Published in final edited form as:

J Biomech. 2014 July 18; 47(10): 2306–2313. doi:10.1016/j.jbiomech.2014.04.037.

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 15Nm and a constant cadence of 90rpm (~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 $4.8 \pm 1.8^\circ$. 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.

Keywords

Bicycling; Motor Control; Below Knee Amputation; Sport Biomechanics; Prosthetics

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Conflict of Interest Disclosure: The authors report no conflict of interest.

Introduction

The neuromusculoskeletal system interacts with the environment via end-segments, e.g. the foot, hand, etc. (Jacobs & 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 & Lemperg 1969; Newton et al., 1988; Lilja et al., 1993; Narita et al., 1997; Soderberg & Roentgen, 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 & Sienko, 1988; Sanderson & Martin, 1997; Powers et al., 1998; Selles et al., 2004; 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 & 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 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)).

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 >6hrs 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 & 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.

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 155mm X 50mm X 10mm plate of 6061-T6 aluminum (Figure 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 (Figure 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 watts and self-selected cadence. The volunteers then pedaled at ~150 watts at a constant torque (15Nm) 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 (Figure 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 & 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 & Martin, 2013). Briefly, these calculations involved modeling the residuum as a frustum based on anatomical measurements and assuming 1.01 kg/cm³ for tissue density (Goldberg et al., 2008). Prosthesis rotational inertia parameters were determined using the pendulum technique while mass and center of mass location were determined using a calibrated scale (Smith & Martin, 2013).

Joint moments were calculated using the Newton-Euler equations of rigid body dynamics for the sagittal plane with the addition of the residuum/prosthesis pseudo-joint (Broker & Gregor, 1994). An uncertainty and sensitivity analysis showed the inverse dynamics calculations were most sensitive to joint center calculation, but the uncertainty was relatively small (about ± 4 Nm or 5-8% of peak extension joint moments during cycling, 95%

confidence interval; see Appendix C in Childers, 2011). Pedaling work asymmetries were calculated as the difference in contribution of each limb to net crank torque (Sanderson, 1990).

Surface electromyographic (EMG) bipolar electrodes and a 12-channel system (Myosystem 1400L, Noraxon USA Inc., Scottsdale AZ, USA) recorded muscle activity at 1000 Hz which was subsequently amplified and band-pass filtered (3db at 20 – 500 Hz). Motion artifacts were attenuated by wrapping electrodes and wires with elastic bandage. Electrodes were placed on mid-belly of the Gluteus Maximus (GM), Rectus Femoris (RF), Long head of the Biceps Femoris (BFL), Vastus Medialis (VM), Gastrocnemius (GAS), Soleus (SOL), and Tibialis Anterior (TA) on the dominant limb for the NoAmp group and bi-laterally for the TTAmp group, except for the SOL and TA on the amputated limb. Electrodes were placed on the belly of the muscle, across subjects, identified while the subject flexed the muscle against resistance, e.g. the belly of the amputated GAS was found by having the subject flex their knee against resistance. The dominant limb (DOM) was determined by having the NoAmp group cycle at 15 Nm and 90 rpm for approximately two minutes. Pedal force and position data were recorded and the dominant limb was the limb producing the greater torque.

EMG data were rectified and low-pass filtered (Butterworth fourth order, zero-lag digital filter, 10Hz cutoff frequency) to provide a linear envelope. Twenty consecutive pedaling cycles were normalized to 1000 points per cycle and averaged.

Muscle onset and offset were determined setting a threshold of 20% maximum activation (Baum & Li, 2003). Onset and offset were calculated in software and visually verified.

Statistical significance was set at $p = 0.05$. T-tests for independent groups were used to analyze anthropometric and load condition variables between groups. A one way ANOVA was used to compare mean and peak joint moments, pedal forces, joint kinematics and EMG data between the dominant limb of the NoAmp group (DOM-NoAmp), the amputated limb of the TTAmp group (AMP-TTAmp), and the sound limb of the TTAmp group (SND-TTAmp). If statistical significance occurred with the one-way ANOVA, a Tukey post-hoc test was used to determine significance between limbs.

