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Technical note

The effect of direct measurement versus cadaver estimates of anthropometry in the calculation of joint moments during above-knee prosthetic gait in pediatrics

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Abstract

Joint reaction forces, moments and powers are important in interpreting gait mechanics and compensatory strategies used by patients walking with above-knee prostheses. Segmental anthropometrics, required to calculate joint moments, are often estimated using data from cadaver studies. However, these values may not be accurate for patients following amputation as prostheses are composed of non-biologic material. The purpose of this study was to compare joint moments using anthropometrics calculated from cadaver studies versus direct measurements of the residual limb and prosthesis for children with an above-knee amputation. Gait data were collected for four subjects with above-knee prostheses walking at preferred and fast speeds. Joint moments were computed using anthropometrics from cadaver studies and direct measurements for each subject. The difference between these two methods primarily affected the inertia couple ($I\alpha$ term) and the inertial effect due to gravity, which comprised a greater percentage of the total joint moment during swing as compared to stance. Peak hip and knee flexor and extensor moments during swing were significantly greater when calculated using cadaver data ($p<0.05$). These differences were greater while walking fast as compared to slow speeds. A significant difference was not found between these two methods for peak hip and knee moments during stance. A significant difference was found for peak ankle joint moments during stance, but the magnitude was not clinically important. These results support the use of direct measurements of anthropometry when examining above-knee prosthetic gait, particularly during swing.

Keywords: Above-knee prostheses; Anthropometrics; Kinetics; Gait

1. Introduction

Joint kinetics are important in the interpretation of gait mechanics and compensatory strategies used by patients walking with above-knee (AK) prostheses. Segment anthropometrics (i.e., mass, center of gravity (CG) and moment of inertia (MI)), needed to calculate joint reaction forces, moments and powers, are often estimated using regression equations from cadaver studies (Dempster et al., 1959; Clauser et al., 1969). While this is a reasonable approach for many patient populations, errors may be introduced when calculating anthropometrics for prostheses. Previous studies have reported limitations of anthropometrics based on cadaver data (Dempster et al., 1959; Clauser et al., 1969) for use in certain patient populations (Jensen, 1986).

Direct measurements of the residual limb and prosthesis have been used to calculate anthropometrics in several studies (Seroussi et al., 1996; Gitter et al., 1997; Fowler et al., 1999; van der Linden et al., 1999). Methods for directly calculating anthropometrics include knife-edge balancing to find the CG and pendulum tests to calculate the MI (Seroussi et al., 1996; Fowler et al., 1999; van der Linden et al., 1999). Negligible differences in joint moments calculated using cadaver data versus direct

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Table 1
Subject age, height and weight

Subject	Age (yrs)	Height (m)	Mass (kg)
1	14.1	1.75	67.50
2	15.3	1.75	71.60
3	12.9	1.64	55.45
4	11.9	1.44	32.55

measurements approaches have been reported for below-knee prosthetic gait (Czerniecki et al., 1991). Similar anthropometric studies could not be found for AK prosthetic gait. The purpose of this study was to compare joint moments calculated using anthropometry (1) estimated using cadaver data and (2) calculated using direct measurements of the residual limb and prostheses during AK prosthetic gait.

