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Identification of Trunk and Pelvis Movement Compensations in Patients with Transtibial Amputation using Angular Momentum Separation

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1. Introduction

Over one million Americans currently have a lower-limb amputation, and this number is projected to double by 2050 [1] due to dysvascular pathologies (e.g. diabetes mellitus (DM)) [2]. Patients with dysvascular amputation commonly have multiple comorbidities and 40-50% have limited physical function [3], which require different treatments apart from patients with traumatic amputation. Although patients with dysvascular amputation differ in age, BMI, prosthetic use time, and comorbidities from patients with traumatic amputation [3,4], it is common to combine them into a single group when investigating how amputation affects functional movement characteristics [5,6]. Because patients with DM prior to amputation move differently than healthy controls [7], differences in movement compensations between patients with dysvascular amputation to patients with DM alone could be used as physical rehabilitation targets for movement retraining following amputation.

Patients with unilateral transtibial amputation (TTA) are at increased risk of developing low back pain (LBP) [8], which may relate to necessary movement compensations to achieve forward progression and balance during walking. For example, to accomplish forward progression in the absence of an ankle plantar flexor, patients with unilateral TTA increase hip extensor power during the stance period of the residual limb [9]. Patients with unilateral TTA demonstrate exaggerated lateral trunk lean toward the amputated limb (compensated Trendelenburg) [10] and altered foot placement of the intact limb, which leads to uneven step length, swing time, and stance time [9]. While these compensations may be necessary to accomplish mobility, asymmetric movements are linked to the development of LBP [11]. This coordination of excessive trunk and pelvic motion during walking likely contributes to

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step-to-step asymmetric loading at the low back previously measured in patients with unilateral TTA [12], and may increase the risk of developing LBP, which was previously demonstrated in patients with transfemoral amputation [13,14].

Clinicians rely on observational gait analysis to identify movement compensations which is highly subjective and unreliable for identifying consequential movement compensations in amputees [15]. Although laboratory-based gait analysis is valid and reliable for quantitatively measuring movement, it is accompanied by high computational and economic expenses, and currently impractical in the vast majority of clinical settings. Because clinicians use observational gait analysis to guide interventions and gait retraining in patients with unilateral TTA, the ability to obtain accurate measures of trunk and pelvis movement patterns could help tailor treatment to patients and ultimately prevent injuries, such as LBP.

Identification of segmental strategies used to generate and arrest segmental angular momentum can provide insight into muscle demands following unilateral dysvascular TTA. During walking, muscles are used concentrically and eccentrically as the primary mechanisms to generate and arrest segment angular momentum [16]. Measuring and understanding segmental angular momentum is a promising approach to bridge the gap between observational and quantitative gait analysis. We previously demonstrate a framework to describe clinical movement compensations during gait using separation of translational angular momentum referenced to the stance foot [17]. Total segmental angular momentum can be separated into two components, each with a unique interpretation: 1) Translational Angular Momentum (TAM): angular momentum created by linear velocity of the segment with mass with respect to a point and 2) Rotational Angular Momentum (RAM): angular momentum created by the rotational velocity of an object with inertia [18].

The objective of this investigation was to assess movement compensations in patients with unilateral dysvascular TTA and patients with DM by examining translational angular momentum and rotational angular momentum of the trunk and pelvis during walking for patterns of generating/arresting momentum. We hypothesized that patients with unilateral dysvascular TTA, patients with DM, and healthy control participants would demonstrate similar patterns of generating/arresting TAM of the trunk and pelvis when walking at similar speeds. We also hypothesized that patients with unilateral dysvascular TTA would demonstrate higher RAM of the trunk and pelvis than the other groups, which illustrates potentially consequential movement compensations that can be retrained through clinical intervention.

