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Dynamic analysis of above-knee amputee gait

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Abstract

Background. It is important to understand the characteristics of amputee gait to develop more functional prostheses. The aim of this study is to quantitatively evaluate amputee gait by dynamic analysis of the musculoskeletal system during level walking and stair climbing.

Methods. Dynamic analysis using gait analysis, electromyography and musculoskeletal modeling for above-knee amputees ($n = 8$) and healthy adults ($n = 10$) was performed to evaluate the muscle balance, muscle force, and moment of each major muscle in each ambulatory task. Time–distance parameters and the kinematic parameter of gait analysis were calculated, and a root mean square electromyogram of major muscles and hamstring and tibialis anterior coactivity was measured using electromyography. Lastly, dynamic analyses of above-knee amputee gaits were performed using musculoskeletal models with scaled bones and redefined muscles for each subject.

Findings. Most kinematic parameters showed statistically no difference among the tasks, excluding pelvic tilt, pelvic obliquity, and hip abduction. Major muscle activities and coactivities of the hamstring and tibialis anterior showed that the stair ascent task needed more muscle activity than the stair descent task and level walking. The muscle activity and coactivity of amputees were greater than those of healthy subjects, excluding the hamstring coactivity during stair ascent ($P < 0.05$). Lastly, dynamic analysis showed that weakened abductor and excessive adductor and then inadequate torque during all tasks were quantitatively observed.

Interpretation. Dynamic analysis of amputee gait enabled us to quantify the contribution of major muscles at the hip and knee joint mainly in daily ambulatory tasks of above-knee amputees and may be helpful in designing functional prostheses.

Keywords: Above-knee amputee; Stair climbing; Level walking; Electromyography; Musculoskeletal model; Dynamic analysis

1. Introduction

Above-knee amputation surgery has been the standard method of treatment for most soft-tissue and bone sarcomas. In Korea, 3.9% of injured workers were reported to be above-knee amputees, according to the annual report of the Ministry of Health and Welfare for physically disabled persons. Above-knee amputees tend to wear the prosthesis–socket system, which enables them to ambulate functionally instead of orthoses such as a wheelchair, crutch, and so on. Therefore, after amputation surgery, the patients are requested to repeatedly practice level walking and stair climbing with their own prosthesis–socket system at the amputated part of the lower extremity to improve ambulation ability. But amputee gait, including level walking and stair climbing, is a problem for the elderly and infants because of feelings of insecurity and fear of secondary disorder. While the elderly and infants can overcome these obstacles by reducing gait speed, most

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amputees wholly rely on muscle condition at the dissected limb, called a stump. However, many amputees want to have more functional prostheses to speed up recovery time and activities of daily living (ADL) adaptation rather than to strengthen stump muscles. Thus, when investigating the characteristics of amputees during ADL, it is important to first understand the muscle volume and forces of them.

Muscle volume and forces of amputees were frequently observed in both management of post-surgical treatment for dissected muscles and muscle adaptation after rehabilitation therapy for ADL motion. The condition of the post-surgical muscles depends on the surgeons who perform the amputation surgery. Therefore, dissected muscles after surgery were needed to be re-estimated by MRI (Zhang et al., 1998) and sonomyography (Zheng et al., 2006). Muscle adaptation by an amputee's intact limb was needed to compensate for the lack of stump muscle force during ambulatory motion (Seroussi et al., 1996). Recently, it has also been reported that muscle adaptation for ADL motion depends on the amputated level of the lower limb (Schmalz et al., 2007). Therefore, it is necessary to redefine the muscle volume and force for dissected muscles to understand and analyze amputee gait. However, related research is rare. To improve the efficiency of rehabilitation therapy through amputee gait, integrated approaches are needed that consider not only muscle condition but also dynamic movement of the musculoskeletal system for amputees. However, there have been few studies conducted on amputee gait of patients with orthopedic implants and amputees with artificial limbs. Recently, several studies that considered musculoskeletal condition have been reported for healthy subjects (Heller et al., 2003, 2001), but research on amputees with prostheses is still rare. Therefore, musculoskeletal models of amputees with prostheses were needed to evaluate quantitatively the contribution of each muscle to amputee gait. The aims of this study were primarily to evaluate the muscle condition by acquiring the root mean square electromyogram (RMS EMG) and secondarily to predict muscle forces and moments using three-dimensional musculoskeletal dynamic models of above-knee amputees with prostheses for level walking and stair climbing tasks. In addition, we wanted to evaluate the habitual gait of above-knee amputees to keep pace with the gait speed of healthy persons by measuring muscle activity during two-stairs ascent and two-stairs descent tasks at one time.

2. Methods

2.1. Subjects

The experimental data were collected from eight above-knee amputees (right side) with lower-limb prostheses; each subject had displayed volume stability of the residual limb for at least 2 years and had no skin problem of the stump prior to participation in this study. They had used the same prosthesis for than 5 years on average. Ten healthy individuals served as the control group, free of any musculoskeletal or neurological dysfunction that would affect gait (Table 1).

Table 1
Subject characteristics mean (SD)

	AKA group ($n = 8$)	Control group ($n = 10$)
Age (yrs)	39.88 (7.83)	24.35 (1.73)
Height (cm)	168.01 (4.14)	174.65 (0.81)
Weight (kg)	67.56 (5.54)	64.46 (1.47)

2.2. Simulation procedures

2.2.1. Motion analysis

Motion analysis was performed by using a three-dimensional motion analyzer with seven infrared cameras (VICON 370, Oxford metrics Ltd., Oxford, UK) and two CCD cameras (HDV1080i, SONY, Tokyo, Japan). Fifteen 25-mm reflective markers for the sound limbs of the amputees and both limbs of the healthy subjects were placed on the sacrum, anterior superior iliac spine (SIS, bilaterally), lateral femoral epicondyle (bilaterally), calcaneous and malleolus (bilaterally), metatarsal head (bilaterally), and the lower lateral 1/3 surface of both shanks and thighs (bilaterally) using a wand, respectively. Marker placement on the prosthesis was estimated by using the bony landmarks on the sound limb. All kinematic data were sampled at 60 Hz using a personal computer.

