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Motor adaptation to prosthetic cycling in people with trans-tibial amputation

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ABSTRACT

The neuromusculoskeletal system interacts with the external environment via end-segments, e.g. feet. A person with trans-tibial amputation (TTAmp) has lost a foot and ankle; hence the residuum with prosthesis becomes the new end-segment. We investigated changes in kinetics and muscle activity in TTAmps during cycling with this altered interface with the environment. Nine unilateral TTAmps and nine subjects without amputation (NoAmp) pedaled at a constant torque of 15 Nm and a constant cadence of 90 rpm (~150watts). Pedal forces and limb kinematics were used to calculate resultant joint moments. Electromyographic activity was recorded to determine its magnitude and timing. Biomechanical and EMG variables of the amputated limb were compared to those of the TTAmp sound limb and to the dominant limb in the NoAmp group using a one-way ANOVA. Results showed maximum angular displacement between the residuum and prosthesis was 4.8 ± 1.8 deg. The amputated limb compared to sound limb and NoAmp group produced lower extensor moments averaged over the cycle about the ankle (13 ± 2.3, 20 ± 5.7, and 19 ± 5.3 Nm, respectfully) and knee (8.4 ± 5.0, 15 ± 4.5, and 12.7 ± 5.9 Nm, respectfully) ($p < 0.05$). Gastrocnemius and rectus femoris peak activity in the TTAmps shifted to later in the crank cycle (by 36° and 75°, respectfully; $p < 0.05$). These data suggest gastrocnemius was utilized as a one-joint knee flexor in combination with rectus femoris for prosthetic socket control and highlight prosthetic control as an interaction between the residuum, prosthesis and external environment.

1. Introduction

The neuromusculoskeletal system interacts with the environment via end-segments, e.g. the foot, hand, etc. (Jacobs and van Ingen Schenau, 1992; Mussa-Ivaldi et al., 1985; Winter, 1995). A person with an acquired trans-tibial amputation (TTAmp) however, has lost the foot/ankle complex making the residuum the motor system's new end-segment. While a prosthesis is designed to replace the amputated limb, control of the prosthesis at the residuum/socket interface presents a significant challenge to the sensorimotor system in ultimately controlling the interaction between the prosthesis and the environment. The motor system controls environmental interactions in TTAmps through appropriate control of the residuum/socket interface.

The prosthesis is not directly connected to the skeletal system and potential movement can occur at the residuum/socket interface. Relative position of the residuum with respect to prosthetic socket has been shown to change in various static leg configurations (Erikson and Lemperg 1969; Newton et al., 1988; Lilja et al., 1993; Narita et al., 1997; Soderberg, 2003) using radiographic techniques and during gait (Sanders et al., 2006) using a photoelectric sensor. Childers et al. (2012) addressed this issue by modeling the residuum/prosthesis interface as a pinned *residuum-prosthesis pseudo joint* (RPP) as there seemed to be minimal translational yet potentially large rotational movement. The motor system should account for movement about the RPP joint for prosthetic control yet prior studies examining motor control strategies with amputation assumed there was no motion between the residuum and prosthesis (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers et al., 1998; Selles et al., 2003; Fey et al., 2010). Understanding how the human motor system adjusts to amputation and prosthetic use can provide insight into motor compensation mechanisms to injury and using assistive devices.

The cycling task provides a controlled environment in which rhythmic locomotion can be studied (Gregor and Childers, 2011 for review). Pedaling kinetics have been reported in TTAmp volunteers suggesting pedaling techniques are modified such that sound limb contribution increases (Childers et al., 2011a). This alteration

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is not solely due to strength and/or inertial differences between limbs suggesting there may be other reasons explaining the motor adaptation strategies utilized by TTAmp cyclists (Childers et al., 2011b).