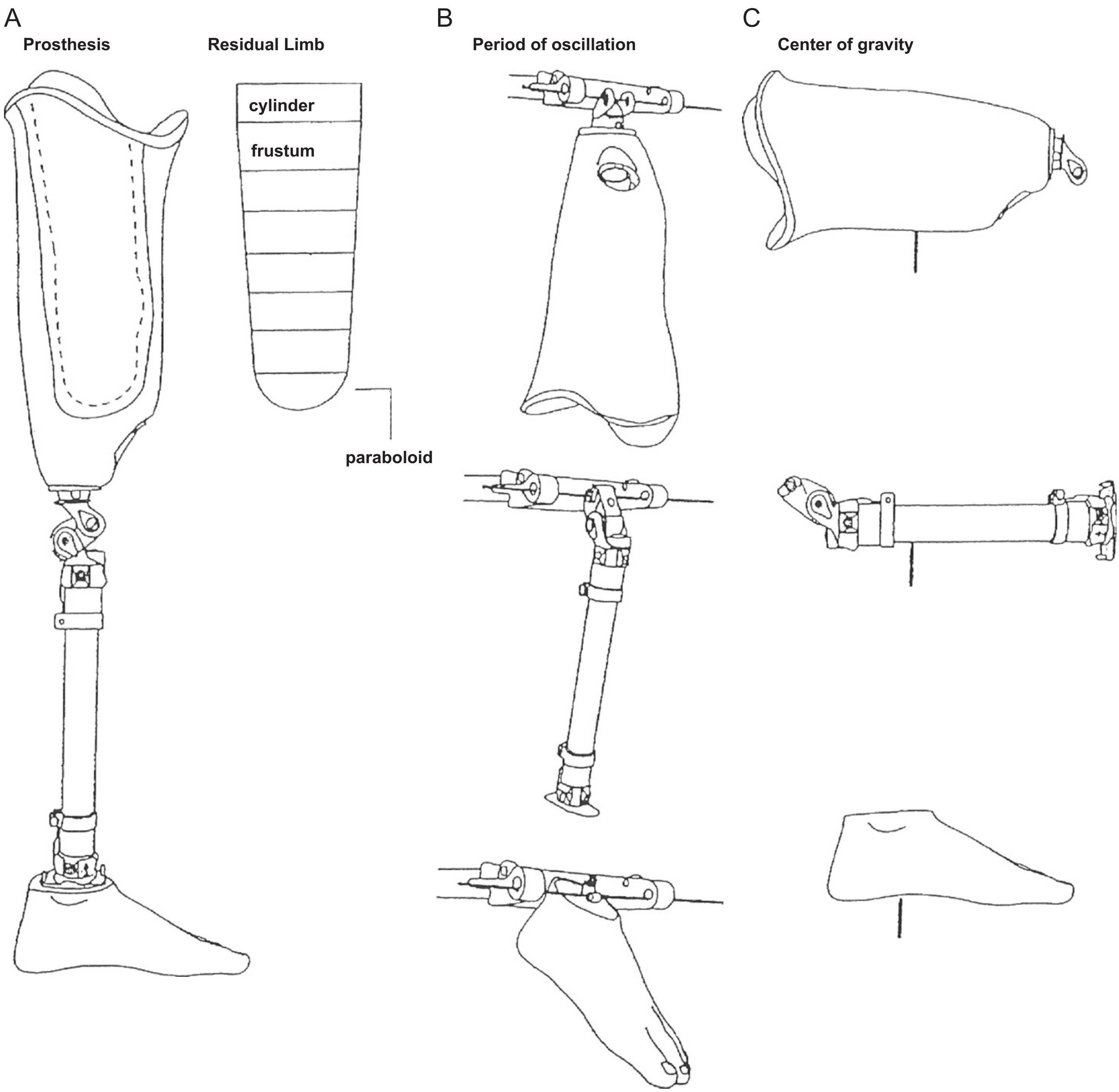

Fig. 1. Schematic diagram of methods used to determine prosthetic limb anthropometrics: (A) modeling of the residual limb, (B) measuring the period of oscillation and (C) measuring the center of gravity. The segments were oscillated to determine the period of oscillation (T). Moment of inertia (MI) of each prosthetic segment was calculated as $\mathrm{MI} = T^2\mathrm{mgd}/(2[\mathrm{pi}]^2)$. Corresponding residual limb and prosthetic components were combined to obtain one mass, center of mass and moment of inertia for each of the affected limb thigh, shank and foot segments.

2. Methods

2.1. Subjects

Four subjects with AK amputations participated in this study (two male and two female). Their average age was 13.5 ± 1.5 years (height $= 1.64 \pm 0.15$ m, weight $= 56.78 \pm 17.54$ kg) (Table 1). A certified prosthetist determined that each subject's prosthesis fit and functioned properly. Each subject could walk independently. Informed written consent was obtained from each subject and a legal guardian.

2.2. Anthropometric data

Anthropometric data for the non-prosthetic limb were calculated from regression equations derived by Dempster et al., (1959). MIs were calculated using reference mass and height (73.44 kg, 1.76 m) multiplied by the ratio of (mass)∗(height)^2/(reference mass)∗(reference height)^2 for each subject. Prosthetic limb anthropometrics were calculated using: (1) Dempster et al., (1959) assuming the anthropometrics were identical to the intact limb and (2) direct measurements for each subject's residual limb and prosthesis (Fowler et al., 1999). The circumference of the residual limb was measured using a tape measure. The distance between each measurement was 3 cm or less, and measurement locations were documented relative to the greater trochanter. Smaller intervals were used in order to obtain a sufficient number of measurements to calculate volume for short residual limbs. The residual thigh was modeled as a cylinder proximally, a series of frustra and a distal paraboloid with an assumed density of 1.0 g/ml to calculate the volume (Fig. 1) (Fowler et al., 1999). Segment mass and volume were used to calculate the MI and CG of the residual limb.