2. Methods

2.1 Participants

Ten patients with DM and unilateral TTA 1-3 years post amputation (AMP) (Table 1) (10 M; age: 56.8 ± 4.3 years; mass: 97.6 ± 15.2 kg; height: 1.8 ± 0.1 m), 11 patients with DM (2F, 9 M; age: 61.4 ± 8.0 years; mass: 94.3 ± 22.0 kg; height: 1.7 ± 0.1 m), and 13 healthy control patients (HC) (3 F, 10 M; age: 63.1 ± 7.7 years; mass: 77.7 ± 13.2 kg; height: 1.7 ± 0.1 m) were enrolled. Eligibility criteria included: age: 50-85 years; BMI ≤ 40 kg/m^2;

independent community ambulation (ability to walk for four minutes without rest or assistive device); 1-3 years post amputation (AMP group); controlled Type-II diabetes mellitus (AMP and DM groups); no traumatic or cancer-related amputation (AMP group); no major amputation on contralateral limb (AMP group); no cardiovascular, orthopaedic, neurologic, wounds, or ulcers that limit physical function; no history of LBP (HC group); no diagnosed rheumatoid arthritis (HC group); no diagnosed osteoarthritis (HC group); and no total hip/knee joint arthroplasty (HC group). Each participant provided a written, informed consent in accordance with the Colorado Multiple Institutional Review Board prior to the start of the experimental session and completed one data collection in which whole body kinematics were collected.

2.2 Motion Analysis

Each participant was instrumented with 63 reflective markers used to obtain whole-body kinematics during gait. Motion was recorded from eight infrared cameras (Vicon) sampled at 100 Hz. Each participant performed three gait trials at 1.0 m/s (± 0.05 m/s) on a 10-m walkway. Motions were averaged across the three trials and used for group comparisons.

2.3 Data Analysis

Kinematic data were low-pass filtered with a 4th-order Butterworth filter (6 Hz cutoff frequency). A 15-segment subject-specific model (head, upper arms, forearms, hands, trunk, pelvis, thighs, shanks, and feet) was created in Visual 3D (C-Motion, Inc.). Segment masses were based on a percentage of total body weight and segment inertias were based on segment geometry [19]. For the AMP group, mass the center of mass position, and inertial properties of the prosthetic shank (residual limb + prosthetic socket) and prosthetic foot were determined using a reaction board technique and oscillation method [20].

TAM (angular momentum of a segment with respect to the stance foot) is described as:

$$\mathbf{h}_{i/\mathrm{Foot}}=(\mathbf{r}_i-\mathbf{r}_{\mathrm{Foot}})\times m_i(\mathbf{v}_i-\mathbf{v}_{\mathrm{Foot}}) \quad (1)$$

where $\mathbf{r}_i$ and $\mathbf{r}_{\mathrm{Foot}}$ are the position vectors of the i^{th} segment and foot, respectively, m_i is the mass of the i^{th} segment, and $\mathbf{v}_i$ and $\mathbf{v}_{\mathrm{Foot}}$ are the velocities of the i^{th} segment and foot respectively. RAM (angular moment of a segment with respect to its center of mass) is described as:

$$\mathbf{h}_i=\mathbf{I}_i\cdot\boldsymbol{\omega}_i \quad (2)$$

where $\mathbf{I}_i$ is the moment of inertia tensor and ω_i is the angular velocity of the segment. To facilitate planar analyses, all angular momenta vectors were expressed in a path reference frame, that is defined by the velocity vector of the body COM: $\mathbf{e}_{\mathrm{frontal}}$ (tangent to the horizontal path of the body COM), $\mathbf{e}_{\mathrm{transverse}}$ (opposite direction of the gravity vector), and $\mathbf{e}_{\mathrm{sagittal}}$ ($\mathbf{e}_{\mathrm{frontal}} \times \mathbf{e}_{\mathrm{transverse}}$). Within the path reference frame, positive momenta values in each plane are defined as: sagittal – posterior rotation away from stance foot, frontal – medial-lateral rotation toward stance foot, transverse – rotation away from stance foot.

2.4 Statistical Analysis

Patient anthropometrics (mass and height) were compared across groups using a one-way ANOVA followed by Tukey HSD for post hoc comparison ($\alpha = 0.05$).

