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A General Model for Estimating Lower Extremity Inertial Properties of Individuals with Transtibial Amputation

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Abstract

Lower extremity joint moment magnitudes during swing are dependent on the inertial properties of the prosthesis and residual limb of individuals with transtibial amputation (TTA). Often, intact limb inertial properties (INTACT) are used for prosthetic limb values in an inverse dynamics model even though these values overestimate the amputated limb's inertial properties. The purpose of this study was to use subject-specific (SPECIFIC) measures of prosthesis inertial properties to generate a general model (GENERAL) for estimating TTA prosthesis inertial properties. Subject-specific mass, center of mass, and moment of inertia were determined for the shank and foot segments of the prosthesis (n = 11) using an oscillation technique and reaction board. The GENERAL model was derived from the means of the SPECIFIC model. Mass and segment lengths are required GENERAL model inputs. Comparisons of segment inertial properties and joint moments during walking were made using three inertial models (unique sample; n = 9): (1) SPECIFIC, (2) GENERAL, and (3) INTACT. Prosthetic shank inertial properties were significantly smaller with the SPECIFIC and GENERAL model than the INTACT model, but the SPECIFIC and GENERAL model did not statistically differ. Peak knee and hip joint moments during swing were significantly smaller for the SPECIFIC and GENERAL model compared with the INTACT model and were not significantly different between SPECIFIC and GENERAL models. When subject-specific measures are unavailable, using the GENERAL model produces a better estimate of prosthetic side inertial properties resulting in more accurate joint moment measurements for individuals with TTA than the INTACT model.

Introduction

Segmental inertial properties (segment mass, center of mass location, and moment of inertia) can be determined using a variety of methods. Variability in estimates of segmental inertial properties, in general, has little influence on joint moment magnitudes during the stance phase of walking (Challis, 1996; Challis and Kerwin, 1996; Smith et al., 2014). This is due to the influence of the large ground reaction force (GRF) acting on the bottom of the foot. During the swing phase, in the absence of a moment due to the GRF, other parameters such as segmental inertial properties and segmental motions have a much larger influence. The importance of accurate segmental inertial properties on joint moment estimates has been shown in several populations including children (Ganley and Powers, 2005), transfemoral amputees (Goldberg et al., 2008; Miller and Childress, 2005), transtibial amputees (Smith et al., 2014), and partial foot amputees (Dillon et al., 2008).

For a transtibial amputation (TTA), the mass of the prosthetic side from the knee down is 30 - 40% less, the center of mass location is 25 - 35% closer to the knee joint, and the moment of inertia is 50-60% less about a transverse axis through the knee joint when compared to the intact limb (Lin-Chan et al., 2003; Mattes et al., 2000; Smith et al., 2014). Although these known differences exist, rarely are these differences accounted for when modeling the limb for inverse dynamics estimates (Gates et al., 2013; Miller, 1987; Sawers and Hahn, 2010).

Smith et al. (2014) recently described an oscillation rack system to directly measure the inertial properties of the prosthesis and residual limb of persons with TTA to provide more accurate, subject-specific inertial properties. In an effort to understand the influence of prosthesis inertial properties on joint moment magnitudes in a group of TTAs, joint moments were estimated based on subject-specific measures of the prosthesis inertia and in a second model that made the assumption that prosthetic side inertial properties were similar to the intact side. Smith et al. (2014) reported significantly lower peak joint moment magnitudes during swing at the knee and hip for the model using subject-specific measures compared to the model using intact limb inertial properties. Despite the fact that the reaction board and oscillation techniques used by Smith et al. (2014) are accurate and reliable methods for directly measuring prosthesis inertial properties, these methods are time-consuming (~30 min) and require specialized equipment. There is a need to develop a prediction method that researchers can use to estimate the inertial properties of a TTA prosthetic limb without the need to directly measure the inertia of the prosthesis.

Thus, the purpose of this study was two-fold: 1) develop a general method (GENERAL) for estimating prosthesis inertial properties of the amputated limb based on means obtained from subject-specific (SPECIFIC) measures in a group of individuals with TTA; and 2) evaluate the validity of the GENERAL approach compared to SPECIFIC and intact limb (INTACT) approaches for inertial measures and evaluate these approaches as inputs into inverse dynamics calculations for lower extremity joint moments during walking. Thus, there were two main phases to this study. Phase I addressed model development and Phase II examined the soundness of the model.