Prior to the experiments, anthropometric measurements (height and weight) of the lower extremity were performed for all subjects (Table 1). Later, the subjects walked across the level walkway until they were accustomed to the level walking task. The healthy subjects walked barefoot while amputees walked with shoes and both walked at the self-selected speed on a 15 m gait pathway that was instrumented with two force-plates (900 mm × 600 mm, Kistler Instrument Corp., NY, USA) to measure the ground reaction force.

For stair climbing, a wooden staircase was custom-built. The staircase with an inclination of 30° included three stairs (height 160 mm, width 300 mm), and two piezoelectric force-plates (400 mm × 600 mm, Kistler Instrument Corp., NY, USA) were embedded in the second and third stair respectively to record the forces generated during stair accent and descent. And another force-plate was embedded on the ground level. Force plate data were sampled at 2500 Hz. The starting points for the stair ascending and descending tasks were in front of the staircase on the ground level and at the top of the staircase, respectively. The dominant foot of each subject was used during these tasks. We analyzed the stair climbing task for a stair stride cycle. During ascent, a stride cycle was defined starting with foot contact on the second step and ending at the next foot contact on the fourth step. During descent, the selected strides started with foot contact on the third step and ended with foot contact on the first step. Foot contact always occurred with the same foot among all subjects (Riener et al., 2002). For the stair climbing task, subjects practiced for several minutes until they were relaxed and felt that they performed a natural motion. Five successful trials were collected for all of the tasks respectively, and each trial was considered appropriate only if one foot per one force-plate was measured. Using a post-process program (Polygon, Oxford metrics Ltd., Oxford, UK), we could get the mean time–distance parameters for five trials of each task.

2.2.2. Musculoskeletal model

The generic musculoskeletal models for the subjects were generated using transformation software (Real-time Motion Module, MusculoGraphics, CA, USA), which enables us to scale the body segments, joint kinematics, and muscle attachment sites of the model to match the size of the subject using the data of captured marker position (see Fig. 1).

Unlike models for healthy persons, several modifications were needed for the amputees' prosthetic limbs. For their anthropometric data, computed tomography (GE Hi-speed, GE healthcare, CT, USA) was taken at 1-mm intervals. The length of the amputated femur was measured using imaging software for real-time 3D visualization based on a PC (Vworks 3.0, Cybermed Inc., Seoul, Korea). MRIs (magnetic resonance imaging) were taken to evaluate the postoperative muscle closure on the end of stump and identified state of muscle closure was applied to model the muscles around the amputated limb bone. With the help of a clinician and technical consideration of myoplasty, the suturing of the ends of residual muscles over the end of the bone, the distal portions of the stumps were reconstructed by restricting the muscle motion using wrapping constraints (Charlton and Johnson, 2001) for four muscle groups, in order: (1) vastus intermedius, (2) adductor magnus, (3) semimembranosus and short head of biceps femoris, and (4) rectus femoris, sartorius, gracilis, semitendinosus, long head of biceps femoris, and tensor fasciae latae. Muscles are defined as biomechanical elements that have lines of action spanning from origin to insertion based on descriptions from the literature (Brand et al., 1986; Duda et al., 1996) and that can generate forces and moments with four parameters and three curves. The four parameters such as peak isometric muscle force (Charlton and Johnson, 2001), optimal muscle-fiber length, pennation angle (Friederich and Brand, 1990), and tendon slack length (Delp and Zajac, 1992; Delp and Loan, 1995; Delp et al., 1990) were defined from the literature. The curves that define the muscle's active force–length relationship, its passive force–length relationship and the force–length relationship of the tendon were taken from the literature (Delp and Zajac, 1992; Delp and Loan, 1995; Delp et al., 1990).

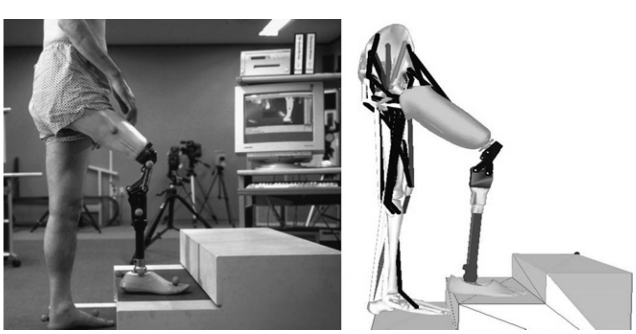


Fig. 1. Clinical experiment (left) and musculoskeletal modeling (right).

2.2.3. Prosthesis model

The prosthesis with the four-bar linkage mechanism consists of 13 components and four joints; the components are connected by a socket system fitted to the stump model. The anthropometric data for the prostheses were obtained from their CAD layout and their masses, mass centers, and moments of inertia were specified for dynamic analysis. In addition, the joint motion was defined as the function of knee flexion angle using the four-bar linkage mechanism.

