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A Musculoskeletal Model for Biomechanical Analysis of Transfemoral Amputees Climbing Stairs

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Understanding the characteristics of amputee gait is key in developing more advanced prostheses. The aim of this study was to quantitatively analyze a stair-climbing task for transfemoral amputees with a prosthesis and to predict the muscle forces and joint moments at musculoskeletal joints using a dynamic analysis. A three-dimensional musculoskeletal model of the lower extremities was constructed from a gait analysis using transformation software for two transfemoral amputees and ten healthy people. The measured ground reaction forces and kinematical data of each joint from the gait analysis were used as input data for an inverse dynamic analysis. Dynamic analyses of an transfemoral amputee climbing stairs were performed using musculoskeletal models. The results showed that the summed muscle forces of the hip extensor of an amputated leg were greater than those of a sound leg. The opposite was true at the hip abductor and knee flexor of an amputated leg. We also found that higher moments at the hip and knee joints of the sound leg were required to overcome the flexion moment caused by the body weight and amputated leg. Dynamic analyses using musculoskeletal models may be a useful means to predict muscle forces and joint moments for specific motion tasks related to rehabilitation therapy.

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1. Introduction

More transfemoral amputees want to use ambulatory prostheses rather than assistive devices such as wheelchairs and crutches. Various mathematical models and ambulation analyses for level walking have been developed from mechanical and kinematical points of view. The results of these studies have had an enormous influence on the development of new prostheses and artificial feet for transfemoral amputees.[1,2] However, compared to level walking, studies on stair climbing are relatively scarce.

To understand the details of walking, we must determine the muscle action for each gait cycle. Normal people generally must use extensors of the hip and knee joints to lower their body when climbing down a stair. When climbing up a stair, the hip and knee joints must continuously extend and generate a torque for their extensors to overcome the flexion moment generated by the body weight.[3,4]

Unlike normal people, most transfemoral amputees do not bend the leg with a prosthesis to attain stability when climbing stairs, but instead depend on the surrounding environment. This gait represents a situation in which the leg with a prosthesis cannot overcome the bending torque generated by the body weight while the sound side is in the swing phase because the extensor of the hip joint has been cut by the transfemoral amputation. The same situation occurs when walking down stairs. However, most transfemoral amputees with a prosthesis often climb two or three steps at once to climb faster, or lightly lift the prosthesis and then leap using their sound side to climb down faster. Unlike level walking, this type of gait applies excessive loads to the hip and knee joints of the sound side, and may cause joint disorders.

To more accurately analyze the stair-climbing gait of amputees, we created a musculoskeletal model for prosthetic ambulation, and analyzed and evaluated the loads on major joints and the contribution of the muscles using inverse dynamics.

2. Materials and Methods

2.1 Experimental Subjects

We examined two male subjects with an amputation right above the knee (transfemoral amputees) who had received rehabilitation training for at least 1 year and walked with prostheses for at least 6 months. The subjects did not have a vascular disease or skin troubles.

2.2 Musculoskeletal Modeling for the Amputated Stump

A computed tomography was conducted to obtain the length of the amputated femur. The photographic image was reorganized into a three-dimensional representation using image-processing software (Vworks version 4.0; CyberMed, Inc., USA) to calculate the length of the amputated femur. The state of the remaining muscles in the amputated leg was checked using a MRI and recreated based on the advice of clinical doctors assuming typical surgery techniques (see Fig. 1).

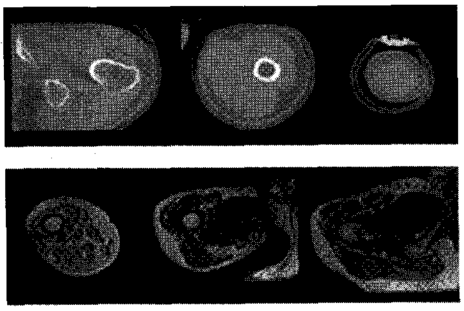


Fig. 1 CT image (upper) and MRI (lower) of a stump

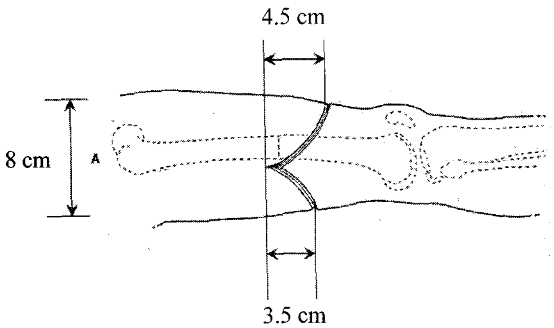
The musculoskeletal modeling of the stump assumed a typical surgery technique that configures a fish-mouth flap around the transfemoral amputated part. The amputated part was divided into front and back sections, both inside and outside, based on the MRI and the advice of clinical doctors, to redefine the insertion points of the amputated muscles (see Fig. 2).