Results

The SND-TTAmp and DOM-NoAmp limbs produced greater peak torque about the crank spindle and torque averaged over the cycle than the AMP-TTAmp limb ($p < 0.05$, Figure 2, Table 2). Work asymmetry (the difference in the percentage each limb contributed to total work, (Sanderson, 1990)) was significantly greater in the TTAmp group ($24.5\% \pm 10.0$) than in the NoAmp group ($4.5\% \pm 3.4$).

The magnitude of the hip extensor moment in the amputated limb (AMP-TTAmp) during the 90-270 degree cycle phase tended to be lower than that of the DOM-NoAmp and SNDTTAmp. The knee extensor moment averaged over the cycle was lower in the AMP-TTAmp than in the DOM-NoAmp and SND-TTAmp limb ($p < 0.05$, Table 2). The amputated limb demonstrated a reduced knee flexor moment magnitude during the 180-270 degree

phase compared to DOM-NoAmp ($p < 0.05$, Figure 3) and a reduced knee flexor moment averaged over the cycle compared to DOM-NoAmp ($p < 0.05$, Table 2). The passive ankle extensor moment averaged over the cycle in AMP-TTAmp was smaller ($p < 0.05$, Table 2) and the peak moment shifted to later in the cycle (Figure 3) when compared to the DOM-NoAmp and SND-TTAmp ankle muscle moments.

The RPP joint was located at the intersection of the distal residuum and the inferior aspect of the prosthetic socket (Childers et al., 2012). The angular range of motion between the residual limb and the socket of the prosthesis was $4.8^\circ \pm 1.8^\circ$ (Figure 4). An RPP joint extensor moment would tend to rotate the prosthesis counter-clockwise (Figure 5). The RPP joint moment was flexor during the first ninety degrees of the crank cycle (Figure 5) in combination with a movement toward flexion (Figure 4). The bottom of the pedal stroke ($90^\circ - 180^\circ$) consisted of an extensor moment at the RPP joint (Figure 5) in combination with a movement toward joint extension (Figure 4). The recovery phase ($180^\circ - 360^\circ$) consisted of a greatly reduced RPP joint extension moment (Figure 5) and a movement toward joint flexion (Figure 4).

Muscle timing, i.e. EMG onset and offset, and time of peak EMG (Figure 6) demonstrated no significant differences when comparing SND-TTAmp to DOM-NoAmp. There were however, notable differences in timing of two muscles between the amputated limb and non-amputated limbs in SND-TTAmp and DOM-NoAmp groups. The rectus femoris (RF) in AMP-TTAmp demonstrated a significant shift in EMG onset, peak and offset to later in the cycle compared to DOM-NoAmp and SND-TTAmp ($p < 0.05$, Figure 6). The GAS in the amputated limb also demonstrated a significant shift in peak activation toward later in the crank cycle. Technical difficulties leading to large signal artifact in EMG samples taken with electrodes within the prosthetic socket resulted in the loss of 3 datasets for the ampGAS. The remaining six datasets were used for the analysis. Both muscles (RF and GAS) in the AMP-TTAmp group failed to show the typical bimodal activation observed in the DOM-NoAmp group (Figure 7).

Data from limb kinematics, pedaling kinetics, and muscle activation are summarized, schematically, in Figure 8 to provide a more integrated picture of limb-prosthesis system mechanics and control.

Discussion

The TTAmp group demonstrated greater cycling work asymmetry than the NoAmp group (Table 2, Fig. 2). These results are similar to data reported previously (Broker & Gregor, 1996; Childers et al., 2011a) suggesting study volunteers were representative of a larger population. A significant observation was that during prosthetic cycling there was substantial rotation at the RPP joint. Additionally, while muscle activity patterns from the gluteus maximus (GM), long head of the biceps femoris (BF) and vastus medialis (VM) in the amputated limb were not different from the SND-TTAmp or the DOM-NoAmp, the timing characteristics and patterns of the muscle activity in rectus femoris (RF) and gastrocnemius (GAS) were significantly altered in the amputated limb. These results support

the hypothesis that selected changes in muscle patterns would occur as a result of amputation.

