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Trunk and pelvic dynamics during transient turns among individuals with unilateral traumatic lower limb amputation

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ABSTRACT

Prior work has identified alterations in trunk-pelvic dynamics with lower limb amputation (LLA) during in-line walking; however, evaluations of other ambulatory tasks are limited. Turns are ubiquitous in daily life but can be challenging for individuals with LLA, prompting additional or unique proximal compensations when changing direction, which over time may lead to development of low back pain. We hypothesized such proximal kinematic differences between persons with and without LLA would exist in the sagittal and frontal planes. Three-dimensional trunk and pelvic kinematics, translational and rotational momenta, and coordination phase/variability were compared among eight persons with unilateral LLA (4 with transfemoral amputation and 4 with transtibial amputation), and five uninjured controls, who performed 90-degree turns to the left ($n = 10$) and right ($n = 10$). Participants self-selected the turn strategy (i.e., step vs. spin) and pivot limb in response to verbal cues regarding when and which direction to turn. Coordination variability and translational angular momenta did not differ between groups in either turn type. During spin turns, frontal rotational angular momenta were larger and frontal trunk-pelvis range of motion was smaller among persons with vs. without LLA. During step turns, pelvis leading transverse coordination was more frequent, frontal trunk rotational angular momentum was smaller, and sagittal pelvis range of motion was larger among persons with vs. without LLA. Altered and task-dependent modulation of trunk-pelvic dynamics among persons with LLA provides additional support for a potential link between repeated exposures to altered trunk-pelvic dynamics with elevated low back pain risk.

1. Introduction

Persons with lower limb amputation (LLA) often walk with compensatory movement strategies involving a prominent reliance on the trunk and pelvis (Goujon-Pillet, Sapin, Fodé, & Lavaste, 2008). Altered kinematic features and coordination of these two segments have been associated with elevated demands on the low back (Hendershot & Wolf, 2014), increased inter-segmental rigidity (Russell Esposito & Wilken, 2014), and larger trunk muscular forces and spinal loads (Shojaie, Hendershot, Wolf, & Bazrgari, 2016; Yoder, Petrella, & Silverman, 2015). These altered loads and asymmetric trunk-pelvis kinematics among persons with LLA have been suggested as key factors in disc degeneration and passive ligamentous strain potentially leading to development of low back pain (LBP; Devan, Hendrick, Ribeiro, Hale, & Carman, 2014; Gailey, Allen, Castles, Kucharik, & Roeder, 2008). As such, differences in trunk/pelvis kinematics between persons with and without LLA have been characterized during in-line walking (Goujon-Pillet et al.,

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2008; Hendershot & Wolf, 2014; Morgenroth et al., 2010). Yet, in-line walking is but one movement among many required for functional independence. Thus, characterizing the extent to which persons with LLA utilize proximal compensations during other (perhaps more demanding) tasks/activities of daily living would facilitate a more comprehensive understanding of biomechanical contributors to LBP risk.

Transient (i.e., non-steady-state) tasks embedded within in-line walking are ubiquitous and often necessary to adequately navigate an environment. Turns, in particular, account for approximately half of daily steps (Glaister, Bernatz, Klute, & Orendurff, 2007; Sedgeman, Goldie, & Iansek, 1994). Biomechanically, turns require a redirection of the body's center of mass, typically as a change in direction between 76 and 120 degrees (Sedgeman et al., 1994) executed using either a step (turn direction is contralateral to pivot leg) or spin strategy (turn direction is ipsilateral to pivot leg; Taylor, Dabnichki, & Strike, 2005). Among persons with LLA, compromised ankle function alters control of braking/propulsive and mediolateral forces during a turn (albeit along a circular vs. orthogonal path; Segal, Orendurff, Czerniecki, Shofer, & Klute, 2008; Ventura, Segal, Klute, & Neptune, 2011), thereby likely necessitating proximal adaptations of the trunk/pelvis to adequately redirect the body's center of mass. Furthermore, proximal compensations during turns may also exist to minimize discomfort within the residual limb-socket interface, particularly as it relates to torsion/shear (Heitzmann et al., 2015).

Inter-segmental coordination and momentum have been used for identification of compensational movement strategies during ambulation. For example, persons with unilateral LLA generate and arrest larger trunk and pelvic segmental momenta during walking (Gaffney, Murray, Christiansen, & Davidson, 2016), as well as alter segmental coordination strategies dependent on the presence of current LBP (Russell Esposito & Wilken, 2014). While recent efforts have similarly identified altered trunk-pelvic coordination strategies in able-bodied individuals (with and without LBP) executing turns (Smith & Kulig, 2016), there exist no studies specifically focused on trunk and pelvic compensations during turns among persons with LLA. Thus, the primary purpose of this study was to characterize proximal compensations using inter-segmental momenta and coordination during transient (90-degree) turns among persons with LLA. Although turns are predominantly associated with movement in the transverse plane, it was hypothesized that persons with vs. without LLA execute turns with altered trunk-pelvic segmental coordination, particularly in the sagittal and frontal planes, to overcome the aforementioned challenges associated with modulating braking/propulsive and mediolateral forces with altered ankle function. Secondarily, we hypothesized that such alterations in trunk-pelvic coordination would also be associated with larger ranges of segmental momenta among persons with vs. without LLA.

