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Compensatory mechanisms of transtibial amputees during circular turning

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ABSTRACT

Turning plays a prominent role in daily living activities and requires the modulation of the ground reaction forces to accelerate the body's center-of-mass along the path of the turn. With the ankle plantarflexors being prominent contributors to the propulsive ground reaction forces, it is not clear how transtibial amputees perform turning tasks without these important muscles. The purpose of this study was to identify the compensatory mechanisms used by transtibial amputees during a simple turning task by analyzing the radial and anterior–posterior ground reaction impulses and sagittal, transverse and coronal joint work of the residual and intact legs. These quantities were analyzed with the residual leg on both the inside and outside of the turn and compared to non-amputees. The analysis showed that amputees and non-amputees use different joint strategies to turn. Amputees rely primarily on sagittal plane hip joint work to turn while non-amputees rely primarily on ankle work in the sagittal plane and hip joint work in the coronal plane. Differences in strategies are most likely due to the minimal power output provided by the passive prosthetic feet used by amputees and perhaps a desire to minimize the risk of falling. Understanding these differences in turning strategies will aid in developing effective rehabilitation therapies and prosthetic devices that improve amputee mobility.

1. Introduction

An important goal for successful rehabilitation of lower-limb amputees is to return patients to normal daily living activities, in which mobility tasks such as turning play a prominent role [1]. To redirect the center-of-mass, turning requires modulation of the radial ground reaction forces (GRFs) to accelerate the body's center-of-mass towards the center of curvature. Glaister et al. [2] and Strike and Taylor [3] analyzed 90° left and right turns performed by non-amputees and found that modulation of the radial ground reaction force impulse (GRI) is critical to successfully perform the turning tasks. Orendurff et al. [4] investigated joint kinematic and kinetic changes in non-amputees while walking along a 1 m radius circular path and found that relative to straight-line walking, the primary changes occurred in the radial GRI. In addition to redirecting of the center-of-mass, turning requires the rotation of the body towards the new heading [5,6], requiring further modulation of lower-limb joint moments. Non-amputee turning required similar magnitudes of plantarflexor power generation relative to straight-line walking, which suggests that reduced ankle push-off power would also impact turning gait. Therefore, compensations in joint work are likely needed for amputees to perform a turn, especially since they are only able to minimally modulate the power output of passive foot-ankle prosthetic devices in response to changes in task demands (e.g., increased walking speed [7]).

Although analyses of three-dimensional joint power and work have provided much insight into the ability of non-amputees to both propel and control the lower limbs during straight-line walking [8–11], a limited number of studies have analyzed the three-dimensional joint powers in amputee gait [12]. Sadeghi et al. [13] examined three-dimensional peak powers at the hip, knee and ankle of transtibial amputees ($n = 5$) during straight-line walking at their self-selected speed wearing a SACH foot and found differences in peak power between the intact and residual legs for the hip in all three planes, the knee in the sagittal and coronal planes and the ankle in the sagittal plane. Since turning likely requires even larger modulations in the other planes of motion, analyzing three-dimensional joint work for each leg may provide further insight into the compensatory mechanisms required for amputees to complete a turning task.

Few studies have analyzed amputee turning. Segal et al. [14] compared the transverse plane joint moments and radial GRIs between amputees and non-amputees while walking along a 1 m radius circular path at the same speed. They found that amputees had decreased internal rotation moments at the residual leg hip

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and knee, which may result from a protective response to reduce stress at the prosthesis-skin interface and facilitate the change in orientation. Perhaps as compensation, the amputees increased the external rotation moment for their intact leg hip compared to non-amputees. Amputees also decreased their inside leg radial GRI compared to non-amputees, possibly to minimize the center-of-mass acceleration in the direction of the turn and maximize stability. The prescription of a torsion adapter, a device with an adjustable transverse plane stiffness, further reduced the internal rotation moments at the knee and hip while on the inside of a turn [15]. However, neither study analyzed the anterior–posterior GRIs nor explored how amputees modulate their joint powers and work to generate needed GRIs in the anterior–posterior or radial directions.

The purpose of this study was to identify the compensatory mechanisms used by amputees during turning by analyzing the anterior–posterior and radial GRIs and the hip (sagittal, transverse and coronal), knee (sagittal) and ankle (sagittal and coronal) joint power and work of the residual and intact legs. These measures were chosen after considering the structure and functional capabilities of each lower leg joint and analyzed with the residual leg on both the inside and outside of the turn and compared to non-amputee turning. In order to avoid confounding effects associated with transient turns, we had the subjects complete a steady-state turn about a defined circle. We tested the general hypothesis that there would be differences in the GRIs and joint work between the intact, residual and non-amputee legs regardless of whether the leg was on the inside or outside of the turn. Understanding how transtibial amputees compensate to successfully perform the turning task will provide needed insight for designing effective rehabilitation therapies focused on specific muscle groups and prosthetic interventions intended to improve amputee mobility.

2. Methods

2.1. Participants

Ten unilateral transtibial amputee participants (9 males, 1 female, age = 56 ± 12 yrs, height = 1.79 ± 0.08 m, mass = 88 ± 11 kg) and ten speed-matched non-amputee participants (6 males, 4 females, age = 44 ± 14 yrs, height = 1.73 ± 0.10 m, mass = 78 ± 21 kg) participated in this institutional review board-approved study. All amputees considered themselves moderately active community ambulators, had walked with a prosthesis for at least two years, wore their prosthesis at least 8 h a day, and were free from known neurological or musculoskeletal disorders. All amputee participants wore their clinically prescribed prosthetic components (Table 1) and provided informed consent prior to the study.