2.3 Prosthesis Model

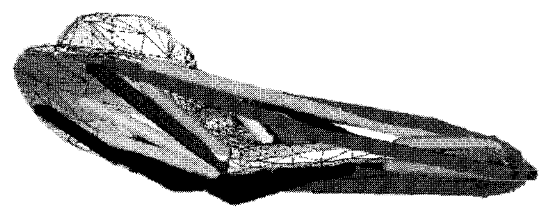
To analyze stair-climbing gaits, we modeled the prosthesis actually worn by each subject. The prosthesis consisted of 13 major components and four joints, and each component was resized to fit the amputated stump of the subject. We used the mechanical functions of each joint based on the movement of the knee joint to implement the mechanical movement of a four-axis link, and modeled the movement of the actual prostheses (see Fig. 3).

2.4 Muscle Model

The muscle model was set up using a revised Hill model that treated each unit of muscle and tendon as serially interconnected contractile elements.[5-7] We required the following five variables to measure the force applied to each muscle: the maximum muscle force derived from the physiological cross-sectional area (PCSA) of the muscle, the pennation angle between the muscle and tendon, the tendon length when the tendon is lengthened and a force transmitted, and the speed at the maximum muscle contraction. The values of these variables for each muscle were determined from existing studies.[8]



(a) Fish-mouth flap used during amputation surgery



(b) Reconstructed musculoskeletal model

Fig. 2 Amputation through the middle third of the thigh

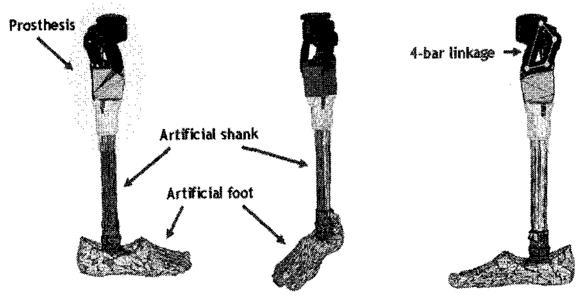


Fig. 3 Prosthesis model with a 4-bar linkage mechanism

We used the muscle force–muscle length, muscle force–speed, and the parallel passive elements of the muscle force–muscle length relational curves to evaluate the force represented by the contractile elements. These curves were normalized by the maximum muscle force and the corresponding muscle tissue length. Ligaments were excluded from this study, which only considered macro-dynamic effects because they only act as passive elements. The damping ratio was set to 0.15 in consideration of the viscoelasticity of the muscles. The pennation angle, muscle activation level, length of the muscle fiber, and speed of the muscle fiber were used as state variables.[9]

2.5 Optimization of the Muscle Model

To redefine the musculoskeletal variables around the amputated part, the amputees exerted the maximum muscle force by assuming the largest muscle mechanism of action. The experimentally measured torque of the hip joint on the amputated side and the sound side were compared to redefine the kinetic variables for the muscles around the amputated part.[10]

2.6 Dynamic Analysis Using the Musculoskeletal Model

We attempted to create a lower-limb musculoskeletal model of the subjects based on the amputated part and prosthesis and using a muscle model. Before creating a musculoskeletal model, we first tuned the skeleton and muscle components for each subject. We measured the height and weight of each subject and the lengths of their femur, leg, and lower limb. We also determined the mass, center of mass, and moment of inertia, which are the kinetic parameters for each body part necessary for a kinetic analysis. These values were calculated by referring to existing studies.[11]

The kinetic parameters of the prosthesis socket, lower limb, and artificial foot worn by each subject were modeled using a design program (Solidworks 2006; Solidworks Corp., USA). These were determined using a kinetic program (ADAMS 2003; MSC, USA). We assigned a range of restraints to prevent excessive movements of each joint by referring to the actual range of motion of a human body, and applied 6 degrees of freedom to each joint (see Fig. 4).