Functionally appropriate changes in motor patterns in response to injury have been documented in the past and are afforded by musculoskeletal redundancy and nervous system plasticity. For example, denervation of select ankle extensors and/or knee flexors in the cat leads to activity changes in intact muscles (Maas et al., 2010; Pearson et al., 1999; Tachibana et al., 2006) that apparently preserve the pre-injury leg/ground interactions during locomotion: the leg length and orientation (Maas et al., 2007; Chang et al., 2009) and the ankle joint moment and power magnitudes (Prilutsky et al., 2011). After more extensive injuries, e.g., limb amputations, compensatory changes in motor output pattern might not be sufficient for full preservation of prosthetic limb interactions with the external environment, as evident from the asymmetric pedaling kinetics reported in TTAmps (Childers et al., 2011a). Documenting adaptive changes in muscle activity during interactions of a person with amputation with the external environment using a prosthesis will help in understanding the motor adaptations in amputees and in improving prosthetic designs.

The purpose of this study was to examine motor adaptations, i.e., changes in kinetics and muscle activity, in TTAmp versus persons without amputation (NoAmp) pedaling against a constant load and cadence. The specific hypothesis tested was that TTAmp subjects would alter muscle activation patterns in the prosthetic leg to perform this cycling task. In particular, the partially amputated GAS (ampGAS) would shift its activity burst to a later (knee flexion) phase in the pedaling cycle, as suggested in Childers et al., 2011a and inferred from animal muscle denervation studies (e.g., (Tachibana et al., 2006)).

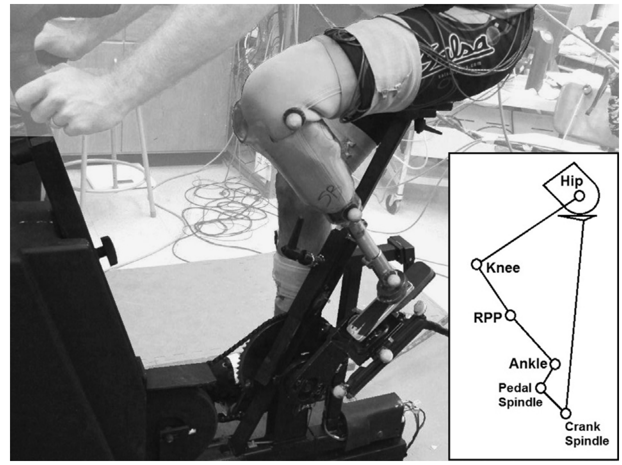

Fig. 1. A subject from the TTAmp group on ergometer with a marker placed on the prosthesis over the 'pseudo-joint' where the pin meets the lock. Removal of the lateral superior wall of the prosthesis socket facilitated placement of the knee marker. The prosthetic foot was a stiff plate of aluminum. Inset defines joint locations.

2. Methods

Nine persons with unilateral TTAmp and nine NoAmp volunteers were recruited (Table 1). All volunteers gave written informed consent approved by the Institutional Review Board at the Georgia Institute of Technology before participating. Volunteers in both groups used cycling for recreation. The NoAmp group was matched to the TTAmp group to ensure similar cycling experience, e.g., road or triathlon experience, self-reported hours of cycling per week, body mass, height, age, and sex. The TTAmp group inclusion criteria were: unilateral transtibial amputation secondary to trauma or cancer, at least one year cycling experience post-amputation, performing cardiovascular exercise >6 hrs per week, between 18–45 years old, and no secondary neuromuscular conditions. These criteria minimized cardiovascular risk during the experiment (ACSM, 2006).

The volunteers pedaled a stationary electromagnetically-braked ergometer (Excaliber Sport, Lode BV, Groningen, The Netherlands) adapted with dual piezoelectric element force pedals (Broker and Gregor 1990) and a commercial "clipless" pedal system (Wheeler et. al., 1992). The saddle height, handlebar reach, drop, and seat tube angle were adjusted to the subject's position as measured from their primary bicycle or (if their bicycle was unavailable) an established bicycle positioning protocol (Pruit, 2004). The cycling shoe (Bontrager Race Mountain, Trek bicycle corp., Madison, WI, USA) was sized to the subject's foot and the pedal interface (Shimano SPD MTN, Shimano inc., Osaka, Japan) was controlled across all subjects.

Table 1
Group characteristics (mean ± SD).