Each subject had an endoskeletal prosthesis with a metal pylon shank, a constant-friction knee and SACH or Seattle-type foot. Each prosthesis was disassembled, and anthropometrics were measured for the thigh, shank, and foot and shoe segments. The mass of each segment was measured on a calibrated scale, the CG was estimated by balancing the segment on a knife edge, and the MI was calculated using the pendulum technique (Fig. 1).

2.3. Gait data

The subjects wore shorts and low-cut comfortable shoes. A modified Helen Hayes marker set (Davis III et al., 1991) was used with reflective markers on the prosthetic limb placed on the thigh, shank, ankle and foot using analogous locations on the intact limb and on the mechanical knee joint center. Motion was recorded using an Eagle 8-camera system (Motion Analysis Corporation, Santa Rosa, CA) sampling at 60 Hz. Two forceplates (Kistler Instrumentation Corporation, Amherst, NY) were concealed in the walkway to record ground-reaction forces sampled at 1 kHz. Gait data were collected during walking at preferred and fast speeds. For fast speeds, subjects were instructed to walk as fast as possible without running. Trials were accepted when the foot hit either forceplate in isolation. A minimum number of five successful gait cycles were collected for each subject's prosthetic side.

Data were processed in Visual 3D 3.79 (C-Motion, Inc., Rockville, MD) using either cadaver data anthropometrics or those calculated from direct measurements. Hip, knee and ankle joint moments were calculated in the anatomical frame and normalized to body weight including the prosthesis. Gait cycles were averaged for preferred and fast walking conditions. Peak moments for the hip, knee and ankle for selected points in the gait cycle were compared using the two different methods. Paired t-tests were computed (for mean peak moments) to determine if the differences were statistically significant ($p < 0.05$).

3. Results

Important differences in anthropometrics were found using the two different methodologies (Table 2). Foot segment MIs were substantially larger using direct measurements, while those of the shank segments were smaller. Thigh segment MIs were smaller using direct measurements for subjects 1 and 2 and larger for subjects 3 and 4.

Exemplar data are presented for subject 1 (Fig. 2), and peak values for all subjects are presented in Table 3. Stance hip extensor and flexor moments using the two different methods varied slightly for both walking speeds. Mean peak moments, however, were not significantly different. During swing, peak hip moments were significantly greater using the cadaver estimate method for both walking speeds ($p < 0.05$).

Mean peak knee flexor moments during stance were similar for both speeds, and a statistical difference was not

Table 2
Moment of inertia (about the proximal end), center of gravity and segment mass using direct measurements and cadaver data for the thigh, shank and foot

	Thigh		Shank		Foot	
	Direct	Dempster	Direct	Dempster	Direct	Dempster
Subject 1						
Moment of inertia (kg m^2)	0.0640	0.0799	0.0147	0.0376	0.0053	0.0005
Center of gravity (% from proximal)	46.9	43.3	20.3	43.3	21.1	42.9
Segment mass (kg)	2.22	6.75	1.08	3.14	0.88	0.98
Subject 2						
Moment of inertia (kg m^2)	0.0366	0.0854	0.0082	0.0402	0.0035	0.0006
Center of gravity (% from proximal)	27.8	43.3	45.5	43.3	16.2	42.9
Segment mass (kg)	4.29	7.16	0.57	3.33	0.69	1.04
Subject 3						
Moment of inertia (kg m^2)	0.0936	0.0577	0.0082	0.0272	0.0029	0.0004
Center of gravity (% from proximal)	32.5	43.3	22.3	43.3	19.5	42.9
Segment mass (kg)	5.25	5.55	0.53	2.58	0.57	0.80
Subject 4						
Moment of inertia (kg m^2)	0.0301	0.0260	0.0062	0.0122	0.0025	0.0002
Center of gravity (% from proximal)	43.8	43.3	27.2	43.3	22.0	42.9
Segment mass (kg)	1.61	3.26	0.52	1.51	0.39	0.47