2. Methods

2.1. Participants

Eight persons with unilateral LLA of traumatic etiology (four with transtibial amputation [TTA], three with transfemoral amputation, and one with knee disarticulation [TFA]) and five persons without LLA (uninjured controls; CTRL) completed this study (Table 1). All participants provided informed consent approved by the Walter Reed National Military Medical Center Institutional Review Board. All participants were free of neurological and orthopaedic injury aside from lower limb amputation, were able to ambulate over even terrain without an assistive device, and were not experiencing any moderate or severe discomfort/pain, regardless of cause, at any point during data collection, as measured by overall pain scores less than 4 cm on a 10 cm Visual Analog Scale (Jensen, Chen, & Brugger, 2003). Of the persons with TTA, 2 wore the RUSH and 2 wore the Vari-Flex XC foot. Of the persons with TFA or knee disarticulation, 2 wore the X3 microprocessor knee and Vari-Flex XC foot, 1 wore the X2 microprocessor knee and Vari-Flex XC foot, and 1 wore the Total Knee 2100 mechanical knee and Vari-Flex XC foot.

Table 1
Demographic information by participant category (CTRL = uninjured controls, TTA = persons with transtibial amputation, and TFA = persons with transfemoral amputation or knee disarticulation). Note, there were no significant differences in demographic information or walking speeds (all $P > .167$).

	Age (yr)	Months Since Amputation	Height (m)	Mass (kg)	In-line Walking Speed (m/s)
CTRL	20		1.8	61.5	1.4
	28		1.7	88.4	1.4
	31		1.9	105.7	1.4
	28		1.9	72.6	1.3
	29		1.8	83.5	1.3
TTA	24	5.5	1.8	90.9	1.4
	27	47.8	1.8	106.9	1.4
	34	133.3	1.9	89.9	1.5
	45	17.7	1.8	135.6	1.5
TFA	34	59.7	1.7	71.4	1.1
	23	15.8	1.9	96.2	1.4
	26	59.0	1.7	74.9	1.4
	25	32.9	1.7	101.2	1.2

2.2. Experimental procedures

Each participant performed 20 turns involving a 90-degree change in direction to the left (n = 10) and right (n = 10). Participants walked at their self-selected speed along a 12-foot straight path and were verbally cued to turn left or right at a specified and consistent location (approximately 6 feet away from the turning point, allowing the participant to ultimately self-select the pivot limb). Turn direction was randomized, and no specific guidance was provided for which foot or type of turn (i.e., step vs. spin) to employ. Full-body kinematics were collected by tracking (120 Hz) 70 reflective markers with a 27-camera motion capture system (Vicon, Oxford, UK). Markers were placed on the C7 and T10 spinous processes, sternal notch, xiphoid process, and bilaterally on the acromia, ASIS, and PSIS. Lower and upper extremities were tracked as 6 DOF segments, with markers placed accordingly (Collins, Ghoussayni, Ewins, & Kent, 2009). All kinematic data were filtered at 6 Hz using a 5th order Butterworth filter.

2.3. Dependent measures and data analyses

The pivot foot and type of turn (step or spin) were first determined using a previously described, automated method (*cf.* Golyski & Hendershot, 2017), and heel strike/toe-off events were calculated using the position of the feet relative to the pelvis (Zeni, Richards, & Higginson, 2008). Step lengths were calculated for the step leading into pivot and the step after pivot as the absolute distance between the positions of heel strikes of each respective step relative to pivot. Stride widths were evaluated using the heel strike positions of the steps before, during, and after the turn (Huxham, Gong, Baker, Morris, & Iansek, 2006).

Three-dimensional trunk segmental kinematics were computed, relative to the pelvis, using Visual3D (Version 5.02.27, C-Motion Inc., Germantown, MD, USA), with local coordinate systems defined by a static calibration trial. Trunk-pelvis range of motion was calculated for each plane over the period from heel strike of the step before pivot to the toe-off of the step after pivot. Individual trunk and pelvic segmental trajectories were also computed and exported to MATLAB (Release 2015a, The MathWorks, Inc. Natick, MA, USA). Using these, tri-planar translational (Eq. (1)) and rotational (Eq. (2)) angular momenta of the trunk and pelvis segments were calculated as described by Gaffney et al. (2016), and normalized by each participant's body mass, height, and self-selected in-line walking speed (Herr & Popovic, 2008). Translational angular momentum (TAM) for the trunk and pelvis segments was calculated as:

$$\boldsymbol{h}_{i/Foot} = (\boldsymbol{r}_i - \boldsymbol{r}_{Foot}) \times m_i(\boldsymbol{v}_i - \boldsymbol{v}_{Foot}) \tag{1}$$

where $\boldsymbol{r}_i$ is the position vector of the segment's center of mass, $\boldsymbol{r}_{Foot}$ is the position vector of the pivot foot, m_i is the mass of the segment, $\boldsymbol{v}_i$ is the velocity vector of the segment's center of mass, and $\boldsymbol{v}_{Foot}$ is the velocity vector of the pivot foot. TAM was evaluated only during the period from heel strike before the turn to toe-off after the turn (i.e. pivot stance). Rotational angular momentum (RAM) for the trunk and pelvis segments was calculated as:

$$\boldsymbol{h}_i = \boldsymbol{I}_i \cdot \boldsymbol{w}_i \tag{2}$$

where $\boldsymbol{I}_i$ is the moment of inertia tensor for the segment of interest and $\boldsymbol{w}_i$ is the segment's angular velocity vector. RAM was evaluated during the same period as trunk-pelvis range of motion. Both TAM and RAM were resolved in the three planes of motion, defined using center of mass velocity (to define a forward direction), gravity, and the resulting cross product; TAM and RAM ranges (i.e., max-min) were extracted within each plane for subsequent analyses.

