INDIVIDUAL MUSCLE FUNCTION IN BELOW-KNEE AMPUTEE WALKING

Anne K. Silverman and Richard R. Neptune

Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA
email: asilverman@mail.utexas.edu, web: http://www.me.utexas.edu/~neptune

INTRODUCTION

Below-knee amputees lose the functional use of the ankle plantar flexors, which are important for providing body support and forward propulsion in non-amputee walking [1]. In the absence of the ankle-plantar flexors, compensations from other muscle groups are needed to provide these necessary walking subtasks. Experimental studies have shown that a variety of compensatory mechanisms are used in amputee gait, such as increased net joint power and work from the residual hip extensors and the intact ankle plantar flexors [2]. Increased muscle activity has also been observed in the knee extensor and biarticular hamstring muscles [3]. However, the biomechanical function of these compensations is not well-understood. The purpose of this study was to use 3D musculoskeletal modeling and simulations of amputee walking to assess individual muscle contributions to body support and forward propulsion to understand how amputees compensate for the lost ankle plantar flexors.

METHODS

A 3D musculoskeletal model with 14 rigid body segments, 23 degrees of freedom, and 38 Hill-type musculotendon actuators per leg was developed in SIMM (MusculoGraphics, Inc.). Musculoskeletal geometry was based on that of Delp et al. [4]. The ankle muscles on the residual leg were removed and the mass and inertial properties of the residual leg were modified to represent a nominal below knee amputee [5]. The prosthesis stiffness was modeled as a second-order passive spring (Eq.1)

$$\tau = a_0 + a_1 \theta + a_2 \omega + a_3 \theta^2 + a_4 \theta \omega$$  \hspace{1cm} (1)

where $\theta$ is the ankle angle, $\omega$ is the ankle velocity, and $\tau$ is the ankle torque. Constants $a_0$–$a_4$ (Eq. 1) were determined using a multiple regression analysis of the experimental data (see below).

The excitation of each muscle was defined with a bimodal excitation pattern. A simulated annealing algorithm was used to identify the optimal excitation patterns to emulate the experimental body segment kinematics and ground reaction forces (GRFs). Each muscle had six optimization parameters including onset, offset and amplitude of the two modes. To provide data for the optimization algorithm, kinematic and kinetic data were collected from 14 below-knee amputees walking overground at $1.2 \pm 0.06$ m/s [2]. The kinematic and GRF data were then averaged across subjects to produce a nominal amputee walking pattern.

Muscle contributions to body support and forward propulsion were quantified by their contributions to the vertical and anterior/posterior (A/P) GRFs, respectively. Individual muscles were then arranged into muscle groups based on anatomical location and muscle function (e.g., three vasti muscles combined into one vasti group). Muscle contributions were analyzed during residual leg stance.

RESULTS AND DISCUSSION

The residual leg vasti, gluteus medius and gluteus maximus muscles were the largest contributors to body support in early and mid-stance (Fig. 1), which was consistent with previous non-amputee simulation studies [6, 7]. The vasti muscles have also been shown to have increased muscle activity in amputee walking [3]. The prosthesis contributed to support throughout stance (Fig. 1), which was consistent with a previous simulation study of 2D symmetric amputee walking showing contributions from the prosthesis to trunk support [8].
The residual leg hamstring muscles were the primary contributors to forward propulsion throughout residual leg stance (Fig. 1). This was consistent with previous non-amputee walking simulation studies that have shown the biarticular hamstrings to contribute positively to the A/P GRF throughout the stance phase [6]. In addition, experimental amputee studies have shown prolonged muscle activity of the hamstrings on the residual leg in early stance [3] and increased hip joint powers and work from the hip extensors to reduce the residual leg braking A/P GRF [2].

In early stance, the biceps femoris short head contributed positively to propulsion, while the rectus femoris, vasti muscles and gluteus medius contributed negatively to propulsion (Fig. 1). In mid-stance, the residual leg gluteus maximus contributed substantially to propulsion (Fig. 1), which has previously been shown in non-amputee walking studies [6]. The prosthesis contributed to propulsion in late stance (Fig. 1) as it returned elastic energy stored in early stance. Previous 2D simulation analyses showed that the prosthesis acts primarily to transfer energy from the leg to the trunk to provide trunk support and forward propulsion in late residual leg stance [8].

CONCLUSIONS

The prosthesis provided considerable propulsion and support in the amputee simulation, but compensations from other muscles were necessary to compensate for the lost ankle muscles. Contributions from the residual leg vasti, gluteus medius and gluteus maximus to support were needed. The amputee simulation also showed contributions of the residual leg hamstrings and gluteus maximus to forward propulsion. Thus, improving the output of these muscles may help improve rehabilitation outcomes and amputee mobility.

REFERENCES


ACKNOWLEDGEMENTS

This work was supported by National Science Foundation Graduate Research Fellowship and Grant No. 0346514.