Methods

Participants

Participants in both phases were between the ages of 18 and 65 years old, had a TTA resulting from trauma, wore their prostheses daily, and were free from other comorbidities that would influence their walking ability. IRB approval and written informed consent were obtained prior to data collections.

Phase I - Model Development

The GENERAL model was developed from the SPECIFIC inertial properties estimated for 11 people with unilateral TTA (9 males, 2 females, measured body mass = 95.5 ± 16.3 kg, height = 1.78 ± 0.07 m). SPECIFIC data were collected from participants in our lab (n = 5) and pooled with data from the literature (n = 6) obtained through personal communication with the authors (Smith et al, 2014).

Using traditional methods, body mass is measured using a standard scale (measured body mass, MBM). However, due to the loss of the limb, mass and MOI estimates of the shank and foot are underestimated in the intact limb. Therefore, the MBM was obtained (while the participant wore their shoes, prosthesis, liner, and any ply) and then adjusted (increased) to account for the missing mass in the intact limb (Smith et al., 2014). This adjusted body mass (ABM) more accurately represents the mass of the individual and was used to calculate the relative percentage of the mass of the amputated limb and foot (see figure 1A in the supplementary material for more details) (de Leva, 1996).

The type of prosthetic foot was limited to energy storing and releasing feet in an effort to limit the influence of other foot types on predictive measures. Prosthetic foot types varied between individuals due to their prescription based on the ability (K-level) of the individual. All participants were community ambulators, able to walk over varied terrains with variable cadence (K3-K4). Participants used various socket suspension systems including: suction, lock and pin, and elevated vacuum. The entire suspension system (including liner and ply) was included in the SPECIFIC measures of the prosthesis inertia.

SPECIFIC measures of prosthesis mass (with shoe), center of mass location (COM), and moment of inertia (MOI) about a mediolateral axis through the prosthesis COM were determined using a standard scale, reaction board, and oscillation techniques, respectively (Smith et al., 2014). The prosthesis (socket, foot, and shoe) was suspended within a support cage and oscillated with an amplitude of less than 5° while the period of oscillation was measured. This period of oscillation was then used to estimate the system's MOI about the oscillation axis. The inner cage with the prosthesis was then removed and the COM of the whole system was determined using a reaction board technique. Then the prosthesis was removed from the inner cage and the inner cage's COM location by itself was determined with the reaction board technique. Finally, the inner cage alone was suspended in the oscillation rack and its period of oscillation was measured. COM and MOI estimates for the inner cage alone trials were subtracted from the trials with the inner cage plus the prosthesis to leave estimates for the COM and MOI for just the prosthesis. Using the parallel axis theorem, the estimated MOI for the prosthesis about the oscillation axis was expressed about a transverse axis through the COM of the prosthesis.

Inertial properties of the residual limb were estimated by modeling the residual limb as the frustum of a right circular cone (Hanavan, 1964; Mattes et al., 2000) and assuming a uniform tissue density of $1.1 g \cdot cm^{-3}$ (Mungiole and Martin, 1990). Residual limb length and two circumferences - one just distal to the knee joint and the other near the distal aspect of the residual limb - were used as inputs into the model. Inertial properties of the prosthesis were combined with those of the residual limb to produce an overall limb inertia from the knee down. This overall limb inertia was then distributed into separate foot and shank segments.

First, the overall limb mass was divided into shank (66%) and foot (34%) masses based on proportions from 9 dismantled prostheses including a 0.9 kg liner. Using percentages from de Leva (1996) for an intact foot segment, the COM and radius of gyration (ROG) of the prosthetic foot were then estimated. These foot estimates were subtracted from the overall inertia of the limb (prosthesis + residual limb) to leave inertial estimates for the combined residual limb and prosthetic shank, which we refer to from this point forward as the prosthetic shank segment. These methods have been described in greater detail in previous papers (Smith et al., 2014; Smith and Martin, 2011, 2013).

After SPECIFIC measures for this group of 11 TTAs were obtained, a GENERAL model was created to estimate prosthetic shank and foot inertial properties from mean values of the group (Table 1). These means of prosthetic foot and shank masses were expressed as a percentage of ABM. Prosthetic shank COM and ROG lengths were expressed as a percentage of the prosthetic shank length (knee joint center to the ankle joint center) and expressed relative to the knee joint. Prosthetic foot COM and ROG were based on percentages reported by de Leva (1996) for an intact foot segment (with the shoe on).