Finally, inter-segmental coordination of the trunk and pelvis in each plane of movement was calculated using a vector coding method described by Needham, Naemi, and Chockalingam (2014). For this, each turn was subsequently divided into two phases: (1) pivot stance; defined as the period from heel strike to toe-off of the foot in stance during the apex of the turn, and (2) pivot swing; defined as the period from pivot foot toe-off to subsequent ipsilateral heel strike. Time-series trajectories of the trunk and pelvic angle defined a 0–360° relative coupling angle, which at each time point is separated into one of eight 45° bins to evaluate the frequency of a given coordination mode (in-phase, anti-phase, trunk-phase, and pelvic-phase) in both pivot stance and pivot swing; circular statistics were used to define the mean and variability of each coupling angle/phase while preserving directionality of the trunk-pelvis relative coupling angle (Hamill, Haddad, & McDermott, 2000; Needham et al., 2014; Watson & Batschelet, 1982). Note, a common alternative method for assessing segmental coordination is continuous relative phase, but this method does not explicitly quantify trunk- and pelvic-phase coordination modes (i.e., dominance of a given segment).

2.4. Statistical analyses

Given that turn type was not controlled as part of the experimental design (i.e., the pivot foot was selected by the participant), and no *a priori* hypotheses were formulated as to how turn type would influence the dependent variables, no explicit comparisons were made between turn strategies. Instead, Mann-Whitney U tests were used to compare all dependent measures between persons with LLA vs. CTRL, separately within each turn type; statistical significance was concluded at $P < .050$. All statistical analyses were performed in SPSS (version 21.0; IBM SPSS Inc., Chicago, IL). Unless otherwise specified, data are reported as medians (interquartile ranges). In total, 60 (of 80) trials/turns from persons with TFA, 71 (of 80) from persons with TTA, and 77 (of 100) from persons without LLA were included as part of subsequent analyses due to marker drop out and/or in-line walking periods of insufficient length before and after the turn.

3. Results

3.1. Turn type and temporal-spatial parameters

Persons with TFA performed 32 step turns (13/19 on the intact/prosthetic limb, respectively) and 28 spin turns (9/19 on the intact/prosthetic limb). Persons with TTA performed 51 step turns (34/17 on the intact/prosthetic limb) and 20 spin turns (19/1 on the intact/prosthetic limb). CTRL performed 51 step and 26 spin turns. During spin turns, no significant differences were observed between persons with or without amputation in step lengths before pivot [LLA: 58.9 (11.2), CTRL: 64.0 (7.9) cm; $P = .343$], step lengths after pivot [LLA: 60.8 (27.0), CTRL: 60.1 (9.5) cm; $P = .734$], and stride widths over the pivot [LLA: 13.8 (11.0), CTRL: 17.6 (7.6) cm; $P = .427$]. Similarly, during step turns no significant differences were observed between persons with and without amputation in step lengths before pivot [LLA: 62.3 (9.2), CTRL: 67.2 (16.8) cm; $P = .310$], step lengths after pivot [LLA: 67.3 (15.6), CTRL: 68.0 (8.8) cm; $P = .586$], and stride widths over the pivot [LLA: 45.0 (6.2), CTRL: 46.4 (5.0) cm; $P = .363$].

3.2. Trunk and pelvic kinematics

During spin turns, sagittal plane range of motion was similar between individuals with vs. without LLA [LLA: 8.5 (3.1), CTRL: 6.9 (5.9)°; $P = 1.000$]. Conversely, frontal plane trunk-pelvis range of motion was significantly smaller in the LLA group than the CTRL group [LLA: 11.4 (3.5), CTRL: 15.3 (6.3)°; $P = .004$]. Transverse plane trunk-pelvis range of motion was not significantly different between groups [LLA: 19.2 (8.4), CTRL: 16.5 (5.0)°; $P = .384$]. During step turns, trunk-pelvis range of motion was larger in the LLA vs. CTRL groups in the sagittal plane [LLA: 8.9 (2.6), CTRL: 6.5 (3.9)°; $P = .047$], but no significant differences between groups were observed in the frontal plane [LLA: 11.7 (5.0), CTRL: 17.5 (7.1)°; $P = .201$] or transverse plane [LLA: 15.4 (5.0), CTRL: 14.6 (2.1)°; $P = .586$].

3.3. Trunk and pelvic angular momenta

3.3.1. Translational angular momentum

During both spin ($P > .157$) and step turns ($P > .087$), group was not a significant main effect for TAM of the trunk or pelvis in any plane (Fig. 1/Table 2).

3.3.2. Rotational angular momentum

During spin turns, trunk and pelvis RAM in the sagittal plane were not significantly different by level of amputation ($P > .115$). However, frontal plane trunk RAM ($P < .001$), and pelvis RAM ($P = .047$) were larger in individuals with vs. without LLA. Additionally, trunk and pelvis RAM in the transverse plane were not significantly different by group ($P > .678$; see Fig. 2). During step turns, trunk and pelvis RAM in the sagittal plane were not significantly different by level of amputation ($P > .698$). Frontal plane trunk RAM was larger among individuals with vs. without LLA ($P < .001$), while frontal plane pelvis RAM was not ($P = .310$). No significant differences were observed in the transverse plane between groups in either trunk or pelvis RAM ($P > .391$; see Fig. 2/ Table 2).

3.4. Trunk and pelvic coordination

No significant differences by group were observed in trunk-pelvis coordination angle variability between persons with vs. without LLA ($P > .098$; Table 3). During spin turns, there were no significant differences in the frequency of any coordination mode in either stance or swing phase ($P > .082$). During step turns, transverse plane pelvis-phase coordination was significantly more frequent in individuals with vs. without LLA ($P = .036$), with no other coordination mode exhibiting significant differences between populations ($P > .068$; Fig. 3).