2.2.4. Dynamic simulation of musculoskeletal structures

To analyze amputee gait using musculoskeletal models during level walking and stair climbing, dynamic parameters such as mass, center of mass, and moment of inertia of the bones, muscles, and components of the prosthesis, respectively, needed to be defined. The bone segments, body height, body mass, and length of the thigh, shank, and foot of each subject were calculated following Winter et al. (Winter, 1984), and the dynamic parameters of the prosthetic socket, shank, and foot were calculated using Adams software (Version, 2003, MSC, CA, USA). In addition, we applied 6° of freedom to each joint and constrained the range of motion to avoid excessive motion. In this study, all muscles at the hip joint were included to simulate hip joint motion such as abduction–adduction and flexion–extension. The muscles of the amputated leg were modeled in consideration for muscle reconstruction; each muscle contraction model was governed by a Hill-type model formulation (Schutte et al., 1997). Musculoskeletal dynamic models for each model were imported and scaled by motion module and SIMM software (Version 4.1.1, Musculographics, Inc., IL, USA). The dynamic equations of motion for the amputees and healthy subjects were derived using SD/FAST (Version 3.3.1, Symbolic Dynamics, Inc., CA, USA), and an inverse dynamics simulation for each model was produced by Dynamics Pipeline (Version 3.0, MusculoGraphics, Inc., IL, USA). The time history of joint angles and ground reaction forces using gait analysis data was used as an input to the solver.

To validate the musculoskeletal models for each task, we wanted to compare the results of the experiments, those of the simulations using cadaveric muscle parameters, and those of the simulation using optimized muscle parameters by static optimization method (Anderson and Pandy, 2001a, Anderson and Pandy, 1999; Anderson and Pandy, 2001b). The joint torques were calculated in the sagittal, frontal, and transverse planes using a proportional-derivative (PD) controller to find corrective torques that will keep the simulation following the input motion, and, lastly, the muscles' forces were estimated using the change of muscle fiber length and moment arm during motion.

2.2.5. Muscle activity and coactivity

Muscle activity was measured by RMS EMG during level walking and stair climbing. To record muscle activity, disposable, self-adhesive Ag/AgCl dual snap electrodes (4×2.2 cm, Noraxon System Inc., AZ, USA) were attached on the muscle bellies of the vastus medial (VM), vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF), semitendinosus (ST), gluteus maximus (GM), soleus (SOL), tibialis anterior (TA), and gastrocnemius lateralis (GA) for the amputees' healthy limbs. For the healthy groups, BF, VL, TA, and GA were involved only to compute muscle coactivity. A 12-channel EMG instrument (Myosystem 1400, Noraxon system Inc., AZ, USA) was used to gain the EMG signals using active EMG lead (1 m) with pre-amplifier (gain 500) and 10–1000 Hz band-pass (part #243 and part #242, Noraxon system Inc., AZ, USA). The band pass filter of 30–400 Hz and band stop filter of 60 Hz were used to reduce noises, and the data sampling rate for each channel was 1024 Hz. To normalize EMG activity measured during level walking and stair climbing, all subjects performed maximal voluntary isometric contractions on a dynamometer (System 3 Pro, Biodex Medical System, NY, USA). Subjects sat on the dynamometer seat, and two shoulder belts and one lap belt secured the upper body. The maximum voluntary isometric contraction testing was performed at 10°, 30°, 45°, 60°, 90°, and 105° for the hip, at 10°, 30°, 50°, 70°, 90°, and 110° for the knee (hip angle 90°), and at 30°, 20°, 10°, 0°, −10°, and −20° at the ankle. While extensors and flexors of hip and knee joints of the right legs of the healthy group were involved in testing, the hip joints at the sound limbs of the amputee group were tested excluding the knee joint of the prosthetic limb. The knee joints at the sound limbs of the amputee group were included in the experiment to evaluate habitual gait of above-knee amputees during two-stairs ascent and two-stairs descent task at one time. During testing, subjects were instructed to perform maximal effort concentric and eccentric contractions. After that, isokinetic contraction testing was subsequently executed at 60° for both joints. To estimate muscle balance, Hamstring coactivity ratios I and II used in the literature (De vito et al., 2003; Hortobagyi et al., 2005) were computed as the quotient of biceps femoris RMS EMG activity divided by vastus lateralis RMS EMG activity multiplied by 100 (BF/VL), referred to in the literature. Gastrocnemius lateralis coactivity relative to the activity of the tibialis anterior (GA/TA) are also calculated during level walking and stair climbing.

In addition, amputees tend to gait rapidly at level walking and stair descent and to climb two stairs at once to keep pace with the healthy. It is speculated that these gait patterns cause great joint load at the ankle, knee, and hip joints in the sound leg due to the absence of function at the amputated leg. Therefore, the effect of these gait patterns was analyzed using the peak RMS EMG of major muscles during level walking (LW) and stair climbing tasks, which divided one-stair climbing and two-stairs climbing per step. SA1 and SD2 represented one-stair ascent and one-stair descent tasks, respectively. In addition, two-stairs ascent and descent at one time were abbreviated to SA2 and SD2, respectively.

2.3. Statistical analysis

For task performance of two groups (healthy persons and amputees), the analysis of variance was accomplished using Minitab software (release 13, Minitab Inc., PA, USA) with respect to task types (level walking and stair climbing) and the coactivity ratio (BF/VL and GA/TA). A Tukey's post hoc contrast was used, and significance was set at $P < 0.05$ to determine significant differences between the mean values.

3. Results

3.1. Gait cycle parameters

For level walking and stair climbing, the mean time–distance parameters were calculated by gait analysis (Table 2). During level walking, all parameters of the healthy group were statistically different from the parameters of the amputees ($P < 0.05$). Especially, the pelvic obliquity of amputees was lower than that of the healthy group as one of the gait characteristics of above-knee amputees. Unlike level walking stance phase, there was no differences between the groups during the stance phase of stair climbing. All kinematic parameters during ascent showed no significant difference, excluding pelvic tilt, pelvic obliquity, and hip abduction ($P > 0.05$). During stair descent, the hip adduction and abduction of amputees were statistically greater than that of the healthy group. From the results, the typical gait characteristics (lateral bending gait) of above-knee amputees could be found quantitatively.