However, we found the use of MBM in the GENERAL model produced similar results as the ABM and was less time consuming (see supplementary material). To use the GENERAL model, the researcher need only measure MBM, prosthetic shank length (knee center of rotation to ankle center of rotation) and foot length (with shoe) and apply these measures to the GENERAL model. These measurements and subsequent calculations take roughly 5 minutes to complete. In contrast, the use of the ABM requires the participant to remove their prosthesis, liner, and ply to take measurements. This process takes approximately 30 minutes, thus increasing the data collection times without producing meaningful improvements in the inertial estimates. To evaluate this choice, we compared the inertial properties and joint moments between the ABM and MBM from the GENERAL model and found no significant differences between the models. The data and results of this evaluation are presented in more detail in the supplementary material for this paper. Therefore, there are only three measures that are need in order to apply this GENERAL model to a transtibial population: 1) MBM, 2) shank length, and 3) foot length with the shoe.

[TABLE 1 HERE]

Phase II - Model Validation

Nine individuals with TTA (6 males, 3 females, measured body mass = 78.2 ± 18.8 kg, height = 1.76 ± 0.10 m), who were not included in the model development portion of the study, participated in the model validation phase. Inertial properties of the amputated limb were calculated using three approaches: (1) INTACT – prosthetic leg inertial properties were assumed to match those predicted for the intact leg estimated using de Leva (1996), (2) SPECIFIC – subject-specific measures as described above, (3) GENERAL– applying the MBM and segment lengths to the GENERAL model developed in Phase I.

To provide an example of the application of our GENERAL model (Table 1) and its influence on commonly reported joint moments, an inverse dynamics model of walking was used with this second group of participants. Participants walked at 1.5 $m \cdot s^{-1}$ along a 10-m walkway with embedded force plates. GRFs (2000 Hz) and motion data (100 Hz) were collected using the Plug-in-Gait marker set. Using a three segment inverse dynamics model, joint moments at the hip, knee, and ankle were computed using the three different inertial models: (1) INTACT, (2) SPECIFIC, and (3) GENERAL. To test for differences between approaches, a single factor (Inertial Model) MANOVA with repeated measures was performed on inertial properties and peak joint moments ($\alpha = .05$).

Post Hoc Analysis

During our analysis of the results, we noted three participants (# 2, 4, 8) had very small shank MOIs in the SPECIFIC model compared to other participants. We attributed these small MOIs to the assumed distribution of mass between the prosthetic shank and foot in the SPECIFIC model. The distribution of the prosthetic socket (66%) and foot (34%) masses from the overall prosthetic mass was based on the mean of 9 dismantled prostheses (including a 0.9 kg liner). However, as a percentage, prosthetic socket masses ranged between 54 - 75% and foot masses ranged between 26 - 46% of total prosthetic mass. Therefore, after our original analysis (presented in the results section), we tested how the MOIs of participants 2, 4, and 8 were affected if the distribution of mass was altered. We did this follow-up assessment by recalculating the subject-specific inertial properties using the highest shank/lowest foot mass percentages (75% shank, 26% foot) since the shank MOI appeared to be underestimated in the SPECIFIC model. We refer to these adjusted measures as ADJUSTED. This ADJUSTED model was then applied to the gait data to calculate joint moments.

Results

Model Validation

No statistically significant differences between the SPECIFIC and GENERAL model for estimates of shank mass ($p = .779$) and COM ($p = .056$) location (Table 2) were observed. Shank mass for the SPECIFIC model was ~14% higher than the GENERAL model (ranging from 0.13 – 0.92 kg difference). Similarly, the shank COM location was higher (~35%) for the SPECIFIC model than the GENERAL model. The absolute difference ranged from 0.00 – 0.04 m.

The SPECIFIC and the GENERAL model estimates of shank mass, resulted in a ~22% lighter mass ($p < .033$) than the INTACT model (~0.75kg less). COM location was significantly ($p < .001$) closer to the knee axis of rotation (~10cm, ~55%) for the SPECIFIC and the GENERAL model compared to the INTACT model.