4. Discussion

The aim of this study was to characterize compensatory movements of the trunk and pelvis during transient 90 degree turns in persons with vs. without LLA. We hypothesized that differences in coordination would exist principally in the sagittal and frontal planes among persons with LLA, concurrent to increases in segmental momenta, to overcome limitations associated with altered ankle function. In support of our hypotheses, ranges of motion, segmental rotational momenta, and frequency of coordination modes differed between individuals with and without LLA, depending on the plane and type of turn employed.

4.1. Trunk-pelvis coordination

Coordinated movements of the trunk and pelvis are important for efficient and steady ambulation, and alterations in trunk-pelvic coordination strategy (or its variability) have been associated with current or future risk for LBP (Hamill, Van Emmerik, Heiderscheit, & Li, 1999; Seay, Van Emmerik, & Hamill, 2011). Though more frequent in-phase coordination has been associated with LBP (Seay et al., 2011) and may decrease relative motion of the trunk to the pelvis as a guarding strategy (Russell Esposito & Wilken, 2014; van der Hulst, Vollenbroek-Hutten, Rietman, & Hermens, 2010) by preventing strain on anatomical structures of the low back, in the

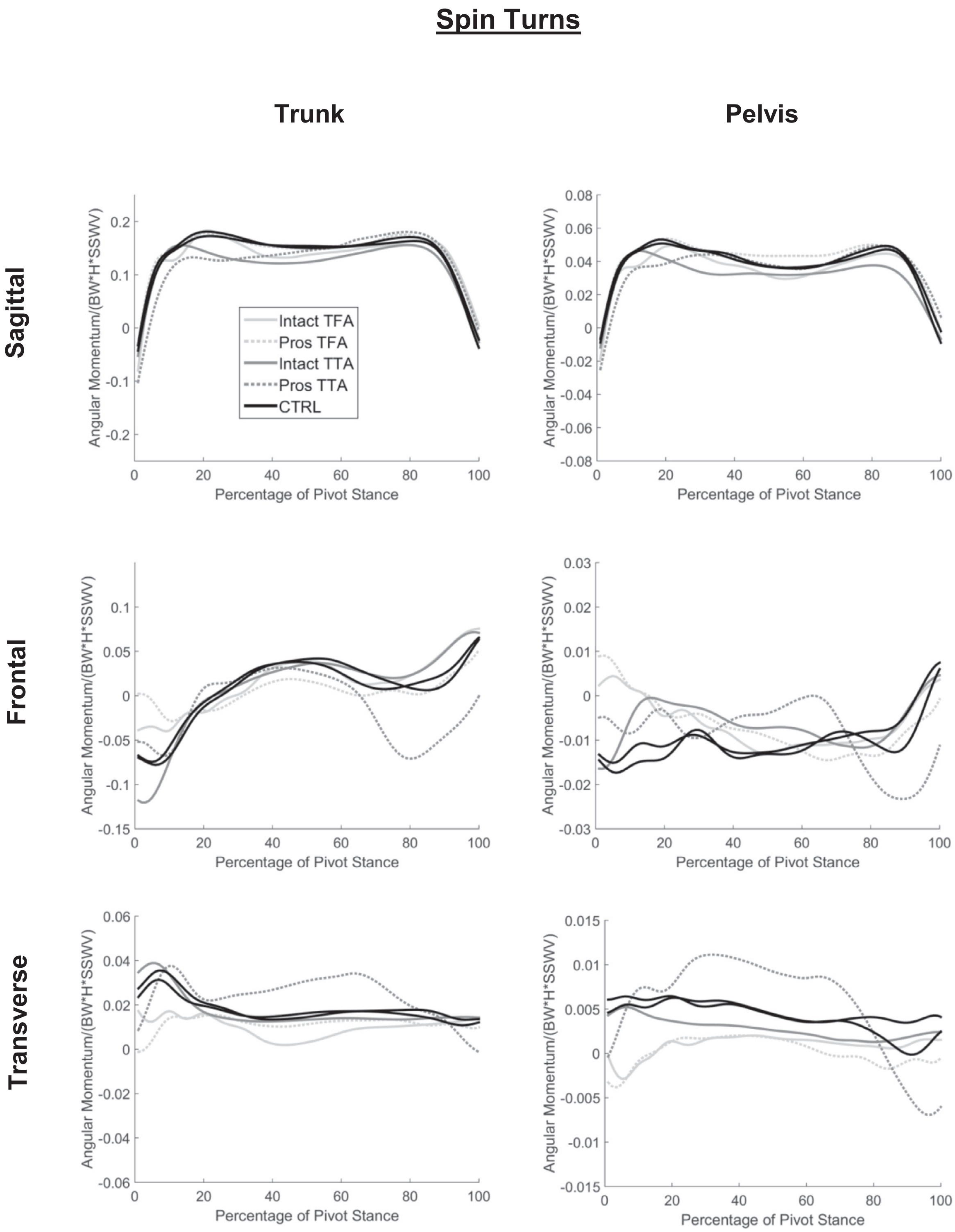

Fig. 1. Ensemble averages of trunk and pelvis translational angular momentum (TAM) in the sagittal, frontal, and transverse planes. Data are normalized by body weight (BW), height (H), and self-selected (in-line) walking velocity (SSWV). The two traces for each turn strategy executed by controls represent turns performed on the right and left feet and are provided as an indicator of healthy variability in angular momenta in each plane. For visualization purposes, frontal and transverse TAM for both the trunk and pelvis were negated for left turns.