All momenta were calculated during one gait cycle (AMP: amputated limb heel strike to amputated limb heel strike; DM and HC: right heel strike to right heel strike). TAM ($\mathbf{h}_{i/\mathrm{Foot}}$), was calculated with respect to the stance foot was analyzed during the stance period. RAM ($\mathbf{h}_i$) was analyzed during the entire gait cycle. To quantify generation and arresting of trunk and pelvis angular momentum, global minimums and maximums were determined.

Magnitudes of the global minima and maxima in each segmental angular momentum variable (TAM and RAM) were compared across the three groups using an ANCOVA (covariates: mass and height) followed by pairwise comparisons using Tukey HSD (α = 0.05). Qualitative analysis was performed to assess when peak momenta values occurred throughout the functional phases of gait: weight acceptance (0-12%), single limb support (12-50%), swing limb advancement (50-100%) [21].

3. Results

3.1 Patient Anthropometrics

Body mass was larger in the AMP group than the HC group ($P = 0.03$). No differences in height existed across groups.

3.2 Translational Angular Momentum

In the sagittal plane, peak posterior trunk and pelvis TAM was lower in the AMP group than the DM group (trunk: $P = 0.01$, pelvis: $P = 0.01$) at the end of single limb support (Figure 1a, Table 1). In the sagittal plane, peak anterior trunk TAM was lower in the DM group than the HC group ($P = 0.03$) at the beginning of single limb support (Figure 1a, Table 2).

In the frontal plane, peak lateral trunk TAM toward the stance foot was lower in the AMP group than the DM group ($P < 0.001$) during weight acceptance (Figure 1a, Table 2).

In the transverse plane, peak trunk and pelvis TAM toward the stance foot was higher in the AMP group than the DM group (trunk: $P = 0.03$, pelvis: $P = 0.01$) at the beginning of single limb support (Figure 1a, Table 2). Peak pelvis TAM away from the stance foot was lower in the AMP group than both the DM group ($P < 0.001$) and the HC group ($P < 0.001$) at the end of single limb support (Figure 1a, Table 2). All other comparisons were not statistically significant.

3.3 Rotational Angular Momentum

In the sagittal plane, peak anterior trunk RAM was higher in the AMP group than both the DM group ($P = 0.02$) and HC group (P = 0.01) at the beginning of single limb support (Figure 2a, Table 3). Peak posterior trunk RAM was lower in the AMP group than both the DM group ($P = 0.04$) and HC group ($P = 0.05$) at the beginning of swing limb advancement (Figure 2a, Table 2). Peak anterior pelvis RAM was higher in the AMP group than both the

DM group (P = 0.04) and the HC group (P = 0.04) at the beginning of single limb support (Figure 2b, Table 3).

In the frontal plane, peak lateral trunk RAM toward the stance foot was higher in the AMP group than the DM group (P = 0.04) during swing limb advancement (Figure 2a, Table 3).

In the transverse plane, peak pelvis RAM toward the stance foot was higher for the AMP group than both the DM group (P = 0.02) and the HC group (P = 0.03) at the beginning of single limb support (Figure 2b, Table 3). All other comparisons were not statistically significant.

4. Discussion

The objective of this investigation was to identify and compare movement patterns in patients with dysvascular transtibial amputation (AMP), patients with diabetes mellitus (DM), and healthy controls (HC) using patterns of generating and arresting trunk and pelvis angular momentum during walking. We observed differences in translational angular momentum in all three planes between the AMP, DM, and HC groups, which indicates unique movement patterns adopted by each group during walking. Loss of ankle function in the AMP group is linked to different movement compensations, and results in higher generation of trunk and pelvis RAM in all three planes compared to the DM and HC groups. Large trunk angular momentum with small pelvis momentum is a compensation in the AMP group that may result in high paraspinal muscle demand, which leads leading to LBP. The identification of movement compensations through analysis of segmental RAM has potential important clinical applications in a gait retraining setting through wearable sensors.

