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# Transtibial amputee joint rotation moments during straight-line walking and a common turning task with and without a torsion adapter

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**Abstract—**Amputees lack movement and control mechanisms at the foot and ankle that result in different strategies for locomotion than nonamputees. The torsion adapter is a prosthetic device designed to minimize shear stress at the residual limb by facilitating rotation in the transverse plane. This study determined if the addition of a torsion adapter alters lower-limb joint rotation moments of transtibial amputees walking in a straight line and turning. Ten transtibial amputees wore either a torsion adapter or a rigid adapter for an acclimation period of 3 weeks in random order. Ten nonamputees were also included for comparison. Kinetics were collected as participants walked in a straight line and around a 1 m-radius circular path at their self-selected turning walking speed. When amputee participants wore the torsion adapter, they demonstrated decreased prosthetic-limb peak internal rotation moments at the inside limb knee and hip compared with when they wore the rigid adapter, which may facilitate changes in orientation by not actively resisting the turn. Nonamputees exhibited larger moments compared with the prosthetic limb for both the amputee participants wearing either the torsion or rigid adapters. No differences were found in the moments for the intact limb between torsion and rigid adapter conditions during turning and for both limbs during straight-line walking.

**Key words:** amputee, kinetics, moment, prosthesis, rehabilitation, rotation, torque, torsion adapter, transtibial, transverse plane, turning.

## INTRODUCTION

Discomfort and residual-limb pain are two main concerns for lower-limb amputees [1]. These problems are often derived from transmitted loads that occur during ambulation and may be exacerbated with turning gait. Turning is quite prevalent in daily ambulation [2–3] and is especially important to the household ambulator who is mobility impaired and whose daily step count mostly comprises short bouts of walking with frequent changes of direction [4]. These individuals have more difficulty turning, since they take more time and steps to complete a turn [5–7] and fall more frequently, resulting in injuries [8–9]. Turning is likely more difficult for lower-limb amputees because of their inability to modulate transverse plane joint torque, one of the turning mechanisms for nonamputees [10]. Because amputees have a portion

**Abbreviations:** NA = nonamputee (participants), RA = (amputee participants wearing) rigid adapter, SD = standard deviation, TA = (amputee participants wearing) torsion adapter.

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of the kinetic chain replaced with a prosthesis, ground reaction torques are transferred through the prosthetic components to the socket-residual-limb interface, which may lead to pain and discomfort. Most prostheses allow limited motion in the transverse plane. Consequently, the addition of a compliant element that facilitates transverse rotation may decrease the transmitted torques and enhance residual-limb comfort.

The torsion adapter is a prosthetic device designed to facilitate transverse plane rotation in the range that exists during nondisabled gait and decrease the torque applied to the residual limb through the prosthetic socket [11]. By enabling transverse rotation between the pylon and the residual limb, these devices may improve gait and minimize stresses on the residual limb, leading to an increase in comfort and mobility during daily ambulation. However, little objective evidence exists related to the effect of torsion adapters on amputee gait. Lamoureux and Radcliffe examined two transfemoral amputees using a prototype torsion device and reported increased rotation between the pelvis and the socket (3.0°–8.5°) as well as decreased torque at the shank [11]. Van der Linden et al. examined two transfemoral amputees and found that for one participant, the torque absorber increased axial rotation of the socket relative to the foot [12]. They suggested a reduction in the relative motion between the residual limb and the socket may decrease shear forces and, therefore, decrease skin breakdown at the residual limb. Finally, Flick et al. reported the mechanical properties of five torsion adapters, including their range of motion and stiffness [13]. Further testing of the effect of torsion adapters on the biomechanics of transtibial amputee gait and maneuvering may help researchers, clinicians, and prosthetic designers better understand the functional capabilities and limitations of the present devices.

Defining how a device that incorporates transverse plane rotation alters the joint rotational torques during a turning task may help clarify how amputees accomplish a turn, what limitations are associated with the present components, and ultimately identify design criteria for improving prosthetic technology. The present study determined if a commercially available torsion adapter can reduce transtibial amputee joint torques when compared with a rigid adapter during straight-line walking and turning gait, as well as how both conditions compare with nonamputee gait.