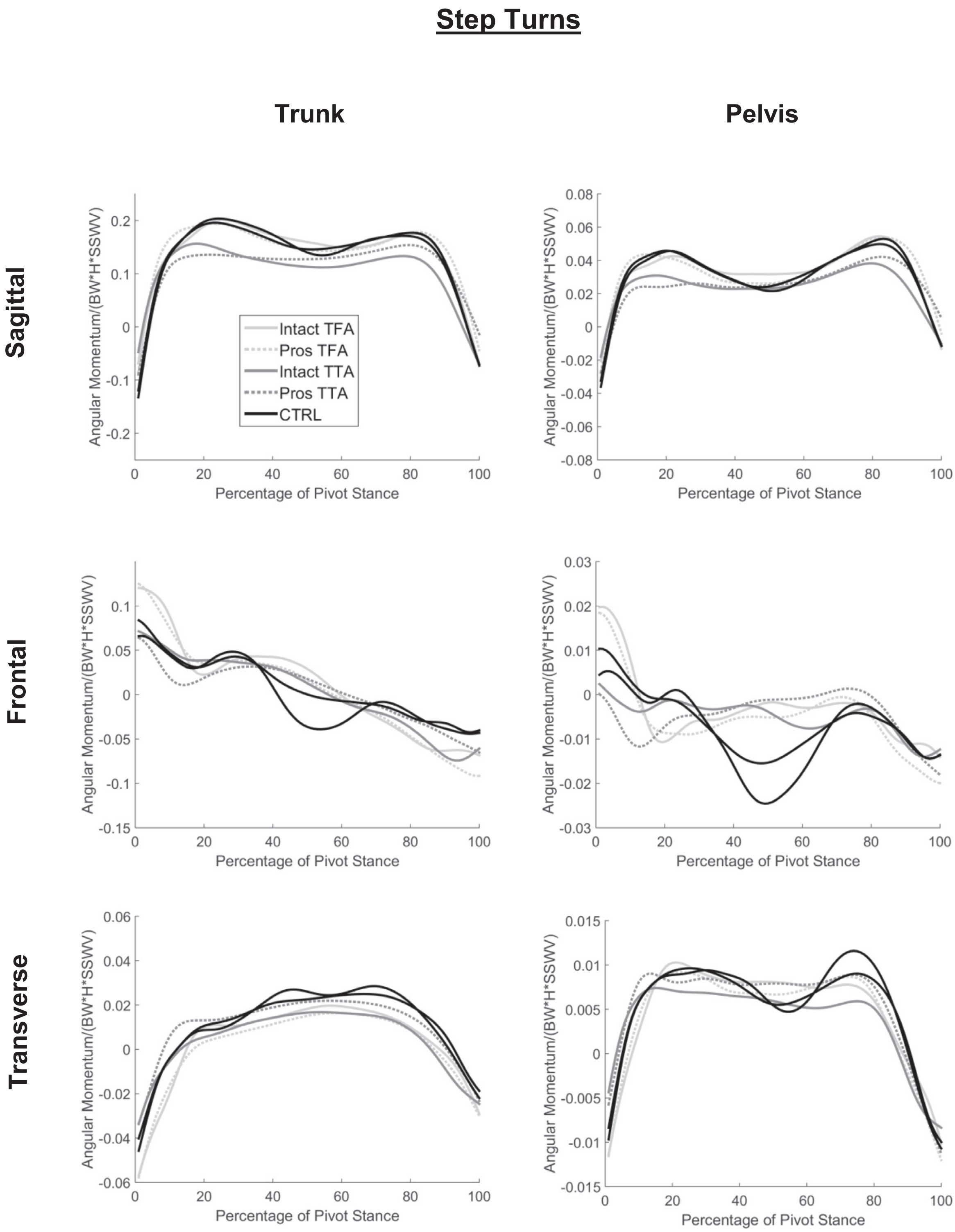

Fig. 1. (*continued*)

transverse plane individuals with vs. without LLA exhibited a lesser (though not significant) frequency of in-phase coordination compared to uninjured controls. Such a decrease in in-phase coordination could be a compensatory mechanism for reduced ankle function, but may indicate an increased risk of repetitive injury. Though no participants reported acute pain during collection, a limitation of the present study was that LBP history was not collected.

To the authors' knowledge, the only previous study of trunk-pelvis coordination during turns evaluated differences in transverse

Table 2
Median (interquartile range) ranges in trunk and pelvic translational angular momenta (TAM) and rotational angular momenta (RAM) for individuals with unilateral lower limb amputation (LLA) and uninjured controls (CTRL), during spin/step turns. TAM was calculated during pivot stance. RAM was calculated during the period from heel strike of the step before the pivot step to toe-off of the step after the pivot step, respectively. For metrics marked by * and **, groups were significantly different at the α = 0.05 and α = 0.001 levels, respectively. All momenta are normalized by body weight, height, and self-selected (in-line) walking speed.