3.2. Validations for musculoskeletal models

To validate the musculoskeletal models, the torque data at the knee joint were compared between the experiment and simulations with cadaver and optimized muscle parameters. The results showed that knee joint torque in the experiment was about 27% greater than that in the simulation using cadaver muscle parameters and was about 0.3% lesser than that in the simulation using optimized muscle parameters on average (Fig. 2). Therefore, optimized muscle parameters were used to perform dynamic simulation in this study.

Table 2
Mean (SD) time–distance and kinematical parameters for level walking and stair ascent/descent

	Level walking		Stair ascent		Stair descent	
	Healthy ($n = 20$)	Amputee ($n = 8$)	Healthy ($n = 20$)	Amputee ($n = 8$)	Healthy ($n = 20$)	Amputee ($n = 8$)
Time–distance parameters						
Gait speed (m/s)	1.36 (0.99)	0.82 (0.15)	0.49 (0.14)	0.35 (0.17)	0.87 (0.14)	0.65 (0.48)
Cadence (step/min)	112.08 (1.68)	88.23 (8.92)	94.08 (9.36)	87.18 (15.14)	108.96 (8.04)	96.69 (17.49)
Cycle duration (s)	1.01 (0.03)	1.62 (0.20)	1.28 (0.14)	1.27 (0.04)*	1.10 (0.09)	1.15 (0.10)*
Stride length (m)	1.39 (0.10)	1.29 (0.16)	0.63 (0.18)	0.47 (0.18)	0.96 (0.13)	0.75 (0.48)
Stance phase (%)	61.14 (1.67)	58.91(2.72)	62.71 (2.81)	65.31(9.04)*	62.59 (1.86)	61.53 (2.50)*
Kinematical parameters						
Pelvic tilt	9.66 (1.56)	6.83 (0.74)	14.93 (3.92)	17.09 (1.42)	10.79 (3.38)	13.37 (0.71)
Pelvic obliquity	0.65 (5.68)	−4.64 (8.73)	0.08 (2.61)	−1.62 (2.52)	−0.1(1.74)	−2.08 (1.26)
Hip flexion	15.89 (5.39)	7.78 (1.20)	32.68 (5.56)	33.13 (1.48)*	14.6 (5.89)	16.96 (0.74)*
Hip extension	23.15 (5.39)	13.20 (1.45)	33.18 (12.45)	37.19 (19.60)*	26.72 (6.87)	30.74 (9.80)*
Hip abduction	5.51 (2.83)	−3.87 (0.98)	−0.35 (4.18)	−4.1 (1.95)	−1.74 (3.51)	−7.18 (4.70)
Hip adduction	1.75 (2.91)	4.72 (0.82)	5.16 (4.70)	5.24 (3.17)*	5.37 (3.51)	8.81 (1.58)
Knee flexion	30.33 (5.21)	19.69 (1.82)	28.14 (5.22)	26.36 (5.87)*	23.66 (5.56)	26.16 (2.94)*
Knee extension	29.01(5.44)	17.53 (1.98)	47.61 (14.68)	51.51(24.04)*	48.44 (8.86)	51.66 (12.02)*

Left limb data for healthy and sound limb (=left limb) data for amputees were used to compute time–distance and kinematical parameters. For pelvic obliquity, negative value means pelvic drop. For hip abduction, negative value means hip was adducted.

* P values > 0.05 revealed no significant differences between healthy and amputee group through statistical analysis.

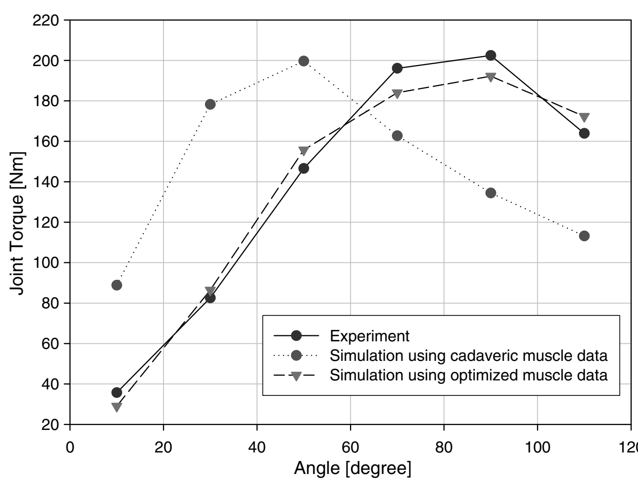


Fig. 2. Knee joint torque compared among experiment, simulation using cadaveric muscle data, and simulation using optimized muscle data.

3.3. Dynamic simulation using musculoskeletal models

3.3.1. Results for level walking

The pattern of muscle forces and moments of hip flexor and extensor in the sagittal plane of the amputees were shown to be quite similar to the pattern in healthy subjects even though the magnitudes are smaller. In the transverse plane, the muscle forces and moments showed similar patterns for both groups during the swing phase of the internal–external rotation. However, the pattern of the muscle force of amputees was shifted back about 10–20% from that of healthy subjects during the stance phase. Unlike the results in the transverse plane, the results in the coronal plane showed noticeable differences in magnitude and pattern. And there were big differences between the two groups in the forces and moments of the adductor during the total gait cycle and of the abductor during the stance phase (Fig. 3). Comparing the sum of abductor and adductor