## METHODS

### Participants

The 10 unilateral transtibial amputee participants (1 female) who gave informed consent to participate in this investigational review board-approved study were 56 ± 12 years old (ranging from 27 to 70 years), with a height of 1.79 ± 0.08 m and mass of 88 ± 11 kg. (Values throughout article are expressed as mean ± standard deviation [SD] unless otherwise indicated.) All participants met the following criteria for inclusion in this study:

1. Wore a prosthesis for at least 2 years before the study and 8 hours each day.
2. Considered themselves moderately active community ambulators.
3. Did not use upper-limb aids.
4. Had no history of falls within the previous 6 months.
5. Were free from neurological deficits and underlying musculoskeletal disorders that might have affected gait characteristics by self-report.

Prosthetic prescription and torsion adapter settings for each participant are shown in **Table 1**.

The 10 nonamputee (NA) participants (4 females) who also gave informed consent to participate in this study were 44 ± 14 years old (ranging from 24 to 65 years), with a height of 1.73 ± 0.10 m and mass of 78 ± 21 kg. They reported themselves to be free from known neuromuscular or orthopedic pathologies.

### Prosthetic Components

We chose a transverse plane torsion adapter with adjustable torsion stiffness (4R85, Otto Bock; Duderstadt, Germany) for the present study (**Figure 1(a)**). Settings on the torsion adapter ranged from –5 to 5 in integer increments, where 5 was the stiffest possible setting. This particular device was chosen over other torsion adapters, since torsional stiffness was adjustable and motion was limited to the transverse plane. To blind the participant to the intervention condition, we concealed a rigid adapter (4R103, 2 Clamp Rigid Adapter, Otto Bock; Duderstadt, Germany) with a black plastic cover and weighted it to match the torsion adapter (±3 g) by securing small lead ball bearings with epoxy inside the rigid pylon (**Figure 1(a)–(b)**) at the proximal end (**Figure 2**). All other prosthetic components were held constant throughout the study (**Table 1**).

**Table 1.**
Specific components used by each transtibial amputee who completed study. All participants used total contact patellar tendon-bearing sockets and Alpha® liners* (except participant 8 who used Iceross liner†).

| Participant | Amputation Cause | Socket | Liner | Suspension | Foot | Socks | Torsion Adapter Setting |
|---|---|---|---|---|---|---|---|
| 1 | Traumatic | Carbon laminated | 6 mm uniform cushion | Sleeve | Endolite Dynamic Response‡ | Knit-Rite™ 2 ply§ | 4 |
| 2 | Vascular | Carbon laminated | 6 mm tapered cushion | Sleeve | Freedom Innovation Runway¶ | Royal Knit™ 5 ply§ | 0 |
| 3 | Diabetic infection | Thermal plastic | 9 mm uniform locking | Pin | FS-3000¶ | Knit-Rite™ 5 ply§ | 5 |
| 4 | Traumatic | Carbon laminated | 6 mm uniform locking | Pin | FS-1000¶ | None | 4 |
| 5 | Diabetic/ vascular | Carbon laminated | 9 mm uniform locking | Pin | College Park** | None | 5 |
| 6 | Traumatic | Carbon laminated | 6 mm uniform locking | Pin | Genesis†† | Knit-Rite™ 3 ply§ | 4 |
| 7 | Traumatic | Carbon laminated | 6 mm contoured locking | Pin | Seattle Lite Foot†† | None | 5 |
| 8 | Traumatic | Carbon laminated | 3 mm Iceross | Pin | Vari-Flex Low Profile† | Knit-Rite™ 1 ply§ | 5 |
| 9 | Diabetic infection | Carbon laminated | 9 mm uniform locking | Pin | Low Profile Renegade¶ | Knit-Rite™ 5 ply, 1 ply§ | 5 |
| 10 | Tumor (cancer) | Carbon laminated | 6 mm contoured locking | Pin | Low Profile FS2000¶ | Knit-Rite™ 5 ply§ | 5 |

*Ohio Willow Wood; Mt. Sterling, Ohio.
†Ossur Americas; Aliso Viejo, California.
‡Blatchford, Endolite North America; Centerville, Ohio.
§Knit-Rite Inc; Kansas City, Kansas.
¶Freedom Innovation Inc; Irvine, California.
**College Park Industries; Fraser, Michigan.
††Genesis Prosthetic Arts; Howell, Michigan.
‡‡Seattle Systems; Poulsbo, Washington.