Patterns of trunk and pelvis TAM indicate the use of a postural compensation by the AMP group to maintain balance and achieve forward progression without ankle function. TAM is a function of position and linear momentum of each segment relative to the stance foot (Eq. 1). In the sagittal plane, trunk and pelvis anterior TAM is generated about the stance foot during weight acceptance, is slightly arrested throughout single limb support, and then arrested completely at the transition to swing limb advancement (Figure 1). Without active plantar flexion at the end of single limb support, the AMP group generated smaller posterior angular momentum when compared to the DM group, which is adopted to maintain forward progression when unloading the amputated limb. In the frontal plane, trunk and pelvis TAM toward the stance limb is rapidly arrested during loading response and then is gradually arrested throughout the remainder of single limb support until angular momentum is generated away from the stance limb during the preparation of swing limb advancement as weight is transferred between limbs. In the transverse plane, trunk and pelvis TAM were arrested during loading response and then remained constant throughout the duration of single limb stance. Remarkably, trunk and pelvis TAM at initial foot contact in the AMP group were directed toward the stance (amputated) limb, which is opposite of both the HC and DM groups. This difference is likely a result of excessive propulsion by the intact limb, which creates a transverse rotation toward the amputated limb. Because each group walked at the same speed, the large transverse TAM toward the stance foot throughout the duration of single limb support in the AMP group occurs by a more medial position of the segment with respect to the stance foot. In the frontal plane, this corresponds to a wider step width, which is a commonly observed finding in amputee gait [9].

Segment rotational angular momentum provides a unique framework for identifying differences in movement patterns by highlighting the motion of the segment, which can assist in characterizing and interpreting movement compensations observed in the clinic. In the sagittal plane, large anterior rotational angular momentum in the AMP group leads to a forward trunk lean that is frequently observed during single limb support, and represents an adaptive strategy to maintain forward progression in light of ankle plantar flexor loss [22]. The hip and trunk extensor demands needed to arrest the large anterior trunk rotational angular momentum, which occurs at approximately 10% of the gait cycle, may contribute to overuse injuries in the lower extremity and the low back [23,24].

In the frontal plane, the AMP group generated larger trunk RAM toward the amputated limb during weight acceptance, and arrested trunk angular momentum later in the gait cycle, compared to the DM and HC groups, corresponding to large trunk displacement toward the stance limb. To prevent a fall at this point in the gait cycle, the AMP group must quickly arrest a large amount of angular momentum that has been generated in the trunk, which creates high paraspinal muscle demand [25]. The lateral trunk posture over the stance limb (compensated Trendelenburg posture) corresponds with increased loading in the low back, and is linked to the development of LBP [26].

In the transverse plane, the AMP group generated substantially larger pelvis RAM toward the amputated than the HC and DM groups limb during weight acceptance. This large RAM, due to excessive angular speed, may be linked to the large ankle power in the intact limb needed to achieve forward progression [27]. Therefore, pelvis RAM must be arrested following peak generation to maintain balance during swing and continue progression during gait.

Our results indicate that the AMP group generates and arrests trunk and pelvis momenta differently than either the DM or HC groups, and the associated muscle demands with the observed movement patterns after amputation may be linked to LBP [8,12]. Because a patient with unilateral TTA cannot create propulsive ankle joint moments, the generating demands shifts higher in the kinetic change. Our results show that movement compensations occur in the pelvis and trunk, and indicates that demand on local muscles (e.g. multifidus, erector spinae, obliques, etc.) is likely higher for patients with amputation than other populations, which potentially could be consequential in the development for LBP.

Identifying movement compensations using segmental RAM may have important clinical applications. RAM combines inertia with angular speed of the segment, which is a parameter that can be interpreted in light of the effort needed to generate or arrest the measured momentum. Because observational analysis is based on presence of events and postures (e.g. compensated Trendelenburg sign), a clinician can gain insight into the effort needed to accomplish the observed event by supplementing with angular momentum. Measuring rotational angular momentum is easily facilitated by wearable sensors such as gyroscopes, and would not require additional instrumentation (e.g. force platforms). Use of low-cost wearable sensors have emerged in biomechanics that facilitate spatiotemporal gait characteristics [28,29] as well as segment and joint kinematics [30]. With additional research, angular momentum may provide clinicians and patients with immediate and accurate information on their ability understand when movement compensations occur and increase the efficacy of targeted movement retraining following amputation.