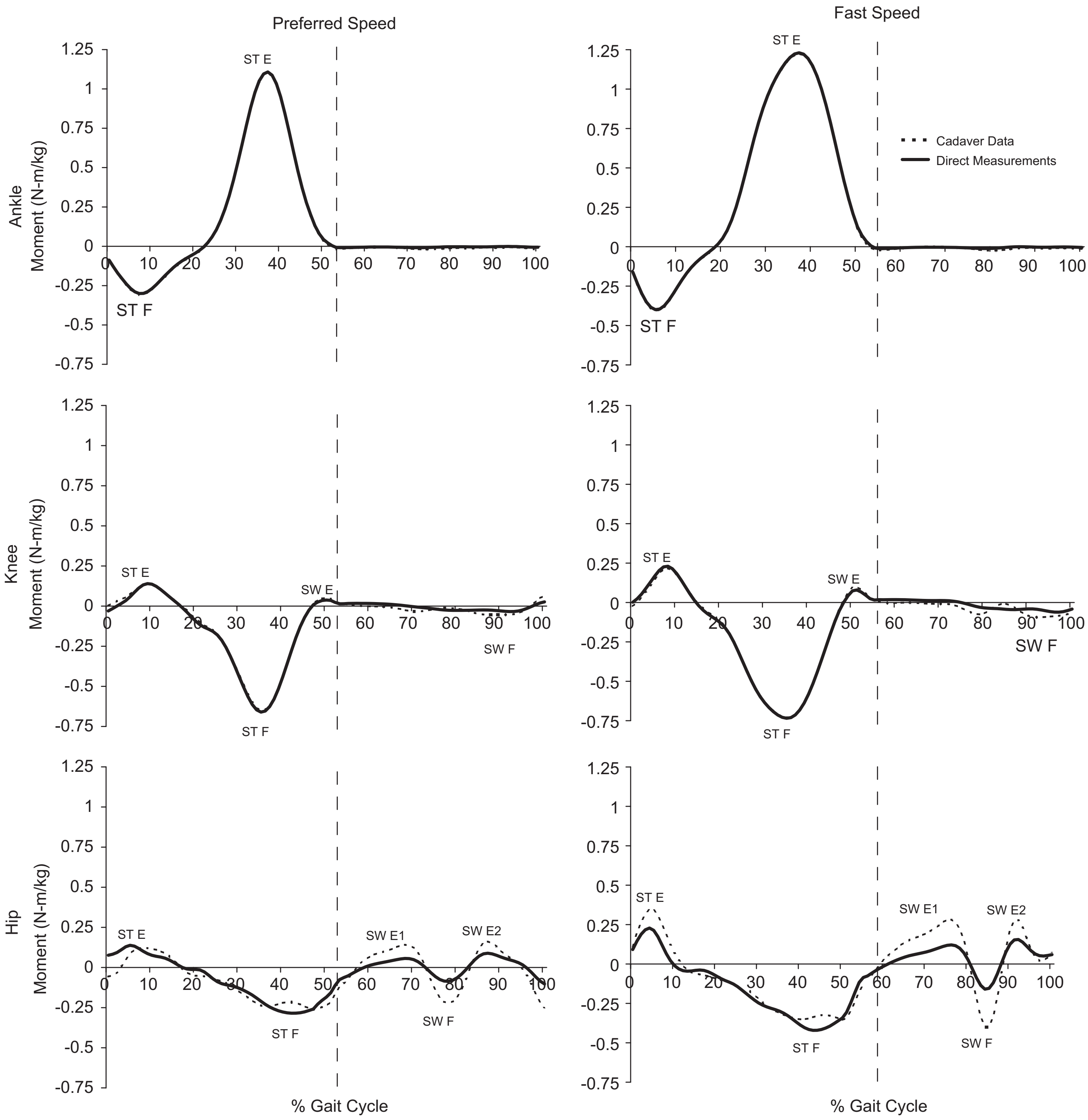

Fig. 2. Exemplar joint moment data calculated with cadaver data and direct measurements from one subject walking at preferred and fast speeds. The joint moment peaks that were selected for statistical analysis are labeled stance (ST), swing (SW), flexion (F) and extension (E). The dashed vertical line represents toe-off for this subject. Positive indicates ankle plantar flexion, knee extension and hip extension.

found between the two different methods (Table 3). During swing, mean peak knee extensor moments (SW E) were not significantly different for the preferred speed, but were significantly different for the fast speed ($p<0.05$). Mean peak knee flexor moments during swing (SW F) were significantly different between the two methodologies at both speeds. When differences were observed, peaks calculated using cadaver data were greater in magnitude (see Fig. 2).

Mean peak dorsiflexion (ST E) and plantar flexion moments (ST F) were significantly greater ($p<0.05$) using the direct method (Table 3), but the difference between the average values was assumed to be negligible (<0.01 Nm/kg).