### Prosthetic Fitting

A research prosthetist initially set the stiffness of the torsion adapter, according to the manufacturer's recommendations based on each amputee's body mass and activity level. Following a 1-week acclimation period to the torsion adapter, each participant discussed the stiffness setting with the prosthetist and he adjusted it as necessary. If an adapter was adjusted, an additional 1-week acclimation period was provided, followed by an assessment. This process continued until both the participant and prosthetist were satisfied. The participant was then randomly assigned to either keep the torsion adapter or switch to a rigid adapter. Both adapters were placed on the proximal end of the pylon for minimizing the effect of added mass during the swing phase of gait (**Figure 2(b)**). A StepWatch™ 2 Activity Monitor (OrthoCare Innovations; Mountlake Terrace, Washington) was attached to the distal end of the pylon throughout the acclimation period for confirming daily usage of the limb. After a 3-week acclimation period, participants returned for the first data collection session for pylon 1. Then, the pylon was switched to pylon 2 and the protocol was repeated. Thus, each subject was evaluated with and without the torsion adapter.

### Experimental Procedures

Thirty-five 14 mm reflective markers were placed on each participant at locations consistent with Vicon's Plug-in-Gait full-body model (Oxford Metrics; Oxford, England). Measurements were taken for each individual according to the Vicon requirements for static and dynamic modeling. Participants were instructed to walk around a 1 m-radius circle at constant walking speed, both

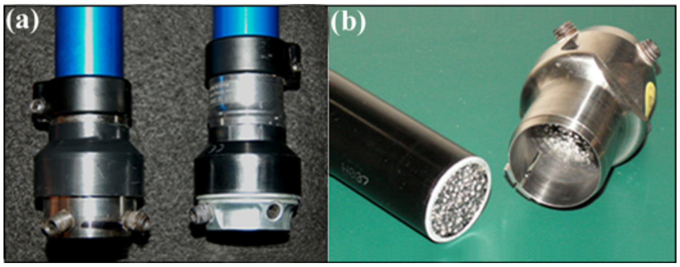


**Figure 1.**
**(a)** Rigid (left) (4R103, 2 Clamp Rigid Adapter, Otto Bock; Duderstadt, Germany) and torsion adapter (right) (4R85, Otto Bock) used in present study. Black plastic covering concealed devices so that they appeared similar. **(b)** Epoxy and ball bearings were added to rigid adapter to match weights of two adapters to within ±3 g. (Note: Adapter in this photograph was cut to display setup.)

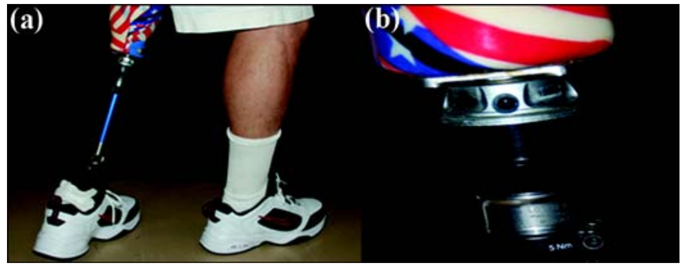


**Figure 2.**
**(a)** Transtibial amputee participant wearing torsion adapter while turning. **(b)** Both torsion and rigid adapters were placed on proximal end of pylon, just below socket.

clockwise and counterclockwise, for three repeated trials for each adapter condition, keeping their body centered on a line marking a circular path. They were told that stepping on the line did not matter. Participants were allowed to practice this task before data collection. Preliminary trials were collected, both clockwise and counterclockwise, until two trials in each direction were within 10 percent of the same walking speed. Continuous walking speed was measured throughout each turning trial by the method of tracking velocity of the outside limb's posterior superior iliac spine marker immediately following each data collection. These preliminary trials allowed the subjects to acclimate to the turning task and established their self-selected turning walking speed. A constant speed 1 m-radius turn was chosen because it represented a typical radius turn found in daily activities (e.g., a 90° hallway turn) but minimized the confounding effects of variable speed gait [14]. Participants also completed three repeated trials walking in a straight line at the same walking speed as the turning trials.