Several limitations should be considered. First, the analysis did not consider consecutive gait cycles; therefore, repeatability of movement compensations was not characterized in these measures. In future investigations we will extend this analysis using repeated over ground trials or a treadmill. Second, we do not know how segmental angular momentum variables correspond with traditional biomechanical variables. In future investigations we will associate how these movement compensations correlate with traditional quantitative biomechanical analyses (e.g. joint moments, joint loading, etc.). Third, neither the AMP and DM groups were screened for LBP at the time of testing; therefore, we cannot determine if any compensatory movement patterns adopted by each group were a habitual movement pattern or a result of LBP. Finally, because only patients with dysvascular amputation were included, we are unable to generalize these findings to patients with unilateral TTA from other causes other than dysvascular disease.

5. Conclusion

This investigation demonstrated the use of segmental angular momenta to identify movement compensations in the trunk and pelvis in patients with unilateral dysvascular transtibial amputation. Coordinated compensations between the trunk and pelvis promote forward progression during locomotion, but may have long-term adverse effects from the demand placed on the musculoskeletal system to generate and arrest segmental momentum.

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Research Highlights

- Examination of transtibial amputees through angular momentum of pelvis and trunk.
- Inference on motion and effort used to supplement observational analyses in clinic.
- Movement compensations linked to consequential muscle demand.