However, the MOI in the SPECIFIC model varied dramatically between individuals. Three participants (#s 2, 4, 8) in particular, had very small MOI SPECIFIC measures compared to the other six participants. When compared to the GENERAL model, SPECIFIC MOI was on average larger than the GENERAL MOI, although no significant differences were found between the GENERAL and SPECIFIC models. The GENERAL model MOI was significantly ($p = .007$) smaller than the INTACT measures.

For the prosthetic foot, only the mass of the foot was significantly greater using INTACT compared with SPECIFIC ($p = .036$) and the GENERAL model ($p = .003$). All other measures for the prosthetic foot were not significantly different between models given that all measures were based on percentages reported by de Leva (1996).

[TABLE 2 HERE]

When the three models were applied to the inverse dynamics model of walking, moment magnitudes were not significantly different at the ankle regardless of inertia model (*Model Validation* group; n = 9). No significant differences were found among the three models at any joint at any time during stance phase. Peak joint moments at the knee ($p < .002$) and hip ($p < .001$) during late swing were significantly smaller for the SPECIFIC and GENERAL models compared with the INTACT model (Figure 1). There were no significant differences in moment magnitudes between SPECIFIC and GENERAL models.

[FIGURE 1 HERE]

Post Hoc Analysis

Results from the ADJUSTED measures produced shank masses and MOIs that were more similar to the rest of the cohort's SPECIFIC measures (Table 3). This suggests that the assignment of the mass distribution was driving the large differences in MOI and shank masses for these individuals.

[TABLE 3 HERE]

Additionally, it was unclear how the ADJUSTED model would alter the joint moments at the knee and hip during swing compared to the SPECIFIC and GENERAL models. Therefore, we applied the ADJUSTED measures for these three participants to the over-ground walking inverse dynamics models. Joint moments were calculated and peak hip and knee joint moments during swing were compared between the SPECIFIC, GENERAL, and ADJUSTED models. The peak moments between the three models did not result in large differences between the three models (Table 4). At the knee and hip, the differences in peak joint moments during swing ranged 0.01 - 0.05 $N \cdot m \cdot kg^{-1}$. These small differences suggest that while the inertial measures improved with adjustments to the mass distribution, the effects on outputs from inverse dynamics analysis was minimal. Therefore, the results from the ADJUSTED model also demonstrate that the SPECIFIC model outputs are reflective of the assumed mass distribution of the shank (66%) and foot (34%). Furthermore, while the distribution of the mass altered the MOI in the ADJUSTED model, when used as inputs into the inverse dynamics model, the resulting moments were similar to those of the SPECIFIC model. This suggests that the influence of the MOI on joint moments was small.

[TABLE 4 HERE]

Discussion

In Phase I of the study, we developed a GENERAL model based on subject-specific measures from 11 persons with TTA and their prostheses to estimate the inertial properties of the prosthetic limb (Table 1). In Phase II, we applied our developed model to a separate group of participants to assess the validity of our model. The results of our validation process suggest our GENERAL model was able to estimate body segment parameters of the amputated limb which did not differ from the SPECIFIC model. With the exception of the shank MOI, the variability of the COM and percent mass for our GENERAL model was very similar to the variability in non-amputees presented by previous literature (Dempster and Gaughran, 1967; Zatsiorsky, 2002).

When the GENERAL model body segment parameters were used in inverse dynamics equations during walking, the resulting joint moments did not differ from the SPECIFIC joint moments. However, both the GENERAL and SPECIFIC models resulted in significantly smaller joint moments than the INTACT. Thus, the GENERAL approach to estimating segment inertial properties is superior to an intact-limb approach especially when researchers will use inverse dynamics.

Given that the kinematic data were consistent between each model, we can conclude the kinetic differences seen between models (SPECIFIC, ADJUSTED, GENERAL, and INTACT) are related to differences in the inertial properties of the shank and foot. Although the MOI showed the largest difference between models, the COM location remained in a similar location between the SPECIFIC and GENERAL models. As a result, these results support previous work which suggests the largest sources of error in motion analysis are COM location (Challis and Kerwin, 1996) and accuracy of kinematic measures (Cappozzo, 1991; Riemer et al., 2008). The GENERAL model approximated segmental inertial properties and joint moments that were more similar to the SPECIFIC model than the INTACT model. This suggests that when SPECIFIC measures are not available, either GENERAL model (ABM or MBM) can be used to approximate segmental inertial properties.