Gait kinematics and kinetics were collected with use of a 12-camera Vicon system, Plug-in-Gait model (Oxford Metrics; Oxford, England) and two force plates (Bertec; Columbus, Ohio) embedded in the walkway. This data collection procedure was completed by the same amputee participants wearing the rigid adapter (RA) and then again by amputee participants wearing the torsion adapter (TA). NA participants only completed the data collection once. (For purposes of this article, further references to RA and TA encompass the amputee participants wearing the rigid and torsion adapters, respectively, and NA encompasses the NA participants.)

## Data Analysis

Gait kinematics and kinetics were calculated with workstation software, and peak internal (positive by convention) and external (negative by convention) transverse plane rotation moments at the ankle, knee, and hip were extracted from each trial using the Event Analyzer software (Vaquita Software; Zaragoza, Spain) and then averaged across trials for each condition. Transverse plane rotation adapter motion was resolved at the knee joint because the shank wand defining rotation was located distal to the transverse plane rotation adapter. The rotation moment calculations were unaffected by the location of the adapter because the location of the joint centers and the orthogonal distance of the force to the axis of rotation of a uniform pylon do not change with the addition of a 1 degree of freedom device placed proximal to the sagittal ankle joint. The prosthetic limb, intact limb, and NA data were analyzed at the inside and outside limbs while subjects turned as well as when they walked in a straight line, with use of repeated measures one-way analysis of variance. Analyses were conducted with the use of R 2.6.1 [15]. Statistical significance was set at $p < 0.05$. Specific $p < 0.1$ values are reported to enable the reader to interpret the strength of each relationship.

# RESULTS

The average self-selected turning walking speeds for NA, RA, and TA were within 10 percent of each other: 0.92 ± 0.07, 0.88 ± 0.10, and 0.87 ± 0.13 m/s, respectively. The straight-line walking trials were controlled to be within 10 percent of the self-selected turning walking speeds and measured as 0.94 ± 0.06, 0.90 ± 0.10, and 0.90 ± 0.13 m/s for NA, RA, and TA, respectively. The peak external moment occurred at ~12 percent of the gait cycle and the peak internal moment occurred at ~50 percent of the gait cycle (**Figure 3**). The peak rotation moments (mean ± 1 SD) at the ankle, knee, and hip are reported in **Tables 2, 3**, and **4** for the inside limb during turning (**Table 2**), outside limb during turning (**Table 3**), and straight-line walking (**Table 4**) for RA, TA, and NA.