Table 3
Average peak joint moments across all subjects throughout the gait cycle

	Preferred Speed			Fast Speed		
	Direct	Dempster	Difference	Direct	Dempster	Difference
Peak ankle joint moment (N m/kg)						
ST F	-0.25 ± 0.07	-0.26 ± 0.07	$0.01\pm0.00^*$	-0.26 ± 0.14	-0.26 ± 0.14	$0.01\pm0.00^*$
ST E	1.13 ± 0.16	1.12 ± 0.16	$0.01\pm0.00^*$	1.28 ± 0.21	1.27 ± 0.21	$0.01\pm0.00^*$
Peak knee joint moment (N m/kg)						
ST F	0.51 ± 0.18	0.51 ± 0.17	0.00 ± 0.01	0.59 ± 0.21	0.60 ± 0.20	0.00 ± 0.01
SW E	-0.08 ± 0.06	-0.10 ± 0.08	0.02 ± 0.03	-0.15 ± 0.07	-0.19 ± 0.08	$0.04\pm0.02^*$
SW F	0.06 ± 0.03	0.12 ± 0.05	$-0.06\pm0.03^*$	0.08 ± 0.02	0.16 ± 0.05	$-0.08\pm0.04^*$
Peak hip joint moment (N m/kg)						
ST E	0.21 ± 0.22	0.21 ± 0.20	0.00 ± 0.10	0.41 ± 0.30	0.51 ± 0.30	-0.09 ± 0.15
ST F	-0.46 ± 0.20	-0.49 ± 0.28	0.03 ± 0.13	-0.69 ± 0.25	-0.77 ± 0.40	0.09 ± 0.19
SW E1	0.04 ± 0.01	0.13 ± 0.02	$-0.09\pm0.03^*$	0.09 ± 0.03	0.25 ± 0.04	$-0.16\pm0.04^*$
SW F	-0.09 ± 0.05	-0.25 ± 0.14	$0.16\pm0.10^*$	-0.14 ± 0.08	-0.36 ± 0.21	$0.22\pm0.13^*$
SW E2	0.13 ± 0.05	0.29 ± 0.11	$-0.16\pm0.07^*$	0.17 ± 0.02	0.38 ± 0.09	$-0.20\pm0.08^*$

$^*p<0.05$, ST = stance, SW = swing, E = extension, F = flexion.

4. Discussion

The objective of this study was to compare the effect anthropometry has on the joint moments of children with AK prostheses. Direct measurement of the MI calculations varied greatly between subjects and as compared to data calculated from cadaver studies. Significant differences in joint moment peaks were found in hip and knee joint flexor and extensor moments during swing. As suggested in previous literature (Miller, 1987; Czerniecki et al., 1991), differences in anthropometric data had little effect on stance phase hip and knee peaks. The change in anthropometrics affects the inertia couple ($I\alpha$ term) and the inertial effect due to gravity, which make up a small percentage of the joint moment during stance (Miller, 1987). During swing, the ground-reaction force, which contributed substantially to hip and knee joint moments in stance, were absent. In addition, hip and knee angular accelerations were greater during swing. As a result, the inertia couple and gravitational component accounted for a greater percentage of the moment. For example, in subject 1, the inertia couple made up $<10\%$ of the hip extension moment for 65% of stance, while the gravitational component made up $<10\%$ of the moment for 97% of stance. In contrast, the inertia couple made up $>50\%$ of the hip extension moment for 96% of swing, and the gravitational component made up $>50\%$ of the moment for 22% of swing.

One limitation of this study is that repeatability of circumference measurements, which has been shown to affect measurement accuracy (Geil, 2005), was not assessed. Another limitation is the small sample size. This study was limited to pediatric subjects with a wide range of body weight and height (Table 1). Discrepancies may have less of an effect on the results of adults. Despite these limitations, we found that there were large variations between conditions during swing. These prostheses were fairly easy to dissemble, and the entire process of direct prosthetic measurement took approximately an hour per patient. Other prostheses, however, can be more difficult to disassemble and reassemble. In particular, an energy storage and return prosthesis requires cutting to separate the foot and shank segments and cannot be reassembled. Regression equations to determine anthropometrics of varying types and sizes of prosthetic segments would be an important contribution to the literature. Anthropometrics based on regression equations using prostheses would improve accuracy and prevent discrepancies in joint kinetics.

In summary, the results of this study indicate that it may be reasonable to use the anthropometrics from cadaver studies instead of directly calculating the anthropometrics for each prosthetic part when analyzing kinetics during stance. However, this approximation would not accurately assess kinetics during swing as seen by the artificial hip and knee moment peaks created when using cadaver data for moment calculations. Future studies should focus on creating regression equations to more accurately determine anthropometrics.

Conflict of interest

None declared.

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