Functional significance of altered muscle activity in the amputated limb

Peak ampGAS activation occurred significantly later in the cycle (Figures 6 and 7) and was similar to a previous report (Childers et al., 2009a). A trans-tibial amputation surgically alters the gastrocnemius from a two-joint (ankle extensor/knee flexor) to a one-joint (knee flexor) muscle. Given the altered muscle's mechanical actions, ampGAS muscle activation was expected to change in accordance with muscle's new function. AmpGAS peak activation shifted closer to peak knee flexor moment compared to the SND-TTamp and DOM-NoAmp limbs allowing for the partially amputated GAS to contribute to the knee flexor moment together with the hamstrings (Figures 3 and 7). Biceps femoris short head activity, a single joint knee flexor, was not recorded in this experiment but its activity has been predicted via computer simulation with EMG onset, offset and peak timing similar to the experimental data from ampGAS presented here (Neptune & Hull 1998). It appears that the observed ampGAS activation corresponds to that of a one-joint knee flexor, i.e. the activity occurred in-phase with the knee flexor moment (given the electromechanical delay of gastrocnemius (Li and Baum, 2004; Prilutsky and Gregor, 2000; van Ingen Schenau et al., 1995)), in phase with its two-joint synergist (biceps femoris long head), out-of-phase with the knee extensor moment, and out-of-phase with the activity of its one-joint antagonist (vasti) (Prilutsky, 2000) (Figures 3 and 7). A shift in ampGAS peak activation toward the peak knee flexor moment as well as the significant difference in the EMG peak timing from the unaltered GAS in the DOM-NoAmp limb suggest the motor system altered ampGAS activation timing to exploit its new biomechanical role. This example illustrates the neuromuscular system's ability to find functionally appropriate activation patterns to match the capabilities of remaining muscles to control the altered musculoskeletal system.

Effective prosthetic socket control would require appropriate limb positioning such that forces and moments acting at the knee joint and distal residuum generate the necessary motion at the RPP joint to control forces at the foot through the prosthesis (Radcliffe, 1962). Movement at the RPP joint is controlled via musculature that does not cross the joint. Besides dedicating ampGAS activation toward the knee flexor moment, the observed compensatory changes in RF activity in the amputated limb could also contribute toward RPP joint control. The first quadrant of the pedaling cycle ($0^\circ - 90^\circ$) involved large hip and knee extensor moments (Figure 3) and GM, BF and VM activity (Figure 6). The second quadrant ($90^\circ - 180^\circ$) involved a knee flexor moment (Figure 3) with corresponding BF and ampGAS activity (Figure 6). The RPP joint was flexed during the first quadrant of the cycle, extended to a more parallel orientation to the prosthesis in the second quadrant reaching peak extension at 180° in the cycle (Figure 4) in conjunction with BF and ampGAS activity (Figures 6 and 7). Thus, the observed motor patterns in the amputated limb contribute to extension at the RPP joint, i.e. moving residuum in parallel with the prosthetic socket in late power phase thus enhancing limb to pedal force transmission (Figure 8).