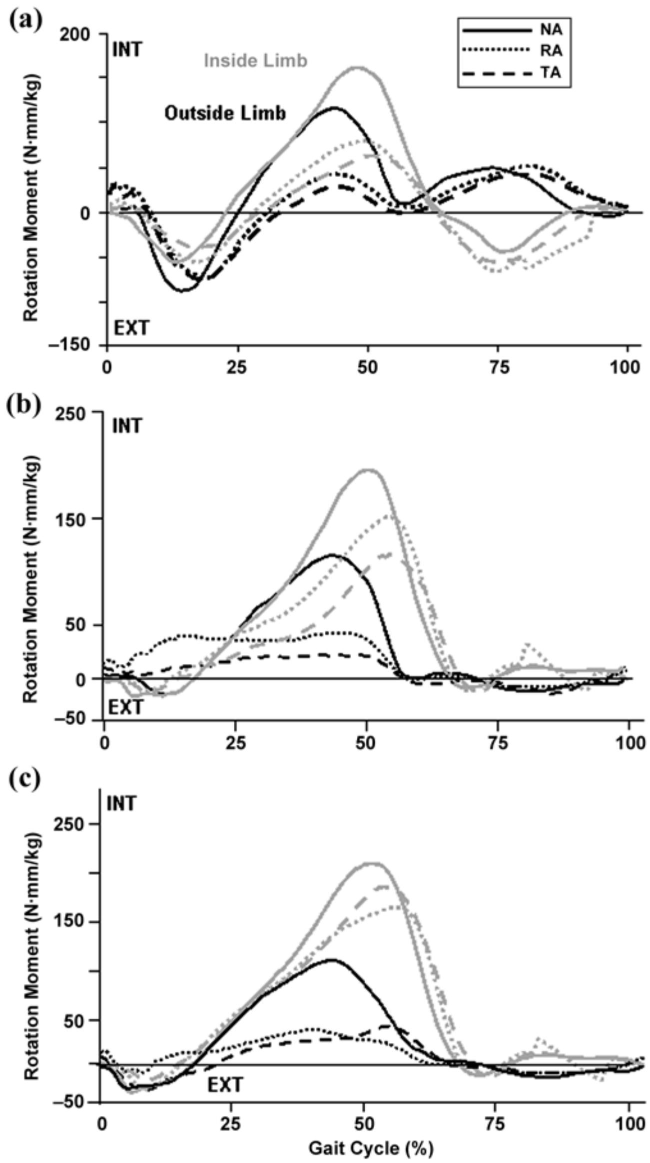


**Figure 3.**
Mean rotation moments (N•mm/kg) during turn for **(a)** hip, **(b)** knee, and **(c)** ankle joints of nonamputees (NA) ($n$ = 10) compared with prosthetic limb of amputees ($n$ = 10) wearing rigid adapter (RA) and torsion adapter (TA). Negative external moment occurs in early stances, and positive internal moment occurs in late stance. Inside limb values are gray and outside limb values are black. EXT = external, INT = internal.

The main comparison of RA versus TA with associated $p$-value was presented first in **Tables 2, 3,** and **4**, followed by the secondary comparisons of RA versus NA and TA versus NA. The intact and prosthetic limbs of the amputee participants were examined separately.

Differences across pylons were found for the prosthetic-limb peak internal rotation moments at the knee on the inside and outside of the turn and at the hip on the inside of the turn: NA group was larger than RA, and RA was larger than TA (**Figures 3–4**). Straight-line walking data at a comparable walking speed to the turning trials are included in **Figure 4** as a comparison to the turning results. The straight-line walking rotation moments did not demonstrate differences across pylons (RA vs TA); however, the peak internal rotation moments for the amputee prosthetic-limb hip and knee of amputees were decreased compared with NAs (**Table 4**). An overall trend was found for TA to be less than RA during straight-line walking, but this comparison did not reach statistical significance (**Figure 4**).

Almost no differences were found at the ankle. The one exception was peak ankle external rotation moment of the prosthetic limb for TA was less than NA for the inside limb turning. At the knee, the outside limb turning peak external rotation moment of the prosthetic limb for RA was smaller than NA.

The intact-limb rotation moments were unaffected by the inclusion of a torsion adapter for all joints. However, some of the comparisons with NA were significant. Peak knee external rotation moment for the outside limb of NA was greater compared with the intact limb of TA (**Table 3**). In addition, the hip joint for the inside limb demonstrated increased peak external rotation moment for the intact limb of RA and TA compared with NA (**Table 2**).

## DISCUSSION

Amputees have reduced lower-limb joint motion compared with NAs, which may significantly affect the joint torques, decreasing efficiency and increasing shear stress at the residual-limb-socket interface. The effect of decreased transverse plane motion may be exacerbated in more complex maneuvering activities such as turning, which incorporates a change of orientation and heading. Therefore, this study was conducted to compare the joint rotation torques of TA, RA, and NA during straight-line walking and a turning task.