		Spin Turns	
		LLA	CTRL
Sagittal	Trunk TAM Range	0.2781 (0.0824)	0.2645 (0.0608)
	Pelvis TAM Range	0.0733 (0.0206)	0.0695 (0.0172)
	Trunk RAM Range	0.0022 (0.0011)	0.0021 (0.0008)
	Pelvis RAM Range	0.0004 (0.0002)	0.0003 (0.0001)
Frontal	Trunk TAM Range	0.1420 (0.1087)	0.1474 (0.0927)
	Pelvis TAM Range	0.0337 (0.0179)	0.0320 (0.0213)
	Trunk RAM Range**	0.0055 (0.0022)	0.0038 (0.0011)
	Pelvis RAM Range*	0.0006 (0.0004)	0.0005 (0.0001)
Transverse	Trunk TAM Range	0.0313 (0.0158)	0.0278 (0.0189)
	Pelvis TAM Range	0.0135 (0.0076)	0.0084 (0.0058)
	Trunk RAM Range	0.0059 (0.0023)	0.0056 (0.0021)
	Pelvis RAM Range	0.0014 (0.0010)	0.0017 (0.0007)
		Step Turns	
		LLA	CTRL
Sagittal	Trunk TAM Range	0.2931 (0.0506)	0.3095 (0.0591)
	Pelvis TAM Range	0.0808 (0.0221)	0.0835 (0.0162)
	Trunk RAM Range	0.0023 (0.0013)	0.0024 (0.0008)
	Pelvis RAM Range	0.0004 (0.0003)	0.0003 (0.0001)
Frontal	Trunk TAM Range	0.1817 (0.1046)	0.1580 (0.0539)
	Pelvis TAM Range	0.0326 (0.0204)	0.0385 (0.0215)
	Trunk RAM Range**	0.0046 (0.0021)	0.0030 (0.0005)
	Pelvis RAM Range	0.0006 (0.0004)	0.0005 (0.0002)
Transverse	Trunk TAM Range	0.0683 (0.0182)	0.0717 (0.0210)
	Pelvis TAM Range	0.0235 (0.0090)	0.0270 (0.0082)
	Trunk RAM Range	0.0058 (0.0034)	0.0070 (0.0030)
	Pelvis RAM Range	0.0015 (0.0011)	0.0014 (0.0006)

Units: Angular Momentum/(Body Weight*Height*Self Selected Walking Velocity).

plane coordination only, in persons with and without LBP during spin turns (Smith & Kulig, 2016). The dominant in-phase coordination in the transverse plane during stance was consistent with this work, though no significant differences were found between groups.

Characterization of trunk-pelvis coordination during in-line walking (Russell Esposito & Wilken, 2014) found higher frequencies of anti-phase coordination in individuals with TFA relative to uninjured controls in the sagittal and frontal planes. In contrast, we only observed a significant increase in transverse plane pelvic-phase coordination in persons with vs. without LLA during step turns, which are more biomechanically similar to in-line walking than spin turns (Taylor et al., 2005). In contrast to the hypothesized changes in sagittal and frontal coordination, the only significant difference between groups was in the transverse plane during step turns, which nonetheless suggests alternative proximal movement strategies within the LLA population. In support of our hypothesis, differences between populations were observed in trunk-pelvis range of motion and angular momenta in the sagittal and frontal planes.

4.2. Sagittal plane

Significantly larger sagittal trunk-pelvis range of motion during step turns among persons with vs. without LLA is consistent with previous observations of trunk-pelvis kinematics during in-line walking (Goujon-Pillet et al., 2008). Such larger trunk flexion angles may be a compensation to facilitate hip extension, which is hampered by hip flexion contractures (Gailey et al., 2008), but this motion may also increase demand on trunk extensors (Hendershot & Wolf, 2014). Moreover, this more extreme sagittal trunk-pelvic movement is also consistent with the larger (though not significant) observed pelvis RAM in persons with vs. without LLA, and in agreement with a previous study of in-line walking (Gaffney et al., 2016).