The top of the pedal stroke ($315^\circ - 45^\circ$) is discomforting in TTamps because the RPP joint is flexed (Figure 4) and the posterior-proximal and anterior-distal walls of the socket apply

pressure to the residuum (Childers et al., 2009b). Excessive RPP joint flexion would increase pressure on the residuum skin (Figure 8) and could potentially cause skin breakdown (Kristinsson, 1993). Unloading the residual limb skin and soft tissue could be considered an additional task performance criterion especially during the top of the pedal stroke. This region of the pedaling cycle is a transition region normally associated with RF activation (Figure 6) to aid hip flexion and knee extension and direct pedal forces more anteriorly in preparation for the power phase (Ting et al., 1999). The DOM-NoAmp and SND-TTAmp limbs demonstrated activation of the RF (Figure 7) similar to other reports (Ryan & Gregor 1992; Ting et al., 1999), whereas the RF in the amputated limb had delayed EMG onset, peak and offset (Figures 6 & 7). The RPP joint was flexed during this region ($315^{\circ} - 45^{\circ}$, Figure 4) placing it in a position of increased discomfort (Childers et al., 2009b). RF activation in this phase would increase pressure on the residuum at the anterior distal tibia and the popliteal area of the knee. RF activity in the amputated limb was delayed until after the prosthesis cleared TDC (Figure 8). That might be partly responsible, in combination with muscle atrophy (Schmalz et al., 2001), for the reduced knee extensor moment early in the crank cycle (Figure 3), and the reduced crank torque of the amputated limb necessitating sound limb compensation (Figure 2). Altered RF activity may help to explain prior research by Childers et al. (2011a) that demonstrated that cycling asymmetry was not solely due to strength/inertial imbalances between limbs. In short, the timing and pattern of RF activity may have been altered to account for additional performance criterion, i.e. minimizing stress on residuum tissues and this resulted in reduced output by the amputated limb.

Possible neural control mechanisms

Users of prosthetic limbs offer a unique model to understand motor control because motor system adaptation to the loss of system's select elements can provide information on neuromuscular locomotor control in an intact human. Locomotion is controlled by central pattern generators (CPGs) whose operation is modulated by descending supraspinal and limb motion-dependent afferent signals (McCrea and Rybak, 2008; Rossignol, 1996). Descending drive from the midbrain locomotor region regulates locomotor rhythm and changes inter-leg coordination (Shik et al., 1966), whereas stimulation of muscle and skin afferents can change duration and magnitude of extensor or flexor motoneuronal activity depending on phase of stimulation (Duysens et al., 2000). The lack of load sensory input from the amputated foot sole and ankle extensors during extensor phase, normally enhancing the duration and magnitude of the CPG extensor half-center in the ipsilateral leg (Pearson, 2008; Rybak et al., 2006), is expected to reduce the extensor knee and hip moments and the crank torque during the power phase (Table 2, Figures 2 and 3). This reduction in extensor activity may contribute to facilitation of flexor activity in the contralateral leg of TTAmp volunteers (Ting et al., 1998) (see Figure 2, SND-TTAmp and Figure 3, SND-TTAmp knee and ankle moments). The compensatory activity adjustments of ampGAS and RF in the amputated leg and extensors of the sound contralateral leg in the amputees (Figures 6 and 7) could be discovered during voluntary modifications of commands to motoneuronal synergistic groups at the CPG pattern formation level (Markin et al., 2012; Yakovenko et al., 2011) and reinforced by skin afferents in the residuum interacting with the prosthetic socket during early stages of learning to pedal with the prosthetic leg. Extended cycling practice

may strengthen the neural pathways in the brain (Cramer et al., 2011) and spinal cord (Wolpaw, 2012) that proved to be advantageous. These possible mechanisms would allow a prosthesis user to adapt to the mechanical constraints of the prosthetic interface while utilizing remaining physiological systems, e.g., muscular, skeletal, and neural that proved to be advantageous.

These data highlight how an altered motor system adapts to control an external device (i.e. prosthesis). The control was accomplished with utilizing the amputated gastrocnemius as a one-joint knee flexor and altering the activity of the rectus femoris. This research highlights the need to understand prosthetic locomotion as an interaction between the residual limb and prosthesis. Future research may compare how an intact neuromuscular system adapts to control similar mechanical systems, e.g. orthotic devices, and how socket designs may be improved through development of geometry altering socket shapes that could enhance control.