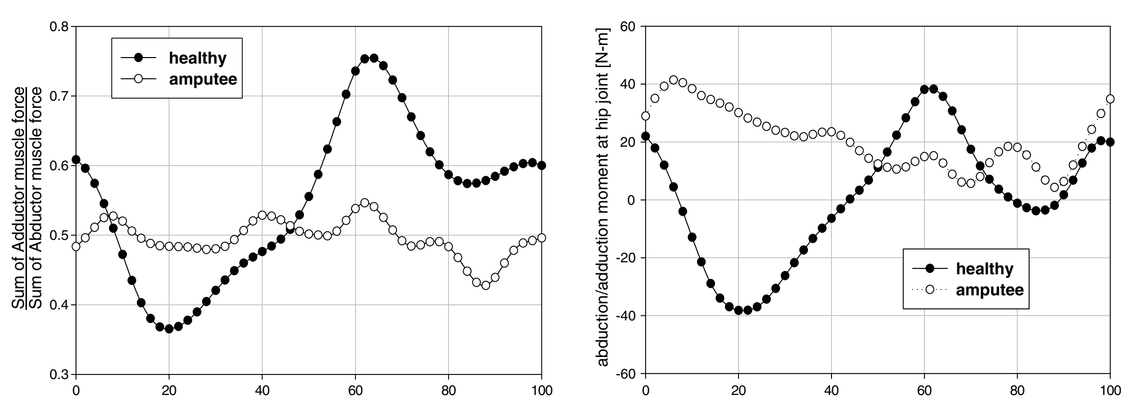


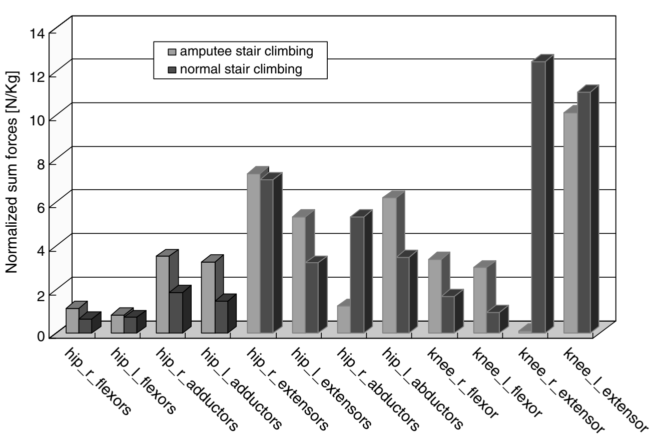
Fig. 3. Ratio of summed muscle forces (left) and moments (right) for hip abductors.

muscle forces for both healthy and amputee gait, the adductor in the amputee gait was less than the healthy gait at the swing phase and the abductor in the amputee gait was also less than that of the healthy gait at the stance phase. Then we calculated the ratio of the sum of the adductor muscle force to that of the abductor muscle force. The weakened abductor during the stance phase and the weakened adductor during the swing phase were observed in contrast to healthy subjects.

3.3.2. Results for stair climbing

We calculated the muscle forces of the hip joint for the frontal and sagittal plane during stair climbing (Fig. 4). In the sagittal plane, there was no difference between the healthy and amputee subjects for hip flexors, but the excessive muscle force of the hip extensors at the hip joint were needed to climb stairs because the summed muscle force of the hip extensors for the sound limbs of the amputees was 1.6 times (5.33 N/kg) greater than that of the healthy subjects. And, in the frontal plane, since the summed muscle force of the hip abductor for the amputated legs was four times (1.25 N/kg) less than that of the healthy legs, excessive hip adduction for both groups was generated. For the knee joint in the sagittal plane, the flexor muscle force of the amputated leg was twice (3.39 N/kg) as great as that of the healthy leg but the extensor muscle force completely depended on the hip extensor and abductor. For the sound limbs for the amputees, 10 times more extensor force (10.11 N/kg) was observed than that of the amputated legs. Also, we found the same results for joint torques (Fig. 4).

In the sagittal plane, the mean net joint moment profiles and magnitudes were very different at the hip and knee joints (Fig. 5). The hip flexion moment of the amputees (1.044 N m/kg) exceeded by seven times that of the healthy subjects during the stance phase, and the mean peak value was found at 16% of the gait cycle. And the peak flexion and extension power, observed at the transition from the stance phase to the swing phase at the hip, were 1.32 W/kg (22% of the gait cycle) and 1.29 W/kg (80% of the gait cycle), which exceeded those of healthy subjects. In contrast to the sagittal plane, the moments and power produced by hip adductors were increased rapidly during the swing phase (70% ∼100% of the gait cycle) in the frontal plane. At the knee, the first peak flexion moment was observed in the sagittal plane at 16% of the gait cycle (0.4 N m/kg) and the second peak flexion moment in the middle of the swing phase (80% of the gait cycle, 0.66 N m/kg). Considerable power was generated by knee flexors in the swing phase similar to the knee moment.



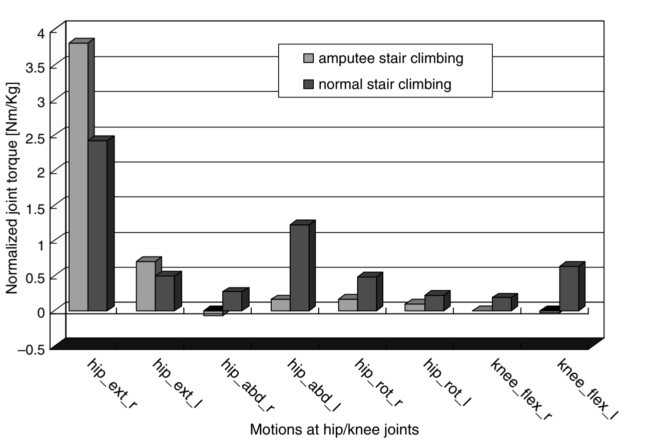


Fig. 4. Results of dynamic analysis using musculoskeletal models during stair climbing. The affix (_r, _l) indicated the right and left leg for the healthy but amputated and sound leg for amputees. The abbreviated affix (_ext, _flex, _abd, _rot) meant extension, flexion, abduction and rotation.