We included adductor and abductor muscles around the hip joint and all muscles responsible for femoral bending and extension. For the amputated knee joint, the muscles were modeled in consideration of the redefined muscles from the orthopedic suture. A governing equation consisting of Hill functions was applied to each muscle,[5,6,12] and a linear differential equation was used to model the activation mechanism of the muscles. The musculoskeletal program SIMM (version 4.01; Musculographics, Inc., USA) provided the mathematical background and visual aspects for clinical physicians.[13]

We attached 15 reflective markers (Helen Hayes Marker Set) to the lower limb of each subject before performing the gait analysis. Then we used a conversion program (Motion Module; Motion Analysis Inc., USA) to obtain the musculoskeletal model for each subject. For stair climbing, we used a wooden stair and installed forceplates (400 × 600 mm; Kistler Instrument Corp., USA) on each step to measure the ground reaction force.[14]

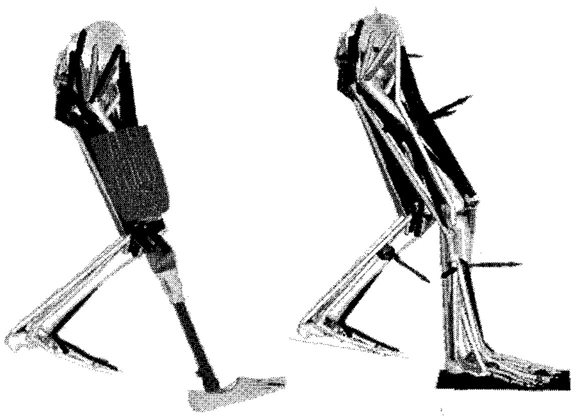


Fig. 4 Musculoskeletal model for an amputee (left) and a healthy subject (right)

We inputted the kinetic variables for each part of the subject and adjusted the range of motion appropriately for each joint. We then derived the equation of motion using SD/FAST (B2.8; Symbolic Dynamics, USA). After inputting the mechanical trace of each joint measured through the gait analysis and the ground reaction force measured by the forceplates, we performed an inverse dynamic analysis using Dynamic Pipeline (version 3.0; Musculo-Graphics, Inc., USA) to analyze the muscle skeleton of a normal person and an amputee while climbing stairs.

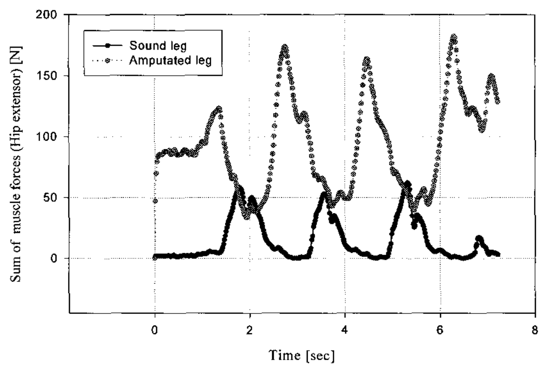


Fig. 5 Summed muscle forces of the hip extensors

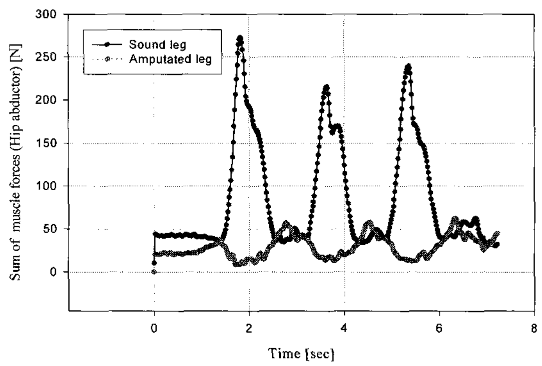


Fig. 6 Summed muscle forces of the hip abductors

3. Results

3.1 Muscle Force

Although the maximum muscle force of the hip joint extensors is normally balanced between the left and right sides of the body, the value on the prosthesis side was three times higher than that on the sound side when an transfemoral amputee climbed a stair, and was about 25% lower than that of a normal person's lower limb (see Fig. 5). The muscle force of the hip joint abductor on the amputated side was at least four times lower than that of the sound side, and the latter was about 10% lower than that of a normal person (see Fig. 6).

We also calculated the muscle forces of the hip and knee joint extensors when climbing stairs. While the muscle forces of the left and right knee extensors of a normal person varied by 11%, the muscle force of the knee joint extensor of the amputated side was 10 times lower as that of the sound side. The muscle force of the sound side extensor of the amputee was 8% greater than that of a normal person (see Fig. 7).

The muscle force of the knee joint extensors (quadriceps) was three times larger than that of the hip joint abductor. However, a comparison with the muscle force of the hamstring of the knee joint did not indicate a large difference. The sound side exerted more muscle force, although the time when the maximum muscle force was exerted was different.