2.2. Data collection

Thirty-eight reflective markers were placed bilaterally on the arms, legs, trunk and head of the participants, consistent with Vicon's Plug-In-Gait model (Oxford Metrics, Oxford, England). A 10-camera Vicon 612 system collected kinematic data at a minimum of 120 Hz. GRFs were collected at a minimum of 1200 Hz from two consecutive Bertec force plates (Columbus, OH) embedded in the walkway. Raw marker trajectories were filtered using Vicon's Woltring quintic spline algorithm with MSE value set to 20.

Participants walked at their self-selected speed along a 1 m radius circular path while keeping their body centered over the path marked on the floor. Participants were told that stepping on the line was allowed and practiced the task before data collection. A constant speed 1 m radius circular path was chosen to explore the mechanisms of turning because it represented a typical turn radius found in daily activities (e.g., a 90° hallway turn) but minimized the confounding effects of variable speed observed during non-circular turning gait [16]. Each discrete trial consisted of participants walking clockwise around the 1 m radius circular path starting a few strides from the force plates and ending a few strides after the plates in order to achieve a steady-state pattern throughout force plate contact. The discrete trials were repeated until three force plate hits per leg were measured, and again for the counter-clockwise direction. Subjects were allowed to rest between trials and no subject reported feelings of dizziness. Walking speed was measured for each turning trial by averaging the instantaneous velocity of the outside leg's posterior superior iliac spine marker across the time period when subjects walked across the force plates.

2.3. Data analysis

Kinematic and GRF data were low-pass filtered in Visual 3D (C-Motion Inc., Germantown, MD) using a fourth-order Butterworth filter with cut-off frequencies of 6 Hz and 20 Hz, respectively. Intersegmental joint angles and moments were determined from marker trajectories using standard inverse kinematics and dynamics techniques (e.g., [17]). Joint powers were calculated as the product of the three-dimensional joint moments and corresponding angular velocities. Positive (negative) joint work was calculated as the time integral of the positive (negative) joint power over the gait cycle.

A rotating reference frame was defined with the origin at the center-of-mass, the x-axis through the center of the 1 m radius circular path and the y-axis (forward progression) perpendicular to the x-axis and tangent to the circle which the subjects were following [14]. Positive radial GRFs (x-axis) were defined in the direction of the circle center. Positive anterior–posterior GRFs (y-axis) were defined in the anterior direction. Positive (negative) GRIs were calculated as the time integral of the positive (negative) GRFs over the gait cycle.

Amputee data were grouped according to leg placement with respect to the circular path (condition). Right-leg amputee data from the clockwise trials were grouped with left-leg amputee data from the counter-clockwise trials to obtain *Residual Inside* and *Intact Outside* data. The *Residual Outside* and *Intact Inside* data were similarly obtained. In non-amputee gait, the right leg has been cited as the dominant and mobilizing leg [18,19]; therefore, both amputee legs were compared to the right leg of non-amputees. *Non-amputee Outside* data was obtained from counter-clockwise trials and *Non-amputee Inside* data was obtained from clockwise trials.

Table 1
Prosthetic components for transtibial amputee participants. Total contact patellar tendon bearing (PTB) sockets, rigid pylons, and Alpha® liners[a] were used by all participants, except subject 8 who used the Iceross® liner.[g]

Subject	Cause of amputation	Socket	Liner	Suspension	Foot/Ankle	Socks
1	Traumatic	Carbon laminated	6 mm uniform cushion	sleeve	Endolite Dynamic Response 2/ Multiflex Ankle[b]	Knit-Rite™ 2 ply[h]
2	Vascular	Carbon laminated	6 mm tapered cushion	sleeve	Freedom Innovation Runway[c]	Royal Knit™ 5 ply[h]
3	Diabetic infection	Thermal plastic	9 mm uniform locking	pin	FS-3000[c]	Knit-Rite™ 5 ply[h]
4	Traumatic	Carbon laminated	6 mm uniform locking	pin	FS-1000[c]	None
5	Diabetic/Vascular	Carbon laminated	9 mm uniform locking	pin	Dynamic Response/ Multi-axial ankle[d]	None
6	Traumatic	Carbon laminated	6 mm uniform locking	pin	Genesis[e]	Knit-Rite™ 3 ply[h]
7	Traumatic	Carbon laminated	6 mm contoured locking	pin	Seattle Lite Foot[f]	None
8	Traumatic	Carbon laminated	3 mm Iceross®	pin	Vari-Flex Low Profile[g]	Knit-Rite™ 1 ply[h]
9	Diabetic infection	Carbon laminated	9 mm uniform locking	pin	Low Profile Renegade[c]	Knit-Rite™ 5 ply, 1 ply[h]
10	Tumor (cancer)	Carbon laminated	6 mm contoured locking	pin	Low Profile FS2000[c]	Knit-Rite™ 5 ply[h]

[a] Ohio Willow Wood, Mt. Sterling, OH.
[b] Blatchford, Endolite North America, Centerville, OH.
[c] Freedom Innovation Inc, Irvine, CA.
[d] College Park Industries, Fraser, MI.
[e] Genesis Prosthetic Arts, Howell, MI.
[f] Seattle Systems, Poulsbo, WA.
[g] Ossur America's, Aliso Viejo, CA.
[h] Knit Rite Inc., Kansas City, KA.