3.4. Muscle activity and coactivity

During level walking, the muscle activities of the major muscles for the sound limbs of the amputees were 20.45% for quadriceps and 87.87% for hamstring, lower than for healthy persons. But the muscle activities of tibialis anterior and gastrocnemius were 14.5% and 15.57%, respectively, greater than those of healthy subjects. The hamstring coactivity (BF/VL) had threefold greater coactivity, but the tibialis anterior coactivity (TA/GA) had 1.39-fold lower coactivity than the healthy subjects ($P < 0.05$). During ascent, all of the muscle activities were greater, excluding the hamstring. Contrary to ascent, there was no exception in stair descent and the magnitude of muscle activities was about 14–207% greater than the healthy group. Especially, quadriceps muscle activity was 1.64-fold greater, and hamstring muscle activity was 1.72-fold greater during descent, compared to those of the healthy group. For muscle coactivity during ascent, there was no significant difference in hamstring coactivity between the two groups ($P > 0.05$), but the tibialis anterior coactivity of the amputees was 2.61-fold greater than the healthy group. During descent, the hamstring coactivity of the healthy group had 1.27-fold greater coactivity, but the tibialis anterior coactivity of the amputees had 3.42-fold greater coactivity than the healthy group (Table 3).

For two-stairs ascent, the result on muscle activity showed that SA2 had also 4.31-fold greater activity on the average than the LW task. For the muscle activity of the rectus femoris, the major extensor at the knee, SA2

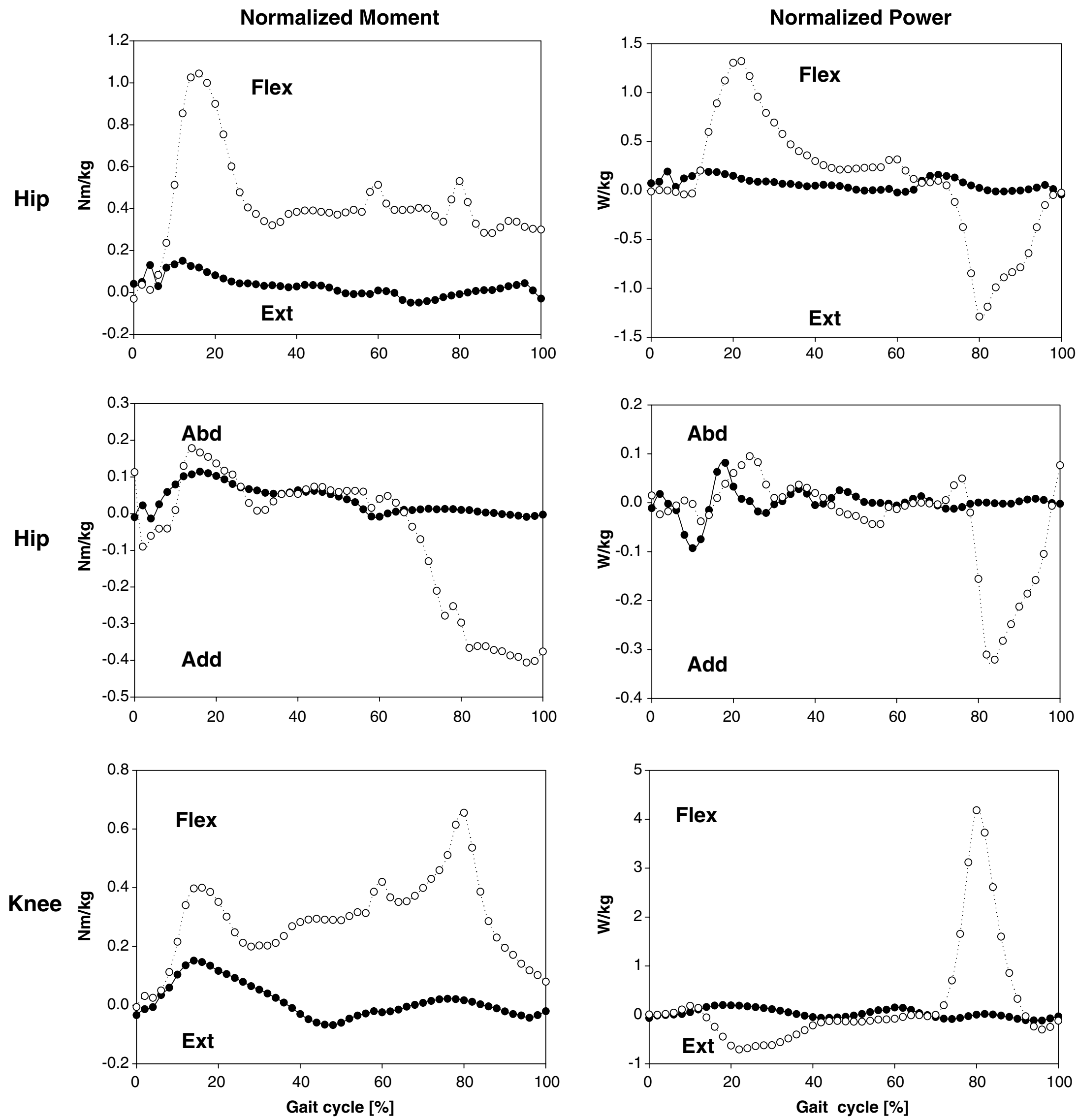


Fig. 5. Average joint moments and power at the hip/knee joint by dynamic analysis using musculoskeletal models during stair climbing for the sagittal and frontal plane of movement during stair climbing for normal subject (black circle) and amputee (blank circle, sound leg).

