 0.90 and 0.98, while symmetry of the same parameters in TT amputees decreased to a range between 0.74 and 0.92 (2, 8).

Since asymmetry of various gait parameters in TT amputees is well known, it can be assumed that asymmetry characterizes the activity of muscles in the lower limbs. To verify this assumption, the present study was addressed to investigate the vastus medialis and biceps femoris muscles, which control the knee joint.

In healthy subjects' gait, the vastus medialis activity was found to be exceptionally consistent during the swing-stance transition period (9). Activity of the vastii begin in terminal swing (90% of GC), reaching peak intensity in early loading response at 5% of GC, and standstill from 15% of GC. This muscle acts eccentrically in early stance, providing the kinematic evidence that flexion of the knee occurs during loading response (10). In TT amputees, the difference between peak activities timing was insignificant since the vastus medialis reached peak activity at 6.06 ± 4.9% of GC in the sound leg, and at 8.84 ± 3.6% of GC in the amputated leg.

Also, the cushioning effect of the knee during weight acceptance is related to the amplitude of the joint range of movement, controlled by the eccentric activity of the vastus medialis. It was shown that the maximal knee flexion obtained in TT amputees during loading response was 14.4 ± 4.1° in the sound leg, while this angle measured in the amputated leg knee joint (7.9 ± 3.8°) was significantly smaller (2). Therefore it can be assumed that the vastus medialis role in attenuation of the initial contact impact is present to smaller extent, and that the muscle is less active in the amputated leg.

During normal gait, the hamstring muscles reach peak activity at late mid- and terminal swing, acting eccentrically to decelerate the passively extending shank (10). In the present study, the sound leg biceps femoris muscles reached peak activity just before HS, at 92.43 ± 6.6% of GC. In the amputated leg the peak activity of the same muscle was reached exclusively after HS, at 9.81 ± 4.8% of GC, and this difference was found to be significant. The timing variation of the biceps femoris may also depend on the muscle strength, since it is known that the amputated leg thigh muscles' strength and volume decrease significantly in TT amputees (5–7). Strength of hamstrings and quadriceps was not measured in this study and it was therefore not possible to correlate it with biceps femoris activity. Stump length of all amputees was measured during the physical examination that anteceded the tests. There was no correlation between stump length and the activity timing variation of the biceps femoris.

The delayed peak activity of the muscle was also evident from the significantly different ratio activities. In the sound leg, the biceps femoris was equally active during the second half of swing and the first half of stance. Nevertheless, this muscle in the amputated leg was almost 4 times more active in the first half of stance. The significant extensive activity of the biceps femoris in the amputated leg is also maintained during the second half of stance compared with the first half of swing period. It can be concluded from the obtained swing/stance periods ratio activity of the investigated muscles that during stance the quadriceps is similarly 3 times more active in both legs while the biceps femoris is 4 times more active in the amputated leg only. This explains why atrophy of the hamstrings muscle in TT amputees was found to be less pronounced than that of the quadriceps muscles. Furthermore, in the amputated leg a lesser reduction in the knee flexors' strength compared with the knee extensors was found (7).

The overall principle of lower limb support during stance phase was extensively investigated and described (11). Support during stance is achieved by a net extensor pattern of moments at the ankle, knee and hip joints. As long as the net supporting moment in the lower limb is positive there is a net tendency to support the limb and prevent collapse. When one joint opposes or does not contribute to lower limb support, one or both of the other joints will compensate for the non-contributing joint. Therefore, collapse during weight bearing is prevented by collaboration of muscle at all three joints of the lower limb. The pattern of the ankle joint moment is the only one to be consistent and it is characterized by a small dorsi flexion moment followed by a strong plantar flexion moment (11). Relying on the above-described principles, the ability of the TT amputee to support himself with a prosthesis during gait can be accomplished only by compensation of the absent ankle muscles. Hence, the required large extensor hip moment is reflected by the compensating extensive activity of the biceps femoris in the amputated leg, as detected in this study.

The obtained asymmetries in timing activities of the biceps femoris during TT amputees' gait can also be interpreted with respect to the suggested whole-body inverted pendulum model (12). In a healthy subject's gait, total body balance in the coronal plane is achieved primarily by the placement of the foot relative to the horizontal location of the center of mass. Foot placement is dependent upon the hip abductor/adductor moments generated during swing while foot placement errors are corrected at the level of the supporting subtalar or hip joints that work in synergy. Control of head, arms and trunk balance about the supporting hip is achieved by the hip abductors while spinal muscles and the coupling between the pelvis and upper trunk regulate posture of the trunk.

The model of whole body balance control can not be implemented in TT amputees since the anatomical ankle joint of the amputated leg is missing. Hence, the obtained differences between the recruited muscles, which act upon the supporting hip joints to control whole body balance, are reflected by the asymmetries in timing activities of the biceps femoris.

In TT amputees, the almost rigid ankle–foot prosthesis's component negatively interferes with the normal progress of gait. During weight acceptance, the ankle does not plantar flex and the plantar flexors controlling the forward rotation of the shank are missing. In order to prevent collapse of the entire supporting limb, the hip extensor activity is reinforced to co-act with the knee extensors. Indeed, the present study sustains the principle of lower limb support in case of usage of prosthesis with a rigid ankle–foot unit. The extensive compensating activity of the biceps femoris during the entire stance period, and especially during weight acceptance, characterizes the TT amputees' gait.

In conclusion, some characteristics of the knee muscles activity during ambulation of TT amputees are described. Asymmetry between the hamstrings activity, as well as asymmetry in other gait parameters, is seemingly related to the prosthesis stiffness.

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