Measure	TTAmp Group	NoAmp Group
Number of participants	**9**	**9**
Cycling Experience (yrs)	**7.1 ± 9.8**	**9.2 ± 11.4**
Cycling time per week (hrs/wk)	**4.7 ± 2.0**	**6.1 ± 2.8**
Aerobic Exercise time per week (hrs/wk)	**14.8 ± 6.4**	**13 ± 5.8**
Body mass (kg)	**83.8 ± 14.9**	**82.4 ± 11.7**
Height (cm)	**183.0 ± 8.0**	**182.0 ± 5.0**
Age (yrs)	**34.1 ± 8.7**	**34.7 ± 8.8**
Time since amputation (yrs)	**12.9 ± 11.9**	**N/A**
Residuum length (cm)	**20.9 ± 3.8**	**N/A**

The prosthetic foot and geometric socket/pedal relationship were similar across all TTAmp subjects and described in prior research (see STIFF foot condition in Childers et al., 2011a). The prosthetic foot was a stiff 155 mm × 50 mm × 10 mm plate of 6061-T6 aluminum (Fig. 1). This foot was found to minimize work asymmetry and pedal stroke variability in a prior experiment (Childers et al., 2011a). Cleat anterior/posterior position was adjusted to match the sound limb using a prosthetic slide adapter. A thermoplastic socket was fabricated by a professional fabrication facility (PDI, Dayton OH, USA) for each TTAmp volunteer by duplicating the volunteer's personal socket with an electromagnetic shape capturing device (TracerCAD, Ohio Willow Wood co. inc., Columbus OH, USA). A portion of the lateral wall was removed allowing for placement of a knee center marker (Fig. 1) and incorporated a pocket in the posterior portion for EMG electrodes over the ampGAS. Prosthetic suspension included a silicon liner with mechanical pin type suspension (X-PSH-PLUS, PDI, Dayton OH, USA).

The volunteers were given a 5–10 minute warm up cycling period at 75 W and self-selected cadence. The volunteers then pedaled at ~150 W at a constant torque (15 Nm) and cadence (90 rpm) during data collection. Volunteers received cycling cadence feedback via an ergometer mounted tachometer. A heart rate monitor (CS400, Polar Electro OY, Kempele, Finland) was worn to verify the workload was submaximal as defined by ACSM (2006) intended to minimize any effects of fatigue. Data were collected for 30 seconds after two minutes of cycling.

Pedal reaction forces were recorded at 300 Hz and digitally filtered using a fourth-order zero-lag Butterworth filter with a 15 Hz cutoff frequency. Kinematic data were collected at 60 Hz (Peak Performance Technology Inc., Englewood, CO, USA) and digitized using Peak Performance software. An electronic pulse synchronized force, EMG and video records. Pedal angle was calculated based on pedal mounted reflective markers. Crank angle was determined using a gear driven continuous turn potentiometer. Nine markers were used to define limb segments. Marker locations include the volunteer's sacrum as well as bilaterally the greater trochanter, anterior superior iliac spine (ASIS), lateral epicondyle of the femur, and lateral malleolus. The TTAmp group had an additional marker placed at the residual limb-prosthesis joint (Fig. 1). Translational movement in this region is less than 4 mm, which allows for treating this marker as a joint center location and for calculating angular displacements between the residual limb and prosthesis (Childers et al., 2012). Marker coordinate data were smoothed using a quintic spline in Peak Motus before exporting to Matlab (MathWorks, Inc., Natick, MA, USA). The kinetic and kinematic data from each of eight complete crank cycles were time normalized to 100 data points and averaged together. The ankle joint center of rotation was calculated based on equations from Vaughan et al. (1999) and knee instantaneous center of rotation was calculated based on work of Smidt (1973). A static calibration trial, placing markers over the greater trochanter, sacrum and ASIS, was used to establish the relationship between these body markers (Neptune and Hull, 1995).

Limb segment center of mass, mass, and moment of inertia were calculated from regression equations (Zatsiorsky et al., 1990). Residual limb and prosthesis inertial properties were calculated using methods described by Goldberg et al. (2008) and (Smith and Martin, 2013). Briefly, these calculations involved modeling the residuum as a frustum based on anatomical measurements and assuming

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