Table 3
EMG coactivity data for healthy and amputees during level walking and stair climbing

	Level walking		Stair ascent		Stair descent	
	Healthy ($n = 20$)	Amputee	Healthy ($n = 20$)	Amputee	Healthy ($n = 20$)	Amputee
Muscle activity (uv)						
Quadriceps	19.77 (1.72)	16.41 (0.99)	11.98 (1.83)	56.02 (4.14)	19.24 (4.30)	31.53 (4.47)
Hamstring	44.45 (5.31)	23.66 (1.74)	21.62 (4.50)	14.03 (2.86)	10.97 (1.56)	18.86 (3.44)
Gastrocnemius	61.50 (18.11)	71.93 (8.62)	21.62 (4.50)	32.47 (6.21)	10.97 (1.56)	33.70 (5.87)
Tibialis anterior	48.10 (9.50)	56.97 (16.91)	60.60 (10.17)	95.83 (33.92)	49.91 (8.87)	56.97 (16.91)
Muscle coactivity (unitless)						
BF/VL	45.48 (28.16)	160.16 (97.03)	99.49 (30.06)	97.59 (36.60)*	116.84 (11.71)	92.19 (49.46)
TA/GA	429.58 (413.53)	309.94 (275.95)	130.15 (56.10)	339.39 (311.88)	68.79 (40.30)	234.97 (133.58)
Peak muscle activity (uv)						
Vastus lateralis	50.6 (2.51)	105.13 (4.47)		Gastronemius	49.4 (2.51)	75.37 (10.88)
Biceps femoris	26.8 (4.44)	64.03 (4.99)		Tibialis anterior	48.1 (4.23)	94.33 (10.43)

Left limb data for healthy and sound limb (=left limb) data for amputees were used to compute EMG coactivity data.
* P values > 0.05 revealed no significant differences between healthy and amputee group through statistical analysis.

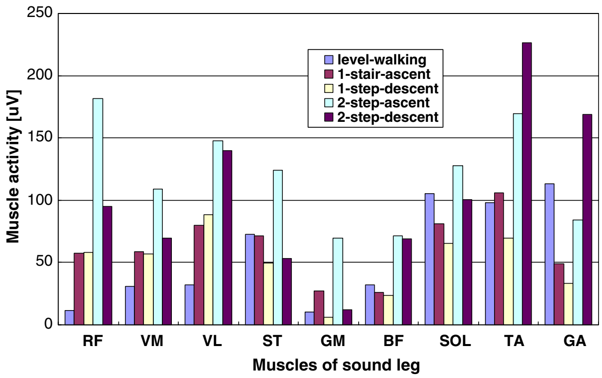


Fig. 6. Comparison of muscle activity for nine major muscles of sound leg of amputee among level walking, one step and two steps stair climbing. Muscle activity was the average value of peak RMS EMG during each motion. (RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; ST, semitendinosus; GM, gluteus maximus; BF, biceps femoris; SOL, soleus; TA, tibialis anterior; GA, gastrocnemius lateralis).

was 3.17-fold greater than SA1 and 16-fold greater than LW. For muscle activity of the gluteus maximus, the major extensor at the hip, SA2 was 2.71-fold greater than SA1 and 6.97-fold greater than LW (Fig. 6). For two-stairs descent, SD2 was 2.65-fold greater than LW and the difference in muscle activity was lower compared to the muscle activity during stair ascent. The muscle activity of the rectus femoris during SD2 was 8.39-fold greater than LW and was 1.63-fold greater than SD1.

4. Discussion

Unlike healthy persons, it is so difficult to evaluate the dynamic analysis for the amputated leg of above-knee amputees in view of the musculoskeletal system. There have also been several studies on muscle activation of the amputated leg by attaching surface electrodes in the prosthetic socket (Isakov et al., 2000; Jaegers et al., 1996). But it is also not easy since the electrode sensor size and connecting wire interfered with close adhesion between the amputated leg and the prosthetic socket. Though a modified socket that enables sensors to stick to the amputated leg was made, it is expected that the adaptation for the modified socket affected gait pattern (Esquenazi, 2004; Rietman et al., 2002; Czerniecki, 1996). Therefore, this study was performed to evaluate not only the sound limb but also the prosthetic limb of amputees by analyzing EMG activity and dynamic simulation using musculoskeletal models during level walking and stair climbing.

4.1. Gait parameters

For amputees, most of the time–distance parameters in both tasks were statistically different from the healthy subjects, but a percentage of the stance phase was not ($P >$ 0.05, Table 2). During stair ascent, the healthy climbed one stair per foot, but amputees have a gait pattern in which the foot of the amputated leg was pulled following the foot of the sound leg climbing the stair first. Therefore, though the magnitude of stride length and gait speed was lower compared to those of the healthy, there was no significant difference in the percentage of the stance phase ($P > 0.05$). It is well-known that the pelvis drops toward the opposite side during the stance phase, and excessive adduction is observed during the swing phase due to abductor weakness (Gitter et al., 2002). Our results for kinematic parameters also showed the typical characteristics of amputee gait so that pelvic obliquity, hip abduction, and hip adduction were statistically different from the healthy subjects.