Previous reports of joint kinetics during the swing phase of walking have shown small differences (~20%) between the intact and amputated limbs (Ferris et al., 2012; Powers et al., 1998). This study has shown that when the inertial properties of the limb are adjusted using the GENERAL model, the differences between the INTACT and GENERAL methods is nearly 50% during swing phase. Furthermore, the joint moments at the knee and hip during swing phase were significantly different between the INTACT measures and the SPECIFIC measures and GENERAL model. At the knee, joint moments were 46% and 33% smaller in the SPECIFIC and GENERAL models respectively than the INTACT measures. Hip moments showed even larger differences from the INTACT measures (90.8% SPECIFIC, 43.7% GENERAL). These results are similar to those shown by Smith et al. (2014) when comparing direct measures of prosthesis inertial and intact limb inertial measures. They similarly found significant differences between the SPECIFIC and INTACT measures during swing phase but not during stance. Thus, we conclude these new measures result in more accurate joint moments during swing phase rather than assuming the amputated limb has inertial properties similar to the intact limb.

The current study provides a simple solution to a long standing problem with lower extremity amputee research: the use of intact limb inertial property models are not appropriate for this population (Dillon et al., 2008; Goldberg et al., 2008; Miller and Childress, 2005; Sawers and Hahn, 2010). Future researchers working with individuals with below-knee prostheses now have a method for estimating inertial properties of the prosthetic side when subject-specific measures of the prosthesis are unavailable.

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Conflict of Interest Statement

All authors have no conflicts of interests to declare.

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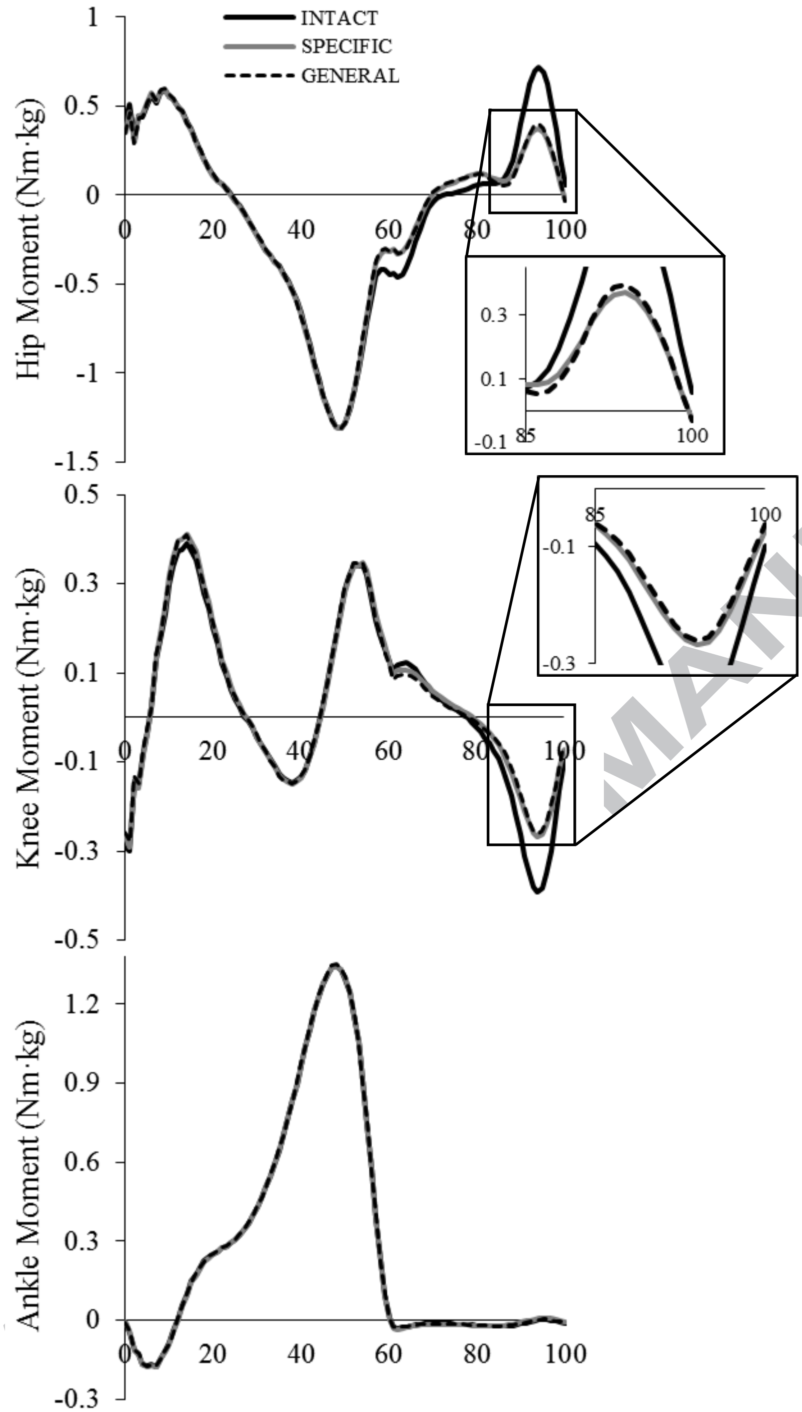