**Table 2.**
Peak internal and external rotation moments (mean ± 1 standard deviation) of inside limb turning for ankle, knee, and hip of nonamputee (NA) participants ($n$ = 10) and of participants ($n$ = 10) wearing rigid adapter (RA) and torsion adapter (TA). Intact and prosthetic limbs of amputees were examined separately. Comparisons were tested for significance with repeated measures one-way analyses of variance. Statistical significance was set at $p < 0.05$. Values of $p > 0.15$ were designated not significant (NS) statistically.

| Inside Limb | RA | TA | RA vs TA (*p*-Value) | NA | RA vs NA (*p*-Value) | TA vs NA (*p*-Value) |
|---|---|---|---|---|---|---|
| Ankle (N•mm//kg) | | | | | | |
| Internal Prosthetic | 186 ± 80 | 206 ± 83 | NS | 224 ± 52 | NS | NS |
| Internal Intact | 230 ± 103 | 225 ± 80 | NS | 224 ± 52 | NS | NS |
| External Prosthetic | –36 ± 48 | –25 ± 16 | NS | –39 ± 24 | NS | 0.039 |
| External Intact | –37 ± 30 | –42 ± 23 | NS | –39 ± 24 | NS | NS |
| Knee (N•mm/kg) | | | | | | |
| Internal Prosthetic | 157 ± 43 | 128 ± 37 | 0.046 | 206 ± 36 | 0.002 | <0.001 |
| Internal Intact | 216 ± 61 | 216 ± 76 | NS | 206 ± 36 | NS | NS |
| External Prosthetic | –26 ± 27 | –24 ± 15 | NS | –30 ± 20 | NS | NS |
| External Intact | –34 ± 34 | –41 ± 26 | NS | –30 ± 20 | NS | NS |
| Hip (N•mm/kg) | | | | | | |
| Internal Prosthetic | 97 ± 42 | 65 ± 45 | 0.006 | 174 ± 43 | <0.001 | <0.001 |
| Internal Intact | 148 ± 87 | 146 ± 59 | NS | 174 ± 43 | NS | NS |
| External Prosthetic | –54 ± 47 | –57 ± 45 | NS | –66 ± 38 | NS | NS |
| External Intact | –129 ± 62 | –140 ± 63 | NS | –66 ± 38 | 0.004 | 0.001 |

**Table 3.**
Peak internal and external rotation moments (mean ± 1 standard deviation) of outside limb turning for ankle, knee, and hip of nonamputee (NA) participants ($n$ = 10) and of participants ($n$ = 10) wearing rigid adapter (RA) and torsion adapter (TA). Intact and prosthetic limbs of amputees were examined separately. Comparisons were tested for significance with repeated measures one-way analyses of variance. Statistical significance was set at $p < 0.05$. Values of $p > 0.15$ were designated not significant (NS) statistically.

| Outside Limb | RA | TA | RA vs TA (*p*-Value) | NA | RA vs NA (*p*-Value) | TA vs NA (*p*-Value) |
|---|---|---|---|---|---|---|
| Ankle (N•mm/kg) | | | | | | |
| Internal Prosthetic | 82 ± 64 | 73 ± 98 | NS | 121 ± 44 | 0.06 | 0.12 |
| Internal Intact | 140 ± 129 | 111 ± 58 | NS | 121 ± 44 | NS | NS |
| External Prosthetic | –28 ± 31 | –42 ± 33 | NS | –42 ± 29 | NS | NS |
| External Intact | –31 ± 34 | –36 ± 44 | NS | –42 ± 29 | NS | NS |
| Knee (N•mm/kg) | | | | | | |
| Internal Prosthetic | 69 ± 26 | 49 ± 18 | 0.007 | 122 ± 37 | <0.001 | <0.001 |
| Internal Intact | 121 ± 67 | 121 ± 56 | NS | 122 ± 37 | NS | NS |
| External Prosthetic | –7 ± 21 | –15 ± 15 | NS | –28 ± 24 | 0.004 | 0.07 |
| External Intact | –16 ± 18 | –13 ± 14 | NS | –28 ± 24 | 0.07 | 0.01 |
| Hip (N•mm/kg) | | | | | | |
| Internal Prosthetic | 53 ± 36 | 35 ± 38 | NS | 124 ± 47 | <0.001 | <0.001 |
| Internal Intact | 88 ± 61 | 104 ± 46 | NS | 124 ± 47 | 0.09 | NS |
| External Prosthetic | –70 ± 54 | –88 ± 45 | NS | –102 ± 60 | NS | NS |
| External Intact | –89 ± 48 | –91 ± 34 | NS | –102 ± 60 | NS | NS |