4.2. Muscle activity and coactivity for level walking and stair climbing

From the results for muscle activity during level walking, we found that the muscle activity of the hamstring and quadriceps for the sound limbs of amputees had lower muscle activity, respectively, compared to those of healthy subjects, but instead that of gastrocneminus and tibialis anterior for amputees were statistically greater. It is thought that excessive power at the ankle joint of the sound limbs of amputees was needed to compensate for that of the prosthetic limb, unlike the healthy subjects. But for muscle coactivity during level walking, BF/VL had a statistically greater percentage of BF/VL than the healthy subjects. This may be caused by impaired quadriceps activation, similar to osteoarthritis patients and aged persons (Hurley, 2003; Hortobagyi et al., 2005). From the results for muscle activity during ascent, the muscle activity of the quadriceps and hamstring for the sound limbs of amputees were both greater than those of the healthy subjects. It is thought that the stair ascent task needs more flexion moment to overcome the flexion moment caused by the mass of the prosthetic limb, and the stair descent task needs to guarantee gait stability, respectively. Similar to this, excessive efforts by the sound leg were needed to compensate for the functional absence at the prosthetic limb.

4.3. Muscle activity and coactivity for two-stairs ascent and descent

From these results for amputees' habitual gait, it is speculated that SA2 and SD2 tasks cause greater load than LW, SA1, and SD1 tasks at the hip and knee joints. The reason for these results is that the sound leg has to support the body mass until the prosthetic limb seeks the stable position of the stair just below. It is also possible that the sound limb experiences excessive load to reduce the impact load when the prosthetic limb contacts the stair since the prosthetic limb does not secure gait stability during the stance phase (Simpson and Kanter, 1997). The results of SA2 and SD2 showed that the gait patterns of amputees to keep pace with climbing speed of the healthy cause unreasonable load to the hip and knee joints, and amputees using prostheses might be easily exposed to several musculoskeletal diseases such as osteoarthritis (Baker et al., 2004; Herzog et al., 2003; Hurley, 1997; Lemaire and Fisher, 1994).

4.4. Dynamic simulation using musculoskeletal models

The musculoskeletal model for healthy and amputee gait was built-up and simulated dynamically. The results of the simulation showed that the weakened abductor results in excessive adduction in the frontal plane; then inadequate valgus torque at the hip joint was generated. For that, previous studies reported the same results for above-knee amputee gait (Tesio et al., 1998; Jaegers et al., 1996; Burger et al., 1996). It is thought that impaired muscle force leads to unstable gait for amputees. From results of moment and power, we realized that the amputee felt so unstable during the swing phase of the sound leg that then excessive flexion moment and power were needed since the amputated leg could not support the sound leg during the swing phase of the sound leg. And from the result of dynamic simulation, the hip abductor needed more muscle force to abduct the femur than the healthy during level walking.

We can divide above-knee amputees wearing prostheses into two groups according to gait tendency in the stair climbing task. One group reduces the walking speed to increase gait stability and to relieve a fear of falling down, similar to aged persons and young children, and the other group tends to climb one or two stairs at once to keep up with the gait speed of healthy people, as mentioned previously. During stair ascent, the excessive moment and power were observed at the hip and knee joints of the sound limbs of amputees from the result of the simulation using musculoskeletal models. That is the reason the muscle forces at the hip and knee joints of the amputated leg were weakened by the dissected parts of the hamstring and quadriceps and those of the sound leg were needed to compensate for the insufficient muscle forces of the amputated leg. In contrast to the first group, it is clear that the gait tendency of the second group will expose amputees to several joint diseases such as inflammation and pain of the joints. Therefore, to improve gait stability and to reduce the risk of joint diseases in stair climbing, it is very important for amputees to strengthen the extensor and abductor muscle forces at the hip and extensor muscle forces at the knee of the amputated leg. Keeping up with the build-up of muscle forces, it is necessary to develop new prostheses that can control the knee extension moment of the amputated leg and prevent knee flexion generated by movement of body weight at the moment from the swing phase to the stance phase.

4.5. Limitations of recruitment and musculoskeletal model

Most of all, we wanted to recruit younger trans-femoral amputees as volunteers rather than older amputees at the beginning of experiments to compare the results of healthy group, but failed to recruit them. Therefore the control group appears to be younger, taller and lighter than the above-knee amputee group. In general, the study for trans-femoral amputees has frequently actual difficulty in recruiting volunteers. For that, it is considered that age-related researches for amputee gait were needed.

For simulations using musculoskeletal models, there are several limitations in these studies. First of all, the condition of the dissected limbs of amputees was not even and could not be analyzed statistically. Although medical histories on the amputees existed, detailed description about the amputation surgery was absent. Therefore, it was too difficult to make an exact musculoskeletal model for the amputated leg. Second, it is not clear whether the muscle parameters that were used to analyze the major muscles were accurate and realistic since most of the parameters referred to cadaver studies (Brand et al., 1986; Delp et al., 1990). Although the limitation of muscle parameters was diminished through static optimization, it is still expected that the calculated muscle force and moment were not equal to true muscle force and moments. This is always limiting in simulations using musculoskeletal models. Therefore, validation using RMS EMG for simulation was always needed. Consequently, we believed that more accurate muscle parameters will enable us to simulate various motions for above-knee amputees in view of the musculoskeletal system.

5. Conclusions

Using musculoskeletal models for above-knee amputees and the healthy, dynamic analyses were performed on level walking and stair climbing tasks. According to gait analysis, most of the kinematic parameters showed statistically no difference among the tasks, excluding pelvic tilt, pelvic obliquity, and hip abduction. The activities of major muscles and coactivities of the hamstring and tibialis anterior for the sound limbs of the amputees showed that the stair ascent task needed more muscle activity than the stair descent task and level walking and that muscle activity and coactivity were greater than those of the healthy, excluding the hamstring coactivity during the stair ascent. In addition, it was possible that the two-stairs ascent/descent tasks might cause unreasonable load to the hip and knee joints of above-knee amputees. Lastly, dynamic analysis showed that weakened abductor, excessive adductor, and inadequate torque at the hip joint of the amputees' prosthetic limbs during all tasks were quantitatively observed and high power and moment at the knee joint of the amputees' sound limbs were found to compensate for the lack of prosthetic limbs.

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