Figure 1. Mean hip, knee, and ankle joint moments of 9 TTAs using INTACT, SPECIFIC, and GENERAL inertial models. *Indicates INTACT measures were significantly different from the

SPECIFIC and GENERAL model (p < .05). No Significant differences were found between SPECIFIC and the GENERAL model.

Table 1. GENERAL model percentages for estimating inertial properties of the prosthetic shank and foot. These data are based on SPECIFIC measures (n = 11) from the model development phase (Phase I)

		Mass (%body mass)	COM (% segment length)	ROG (% segment length)
Shank[b]		3.3	21.0	17.1
	CI_{95}	(3.02, 3.59)	(17.4, 24.3)	(11.9, 22.2)
Foot[c]		1.4	[a]44.15m	[a]27.9m
			40.14w	24.5w
	CI_{95}	(1.39, 1.40)	n/a	n/a

Note: [a] Estimates based on de Leva (1996) were gender specific (m: men, w: women).
[b] Shank length is measured from the knee joint center to the ankle joint center (ankle joint center was mirrored from the intact ankle).
[c] Foot length is measured from the heel to the toe with the shoe on.

Table 2. Means and standard deviations of inertial properties calculated using the SPECIFIC, GENERAL, and INTACT models (n = 9). Shank data are presented in the top half, and foot data are presented in the bottom half. These data are from the model validation phase (Phase II) of the study. Shank COM is shown as a distance from the knee axis. Shank MOI is represented relative to a M-L axis through the shank COM.

		Shank Mass (kg)			Shank COM (m)			Shank MOI (kg·m²)		
ID	**Sex**	**SPECIFIC***	**GENERAL***	**INTACT**	**SPECIFIC***	**GENERAL***	**INTACT**	**SPECIFIC**	**GENERAL***	**INTACT**
1	f	1.94	1.75	2.31	0.084	0.086	0.180	0.064	0.009	0.023
2	m	3.35	2.7	3.54	0.057	0.095	0.200	0.001	0.016	0.044
3	m	2.83	2.22	2.90	0.093	0.084	0.178	0.031	0.011	0.029
4	f	2.18	2.03	2.69	0.077	0.075	0.158	0.003	0.008	0.021
5	m	3.3	3.56	4.72	0.067	0.098	0.207	0.023	0.023	0.063
6	m	2.24	2.37	3.15	0.039	0.076	0.160	0.013	0.009	0.025
7	f	2.64	2.4	3.19	0.102	0.084	0.178	0.054	0.011	0.032
8	m	3.12	2.69	3.54	0.059	0.087	0.182	0.001	0.013	0.037
9	m	2.69	3.53	4.73	0.063	0.078	0.165	0.041	0.014	0.04
Mean±SD		**2.70 ± 0.51**	**2.58 ± 0.62**	**3.42 ± 0.84**	**0.071 ± 0.020**	**0.085 ± 0.008**	**0.179 ± 0.017**	**0.026 ± 0.024**	**0.013 ± 0.005**	**0.035 ± 0.013**
		Foot Mass (kg)			**Foot COM (m)**			**FOOT MOI (kg·m²)**		
ID	**Sex**	**SPECIFIC***	**GENERAL***	**INTACT**	**SPECIFIC**	**GENERAL**	**INTACT**	**SPECIFIC**	**GENERAL**	**INTACT**
1	f	0.91	0.66	1.04	0.036	0.036	0.036	0.005	0.003	0.004
2	m	1.36	1.02	1.43	0.049	0.049	0.049	0.008	0.006	0.006
3	m	1.19	0.84	1.29	0.053	0.053	0.053	0.006	0.005	0.005
4	f	0.94	0.77	1.09	0.040	0.040	0.040	0.005	0.004	0.004
5	m	1.53	1.35	1.93	0.057	0.057	0.057	0.011	0.009	0.010
6	m	0.95	0.90	1.33	0.051	0.051	0.051	0.004	0.004	0.005
7	f	0.88	0.91	1.25	0.042	0.042	0.042	0.005	0.006	0.006
8	m	1.24	1.02	1.37	0.055	0.055	0.055	0.006	0.005	0.006
9	m	1.12	1.34	1.82	0.051	0.051	0.051	0.019	0.023	0.026
Mean±SD		**1.12 ± 0.23**	**0.98 ± 0.24**	**1.40 ± 0.27**	**0.048 ± 0.007**	**0.048 ± 0.007**	**0.048 ± 0.007**	**0.008 ± 0.005**	**0.007 ± 0.006**	**0.008 ± 0.007**