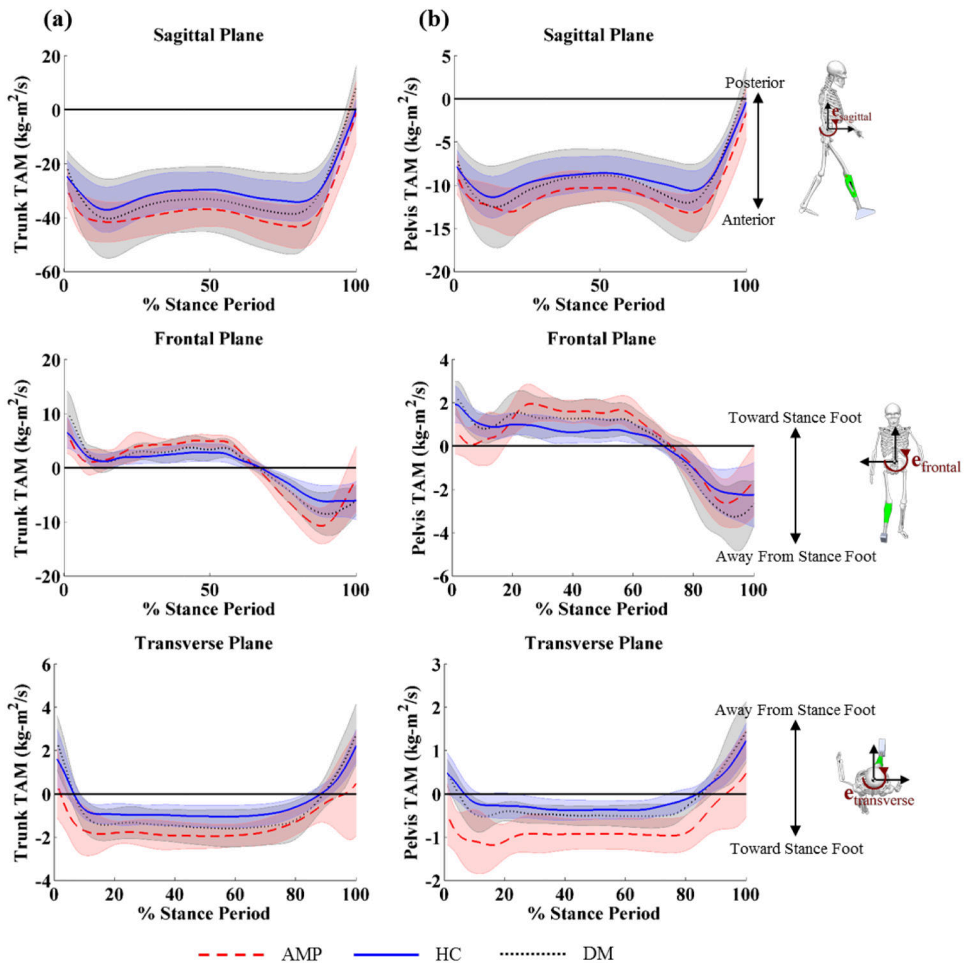

Figure 1.
Translational angular momentum (TAM) of the (a) trunk and (b) pelvis with respect to the stance foot in the sagittal, frontal, and transverse plane healthy controls (blue solid line), patients with diabetes mellitus (DM) (black dotted line), and patients with DM and transtibial amputation (AMP) (red dashed line).

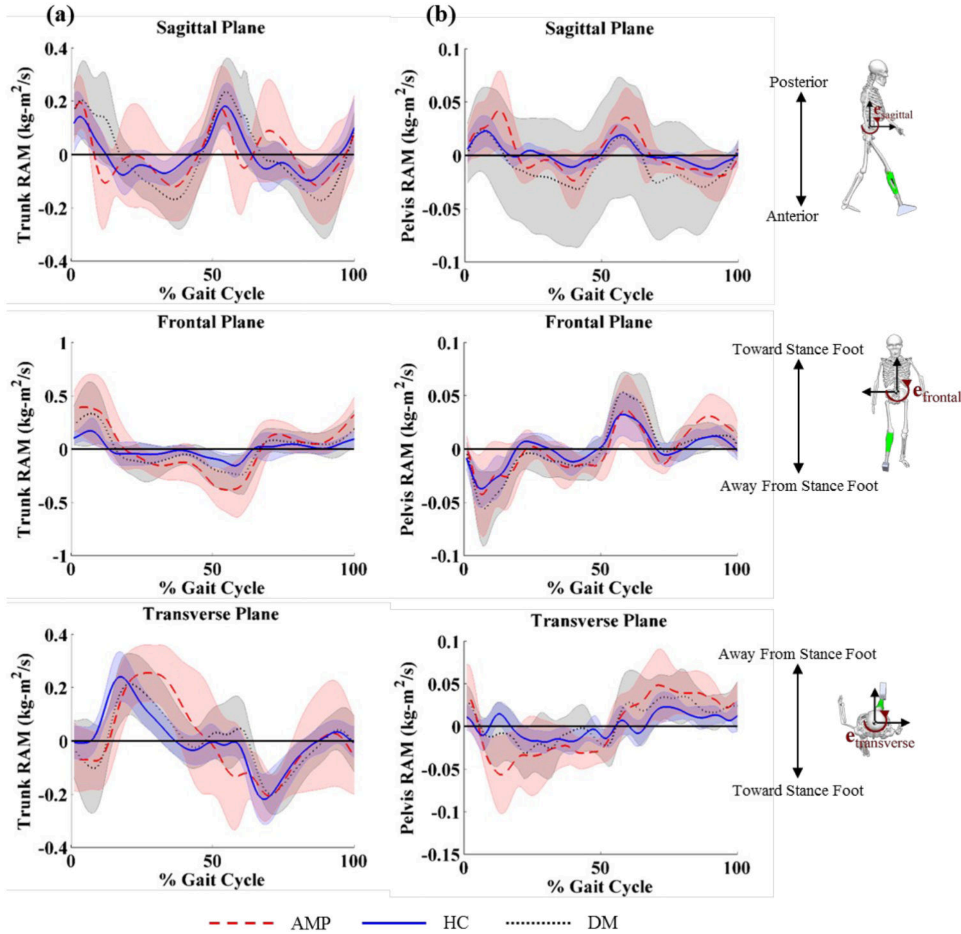

Figure 2.

Rotational angular momentum (RAM) of the (a) trunk and (b) pelvis with respect to the stance foot in the sagittal, frontal, and transverse plane healthy controls (blue solid line), patients with diabetes mellitus (DM) (black dotted line), and patients with DM and transtibial amputation (AMP) (red dashed line).