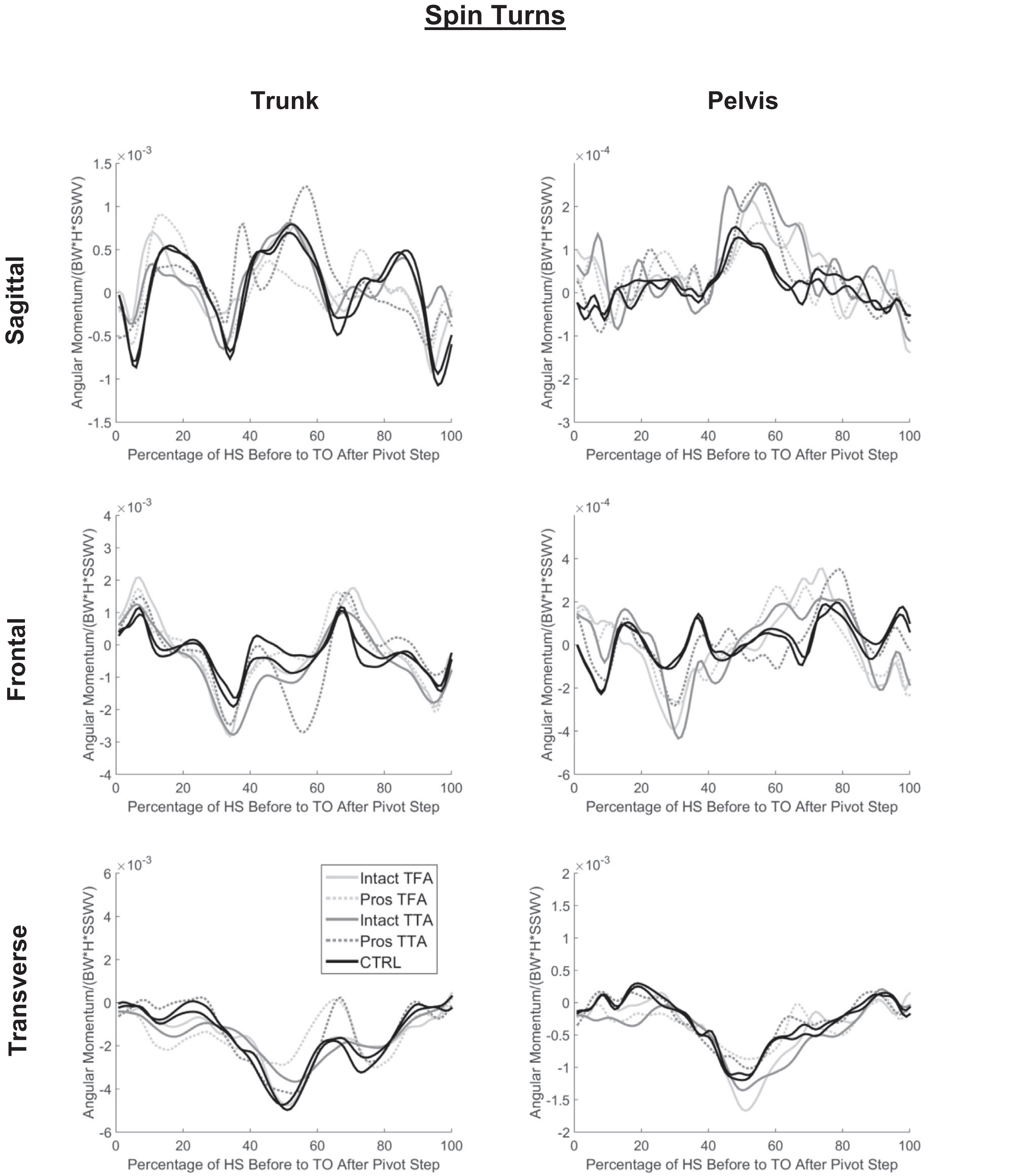

Fig. 2. Ensemble averages of trunk and pelvis rotational angular momentum (RAM) in the sagittal, frontal, and transverse planes during spin/step turns. Data are normalized by body weight (BW), height (H), and self-selected (in-line) walking velocity (SSWV). The two traces for each turn strategy executed by controls represent turns performed on the right and left feet and are provided as an indicator of healthy variability in angular momenta in each plane. For visualization purposes, frontal and transverse RAM for both the trunk and pelvis were negated for left turns.

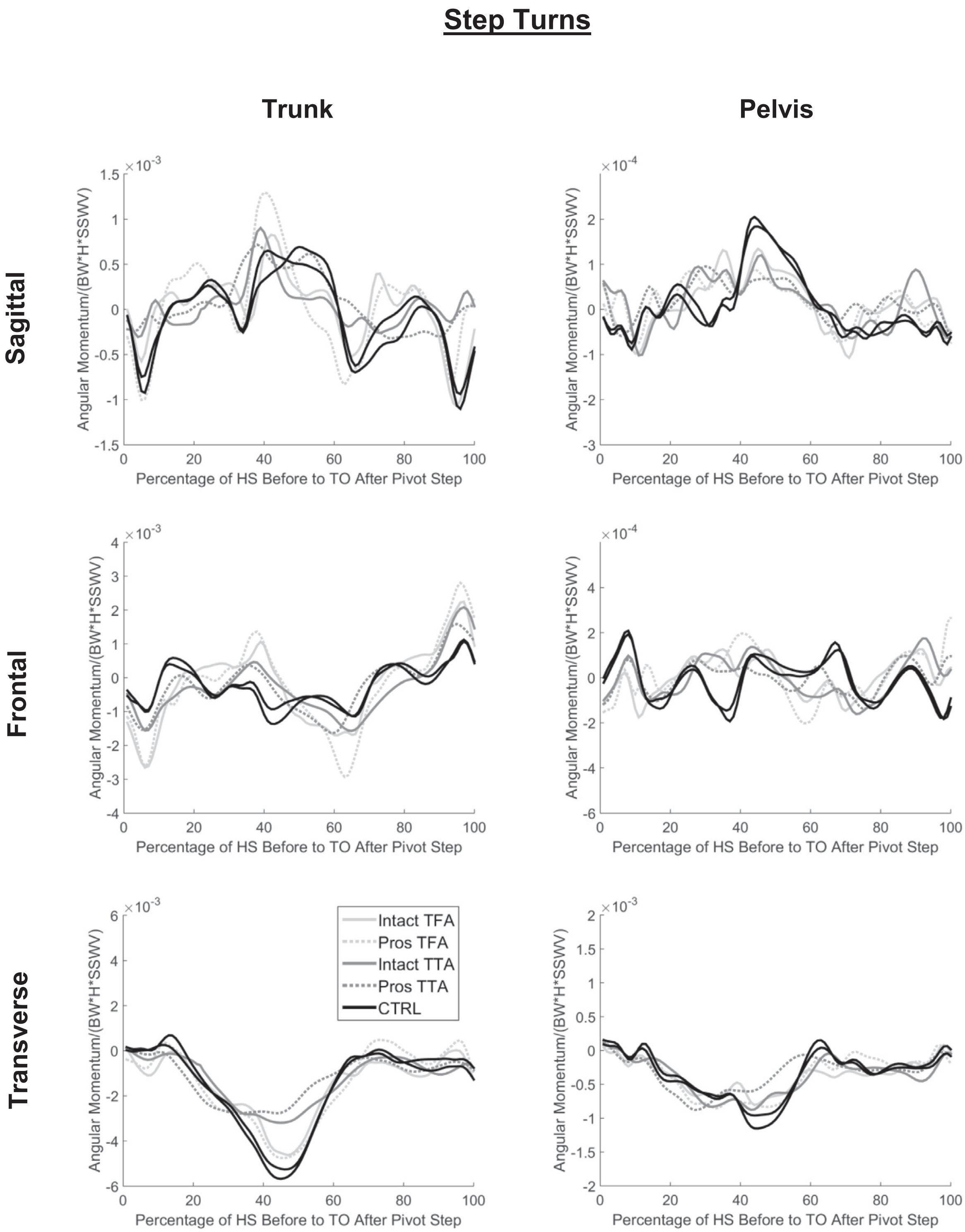

Fig. 2. (*continued*)