The NA peak external and internal rotation moments during a turn were similar to previously reported literature on turning biomechanics in nondisabled participants [10,16]. Orendurff et al. also demonstrated a decrease in hip internal rotation moment for the outside limb for a "turn continuation" step compared with walking in a straight line (straight line: 164.4 ± 51.8 vs turning: 92.8 ± 86.4 N•mm/kg) [10]. In addition, the prosthetic-limb

**Table 4.**
Peak internal and external rotation moments (mean ± 1 standard deviation) of straight-line walking for ankle, knee, and hip of nonamputee (NA) participants ($n$ = 10) and of participants ($n$ = 10) wearing rigid adapter (RA) and torsion adapter (TA). Intact and prosthetic limbs of amputees were examined separately. Comparisons were tested for significance with repeated measures one-way analyses of variance. Statistical significance was set at $p < 0.05$. Values of $p > 0.15$ were designated not significant (NS) statistically.

| Straight Line | RA | TA | RA vs TA ($p$-Value) | NA | RA vs NA ($p$-Value) | TA vs NA |
|---|---|---|---|---|---|---|
| Ankle (N•mm/kg) | | | | | | |
| Internal Prosthetic | 112 ± 57 | 118 ± 108 | NS | 157 ± 40 | 0.03 | NS |
| Internal Intact | 164 ± 93 | 134 ± 53 | NS | 157 ± 40 | NS | NS |
| External Prosthetic | –50 ± 70 | –38 ± 24 | NS | –37 ± 22 | NS | NS |
| External Intact | –46 ± 46 | –33 ± 24 | NS | –37 ± 22 | NS | NS |
| Knee (N•mm/kg) | | | | | | |
| Internal Prosthetic | 84 ± 39 | 74 ± 33 | NS | 155 ± 28 | <0.001 | <0.001 |
| Internal Intact | 156 ± 71 | 147 ± 51 | NS | 155 ± 28 | NS | NS |
| External Prosthetic | –17 ± 17 | –24 ± 18 | NS | –27 ± 16 | 0.04 | NS |
| External Intact | –34 ± 25 | –30 ± 21 | NS | –27 ± 16 | NS | NS |
| Hip (N•mm/kg) | | | | | | |
| Internal Prosthetic | 70 ± 34 | 58 ± 61 | NS | 152 ± 38 | <0.001 | <0.001 |
| Internal Intact | 131 ± 72 | 122 ± 54 | NS | 152 ± 38 | NS | 0.07 |
| External Prosthetic | –87 ± 60 | –93 ± 54 | NS | –85 ± 51 | NS | NS |
| External Intact | –127 ± 56 | –137 ± 56 | NS | –85 ± 51 | 0.03 | 0.03 |

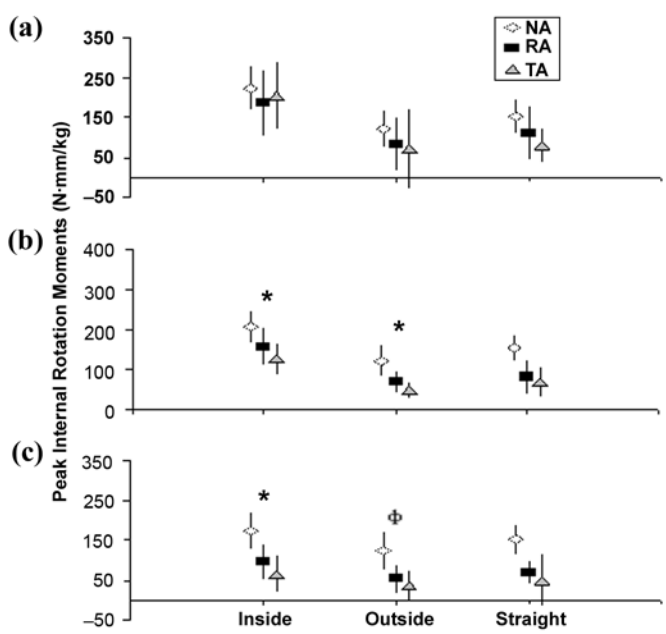


**Figure 4.**
Mean ± 1 standard deviation peak internal rotation moments (N•mm/kg) for **(a)** ankle, **(b)** knee, and **(c)** hip of nonamputees (NA) ($n$ = 10) compared with prosthetic limb of amputees ($n$ = 10) wearing rigid adapter (RA), and amputees wearing torsion adapter (TA). * = statistically significant differences across all comparisons (nonamputee vs rigid vs torsion). Φ = statistically significant differences only between nonamputees versus amputees (both rigid and torsion conditions). Straight-line walking data are included for general comparison.