3.2 Joint Moment

We calculated the extension moment required to overcome the flexion moment in order to climb a stair. The moment of the hip joint of the amputated side was at least twice greater than that of the sound side, which indicates that the load on the hip joint of the sound side was much larger than that of the amputated side. The flexion moment of an amputee was as much as five times greater than normal during the stance phase (see Fig. 8). We also compared the abduction moment of the hip joint with normal values. Little difference was observed during the stance phase, but the abduction moment during the swing phase was at least four times lower than normal, indicating an excessive adduction moment rather than an abduction moment. These results are similar to those obtained for level walking by amputees (see Fig. 9).

The moment of the knee joint of the sound side was as much as five times greater than that of the amputated side. However, compared to normal values, the knee flexion moment was twice lower during the stance phase and six times lower during the swing phase.

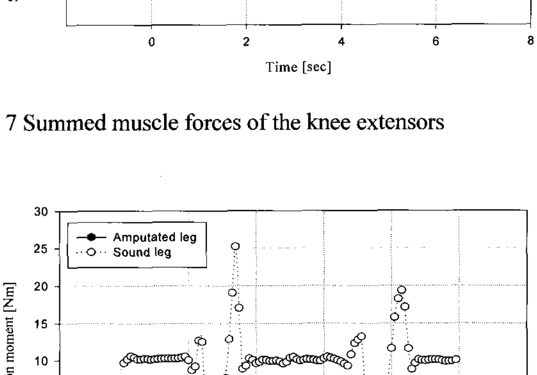


Fig. 7 Summed muscle forces of the knee extensors

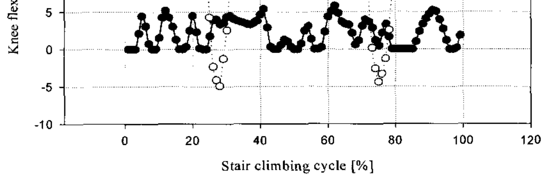


Fig. 8 Flexion moment at the hip joint

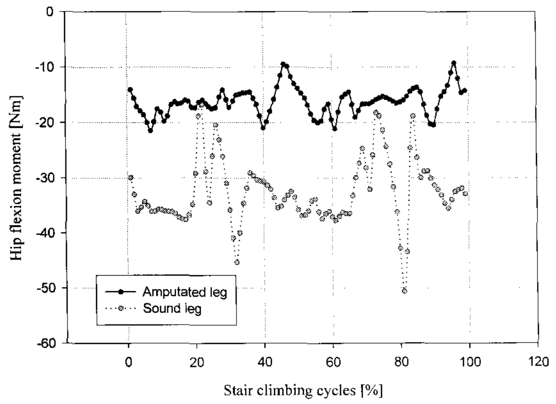


Fig. 9 Flexion moment at the knee joint

4. Discussion and Conclusions

We created a three-dimensional musculoskeletal model to examine the stair-climbing gait of amputees. We performed inverse dynamic analyses using the mechanical and kinetic data obtained from actual gait analyses, and obtained the muscle forces and moments of each joint of amputees with prostheses while climbing stairs.

The muscle extensors on the sound side of the amputees required higher muscle forces than those on the amputated side. The results were the opposite for the abductors. This agrees with the results of prior studies, which found that the abductor of the hip joint weakened after amputation. We calculated the muscle force of the knee joint extensor and determined that the force on sound side was 10 times larger than that on the amputated side. This was observed in the actual walking habits of patients wearing prostheses, and suggests the possibility of future joint disorders in the sound side of patients as they transfer considerable load away from their amputated side.

We also calculated the moment of the hip and knee joints of amputees while climbing stairs. An extension moment occurred on both the amputated and sound sides to overcome the flexion moment due to the body weight, but the extension moment on the sound side was twice larger than that on the amputated side.

During stair climbing, an insufficient extension moment exists on the amputated side to overcome the flexion moment due to the body weight so that the amputee cannot achieve gait stability. We saw little variation in the size of the moment for the knee joint because the sound side adjusted the bending moment to an appropriate level while the amputated side could not during the entire gait cycle.

Therefore, to achieve gait stability during stair climbing by transfemoral amputees, reinforcing the muscle force of the hip joint extensors of the amputated stump is necessary. We also must develop a prosthesis that does not bend due to the moment caused by the body weight when amputees climb stairs, and adjusts the extension moment of the knee joint on the amputated side when the sound side is moving from the swing phase to the stance phase.

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