*Significant difference from INTACT ($p < .05$)

Table 3. Means and standard deviations of inertial properties calculated using the SPECIFIC, GENERAL and ADJUSTED models (n = 3). These data were created post hoc to compare inertial properties after they have been adjusted for shank and foot mass distributions.

	SHANK MASS (kg)			**SHANK COM (m)**			**SHANK MOI (kg·m^2)**		
ID#	**SPECIFIC**	**ADJUSTED**	**GENERAL**	**SPECIFIC**	**ADJUSTED**	**GENERAL**	**SPECIFIC**	**ADJUSTED**	**GENERAL**
2	3.35	3.60	2.70	0.06	0.09	0.09	0.001	0.048	0.016
4	2.18	2.34	2.03	0.08	0.10	0.08	0.003	0.020	0.008
8	3.12	3.35	2.69	0.06	0.09	0.09	0.001	0.038	0.013
Mean ± SD	**2.88 ± 0.62**	**3.10 ± 0.67**	**2.48 ± 0.38**	**0.06 ± 0.01**	**0.09 ± 0.01**	**0.09 ± 0.01**	**0.002 ± 0.001**	**0.035 ± 0.014**	**0.013 ± 0.004**
	FOOT MASS (kg)			**FOOT COM (m)**			**FOOT MOI (kg·m^2)**		
ID#	**SPECIFIC**	**ADJUSTED**	**GENERAL**	**SPECIFIC**	**ADJUSTED**	**GENERAL**	**SPECIFIC**	**ADJUSTED**	**GENERAL**
2	1.36	1.11	1.02	0.05	0.05	0.05	0.008	0.006	0.006
4	0.94	0.77	0.77	0.04	0.04	0.04	0.005	0.004	0.004
8	1.24	1.01	1.02	0.06	0.06	0.06	0.006	0.005	0.005
Mean ± SD	**1.18 ± 0.22**	**0.96 ± 0.17**	**0.94 ± 0.14**	**0.05 ± 0.01**	**0.05 ± 0.01**	**0.05 ± 0.01**	**0.006 ± 0.001**	**0.005 ± 0.001**	**0.005 ± 0.001**

Table 4. Means and standard deviations of peak hip and knee joint moments (Nm·kg) during terminal swing calculated using the SPECIFIC, GENERAL and ADJUSTED models (n = 3). These data were determined post hoc to compare inertial properties after they have been adjusted for shank and foot mass distributions.

	Knee			Hip		
ID #	**SPECIFIC**	**ADJUSTED**	**GENERAL**	**SPECIFIC**	**ADJUSTED**	**GENERAL**
2	-0.33	-0.34	-0.29	0.31	0.33	0.29
4	-0.27	-0.28	-0.24	0.63	0.65	0.49
8	-0.22	-0.24	-0.26	0.32	0.34	0.41
Mean ± SD	**-0.27 ± 0.05**	**-0.29 ± 0.05**	**-0.26 ± 0.03**	**0.42 ± 0.19**	**0.44 ± 0.18**	**0.39± 0.10**