rotation moments of the ankle, knee, and hip for RA were within the previously reported range of rotation moments of transfemoral amputees walking in a straight line with different prosthetic feet [17].

The addition of a torsion adapter decreased peak internal rotation moment for the prosthetic-limb knee (inside and outside limbs) and hip (inside limb) of the transtibial amputee in late-stance during turning. This decreased moment was consistent with a case study that examined a transfemoral amputee during straight-line walking and showed decreased rotation moment with a torsion adapter, as well as increased rotation between the socket and the foot, indicating that the socket rotated with the residual limb [11]. At heel contact on the prosthetic limb, amputees activate the gluteal and hamstring muscles to extend their hip, which coincides with generation of an external rotation moment along the long axis of the fully extended lower limb. A torque absorber appears to alter the loading rate of the torsional shears applied to the residual limb. The spring properties of the device enable loading over a longer period with lower peak stress, decreasing shear at the residual limb [12]. However, without a torsion adapter, the prosthesis is stiff in the transverse plane, increasing shear stress from the external rotation moment. The ability to modulate the internal rotation moment is even more critical for turning because the moment decreases for the outside limb and increases for the inside limb compared with straight-line walking (**Figure 4**) [10]. NA, RA, and TA demonstrated a similar relationship, where the internal rotation moment of the outside limb was less than straight and the internal rotation moment of the inside limb was greater than straight for all conditions. Furthermore, the torsion adapter condition decreased the internal rotation moment in late stance on the prosthetic limb, which may facilitate changes in orientation by not actively resisting the turn.

The inclusion of a torsion adapter did not affect the rotation moments for the intact limb. Since changes occurred for the prosthetic limb at the knee and hip, changes on the intact limb were also expected. However, changes may have also occurred in other parts of the system, in planes besides the transverse plane, or in biomechanical metrics besides the moment (i.e., angle, power, or impulse). The torsion adapter may also have a larger effect on the rotation moments during a turn initiation step or turn termination step for a transient turn than during a turn continuation step at constant walking speed, as presented in this article. This particular constant-speed turning task was chosen because walking speed affects gait biomechanics [14]; therefore, with walking speed controlled, we could most likely attribute any differences found between biomechanical variables to a change in intervention and not in speed.

## CONCLUSIONS

The torsion adapter minimally affected the lower-limb transverse plane rotation moments of transtibial amputees walking in a straight line. However, when a turning maneuver was performed, the peak knee and hip internal rotation moments for the inside prosthetic limb of TA were decreased compared with RA, and the peak knee internal rotation moment for the outside prosthetic limb of TA was also decreased compared with RA. If reduced transverse plane moments lower shear stresses applied to the residual limb, torsion adapters (like the one tested here) will likely reduce discomfort and incidence of injury experienced by lower-limb amputees.

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**Author Contributions:**

*Study concept and design*: A. D. Segal, M. S. Orendurff, J. M. Czerniecki, G. K. Klute.

*Analysis and interpretation of data*: A. D. Segal, M. S. Orendurff, J. M. Czerniecki, J. B. Shofer, G. K. Klute.

*Drafting of manuscript*: A. D. Segal, G. K. Klute.

*Critical revision of manuscript for important intellectual content*: A. D. Segal, M. S. Orendurff, J. M. Czerniecki, J. B. Shofer, G. K. Klute.

*Statistical analysis*: A. D. Segal, J. B. Shofer.

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**Additional Information:** Mr. Michael Orendurff is now with the Movement Science Laboratory, Texas Scottish Rite Hospital for Children, Dallas, Texas.

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