Table 1

Participant characteristics for patients with dysvascular unilateral transtibial amputation (AMP) group.

Time since Amputation (Months)	Residual Limb Length (cm)	Socket Type	Prosthetic Foot
17.4 ± 5.1	14.8 ± 2.5	Total contact carbon fiber	Dynamic elastic response

Table 2

Mean ± SD peak (minimum and maximum) translational angular momentum (TAM) of the trunk and pelvis ($\mathbf{h}_{Trunk/Foot}$ and $\mathbf{h}_{Pelvis/Foot}$) during the stance period for patients with dysvascular transtibial amputation (AMP), diabetes mellitus (DM), and healthy control (HC) groups.

		Trunk TAM (kg·m^2/s)			Pelvis TAM (kg·m^2/s)		
		AMP	**DM**	**HC**	**AMP**	**DM**	**HC**
Sagittal Plane	*Min (Anterior)*	-45.6 ± 9.0	-40.7 ± 14.7	-37.3 ± 8.3	-14.5 ± 2.7	-12.7 ± 5.8	-11.5 ± 2.4
	Max (Posterior)	-0.9 ± 10.8	8.4 ± 7.8	0.5 ± 3.3	-1.6 ± 2.8	1.3 ± 2.4	-0.4 ± 1.0
Frontal Plane	*Min (Away from Stance Foot)*	-11.5 ± 3.1	-9.5 ± 4.2	-8.6 ± 2.4	-3.2 ± 1.1	-3.6 ± 2.0	-3.0 ± 0.9
	Max (Toward Stance Foot)	8.1 ± 2.2	10.2 ± 4.2	7.6 ± 4.0	2.6 ± 0.7	2.7 ± 1.2	2.3 ± 1.1
Transverse Plane	*Min (Toward Stance Foot)*	-2.3 ± 0.9	-1.8 ± 0.9	-1.2 ± 0.5	-1.4 ± 0.6	-0.8 ± 0.4	-0.4 ± 0.3
	Max (Away from Stance Foot)	1.8 ± 1.7	3.1 ± 1.4	2.5 ± 1.0	0.8 ± 0.7	1.5 ± 0.9	1.2 ± 0.4

Table 3

Mean ± SD peak (minimum and maximum) rotational angular momentum (RAM) of the trunk and pelvis ($\mathbf{h}_{\mathrm{Trunk}}$ and $\mathbf{h}_{\mathrm{Pelvis}}$) during the gait cycle for patients with dysvascular amputation (AMP), diabetes mellitus (DM), and healthy control (HC) groups.

		Trunk RAM (kg·m^2/s)			Pelvis RAM (kg·m^2/s)		
		AMP	**DM**	**HC**	**AMP**	**DM**	**HC**
Sagittal Plane	*Min (Anterior)*	-0.34 ± 0.12	-0.24 ± 0.12	-0.16 ± 0.05	-0.05 ± 0.02	-0.03 ± 0.02	-0.02 ± 0.01
	Max (Posterior)	0.39 ± 0.12	0.33 ± 0.15	0.25 ± 0.08	0.07 ± 0.04	0.05 ± 0.03	0.03 ± 0.01
Frontal Plane	*Min (Away from Stance Foot)*	-0.47 ± 0.22	-0.33 ± 0.21	-0.20 ± 0.08	-0.07 ± 0.05	-0.06 ± 0.04	-0.04 ± 0.01
	Max (Toward Stance Foot)	0.47 ± 0.27	0.37 ± 0.29	0.22 ± 0.10	0.06 ± 0.03	0.07 ± 0.03	0.04 ± 0.01
Transverse Plane	*Min (Toward Stance Foot)*	-0.33 ± 0.11	-0.26 ± 0.11	-0.24 ± 0.09	-0.08 ± 0.04	-0.06 ± 0.03	-0.04 ± 0.01
	Max (Away from Stance Foot)	0.34 ± 0.13	0.27 ± 0.12	0.26 ± 0.09	0.08 ± 0.04	0.07 ± 0.04	0.04 ± 0.01