Acknowledgments

The authors gratefully acknowledge the individuals that volunteered for this research and supportive faculty and staff of School of Applied Physiology, especially Dr. Mindy Millard-Stafford for the use of the Lode cycle ergometer.

Funding Source: National Institutes of Health grant T32 HD055180-01A1.

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Figure 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.

Figure 2.

Crank moment for the dominant limb in the group without amputation (DOM-NoAmp, small dashes), the sound limb of the TTA group (SND-TTAmp, large dashes) and the amputated limb in the TTA group (AMP-TTAmp, solid). Positive values indicate clockwise moment putting energy into the bicycle.

Figure 3.

Hip, knee, and ankle joint moments for the dominant limb in the intact group (DOM-NoAmp, small dashes), the sound limb of the TTA group (SND-TTAmp, large dashes) and the amputated limb in the TTA group (AMP-TTAmp, solid).

Figure 4.

Angular motion of the residuum with respect to the prosthesis (mean $\pm$ SD). Zero angle indicates a parallel alignment between the residuum and the prosthesis. Positive values represent joint flexion or the posterior/proximal portion of the prosthesis moving toward the knee joint.

Figure 5.

Moment at the residuum/prosthesis pseudo-joint. Positive values represent an extensor moment. An extensor moment tends to rotate the prosthetic socket counter clockwise about the residuum so that the anterior/proximal portion of the prosthetic socket moves toward the knee joint.

Figure 6.

Timing of muscle EMG onset, peak and offset (mean±SD). Technical difficulty resulted in only 6 complete datasets for the amputated gastrocnemius. †= statistically significant difference between the prosthetic and sound limbs of the TTA group (AMP TTAmp, white bars and SND-TTAmp, gray bars, respectively). ♦ = statistically significant difference between the SND-TTAmp and the dominant limb (DOM-NoAmp, black bars) of the intact group.

Figure 7.

Normalized EMG activity (mean $\pm$ SD) of the Rectus Femoris (top panel) and Gastrocnemius (bottom panel) recorded in the amputated limb (AMP-TTamp, black line) and in intact group (DOM-NoAmp, shaded region representing $\pm$ SD from the mean). Technical difficulty resulted in six datasets for the amputated GAS.

Figure 8.

Schematic of the amputated limb and the prosthesis through the crank cycle. Force vectors indicate forces at the knee and RPP joint calculated with inverse dynamics and measured at the foot/pedal interface. Joint moments indicated by curved arrows, magnitude represented by line style (see legend). Muscle activity is represented by line thickness. Activity for the iliopsoas muscle is assumed based on joint moments as well as Juker et al., 1998.

Table 1Group characteristics (mean $\pm$ SD)

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

Table 2

Joint moment averaged over the cycle (in Nm; mean $\pm$ SD).

	Positive Crank Moment	Negative Crank Moment	Ankle Extensor Moment	Knee Extensor Moment	Knee Flexor Moment	Hip Extensor Moment
NoAmp, Dominant Limb	18.0 $\pm$ 4.3	12.4 $\pm$ 7.7	19.2 $\pm$ 5.3	12.7 $\pm$ 5.9	20.1 $\pm$ 3.3 [†]	42.7 $\pm$ 12
TTAmp Group, Sound Limb	20.0 $\pm$ 4.8	7.2 $\pm$ 1.8	20.0 $\pm$ 5.7	15.4 $\pm$ 4.5	17.4 $\pm$ 5.0	43.8 $\pm$ 9.9
TTAmp Group, Amputate d Limb	11.9 $\pm$ 1.4 $\blacklozenge^{\dagger}$	5.4 $\pm$ 1.9 $\blacklozenge$	13.3 $\pm$ 2.3 $\blacklozenge^{\dagger}$	8.4 $\pm$ 5.0 [†]	14.0 $\pm$ 3.9 $\blacklozenge$	37.4 $\pm$ 10.0

[†] statistically significant difference from the sound limb of the TTAmp group.

$\blacklozenge$ statistically significant difference from the dominant limb of the NoAmp group