4.3. Frontal plane

In contrast to increases in trunk-pelvis range of motion among persons with vs. without LLA during in-line walking (Goujon-Pillet et al., 2008; Yoder et al., 2015), frontal plane range of motion during spin turns was smaller in the LLA than CTRL group (no difference between groups during step turns). During in-line walking, a larger range of motion is primarily due to increased lateral

Table 3
Median (interquartile range) variability of trunk-pelvis coupling angle during spin/step turns in stance/swing of the pivot limb for persons with unilateral lower limb amputation (LLA), and uninjured controls (CTRL). No significant differences were found between groups at the α = 0.05 level.

		Spin Turns	
		LLA	CTRL
Sagittal	Pivot Stance (°)	21.7 (33.9)	23.2 (17.6)
	Pivot Swing (°)	8.7 (20.2)	19.6 (15.0)
Frontal	Pivot Stance (°)	10.8 (27.1)	14.7 (18.7)
	Pivot Swing (°)	20.1 (33.5)	27.5 (25.6)
Transverse	Pivot Stance (°)	12.1 (21.1)	8.8 (6.2)
	Pivot Swing (°)	18.0 (43.0)	26.0 (20.1)
		Step Turns	
		LLA	CTRL
Sagittal	Pivot Stance (°)	23.2 (7.5)	22.7 (17.1)
	Pivot Swing (°)	22.2 (31.2)	13.8 (19.9)
Frontal	Pivot Stance (°)	15.9 (9.1)	13.5 (8.0)
	Pivot Swing (°)	28.2 (31.9)	28.1 (20.9)
Transverse	Pivot Stance (°)	14.7 (6.0)	12.3 (10.4)
	Pivot Swing (°)	23.5 (26.6)	24.9 (15.7)

trunk lean over the prosthetic limb and is considered a compensation, at least in part, for reduced residual limb function (Hendershot & Wolf, 2014; Rueda et al., 2013). Future studies exploring turns on the intact vs. prosthetic side may elucidate the basis for reduced frontal plane range of motion, though we speculate the relative decrease in lateral trunk lean throughout turns may be a result of the more proximal (i.e., hip vs. ankle) strategy and generally not leaning into/away from the turn to minimize excursions of the body center of mass and improve stability (Ventura et al., 2011). Despite the trends in frontal plane trunk-pelvis range of motion being inconsistent with those of existing literature, differences in frontal plane trunk RAM (which is dependent on segmental angular velocity) between groups during both turn types were apparent. Such differences are consistent with our hypothesis and previous work identifying larger ranges in whole body frontal plane angular momentum in persons with LLA (albeit during in-line walking; Silverman & Neptune, 2011). Large changes in whole-body angular momentum in the frontal plane have also been correlated with poorer clinical balance outcomes post-stroke (Nott, Neptune, & Kautz, 2014). Moreover, such deviations in trunk and pelvis angular momentum in the frontal plane are of particular interest since these segments are the principal contributors to whole body angular momentum in the frontal plane (Herr & Popovic, 2008). During spin turns the more extreme frontal trunk angular velocity coupled with smaller trunk-pelvic range of motion could suggest a trunk-stiffening strategy (Arendt-Nielsen, Graven-Nielsen, Svarrer, & Svensson, 1996; Lamoth et al., 2002), similar to the segmental rigidity identified among persons with TFA during in-line walking (Russell Esposito & Wilken, 2014); however, such a stiffening strategy would likely be associated with increased in-phase co-ordination (Wu et al., 2014) – a trend we did not observe here with the vector coding method.

4.4. Transverse plane

Larger axial trunk rotations have been observed in persons with TFA during in-line walking (Goujon-Pillet et al., 2008), which are concerning given the association of such rotations with LBP (Fujiwara et al., 2000; Morgenroth, Medverd, Seyedali, & Czerniecki, 2014). We did not observe differences in transverse plane trunk-pelvis range of motion, though this could be attributed to turns requiring more control over transverse plane angular displacements. However, during step turns, range in transverse trunk RAM was smaller, albeit not significantly, in persons with LLA compared to uninjured controls. As illustrated in Fig. 2, at approximately 50% of the turn the trunk RAM was smaller for turns on both the prosthetic and intact limbs in persons with TFA and TTA vs. controls, indicating a smaller peak trunk angular velocity in the LLA group. This contradicts previous preliminary findings which suggested persons with unilateral TTA execute step turns with larger transverse trunk angular velocities than uninjured controls (Taylor & Strike, 2009).

Though the host of kinematic differences between turns and in-line walking (Taylor et al., 2005) precludes direct comparison of angular momentum components to previous work, qualitatively, transverse trunk and pelvis RAM were the most different in shape between the two ambulation tasks (c.f. Gaffney et al., 2016), stemming from the seemingly necessary peak in transverse angular velocity. Moreover, differences in the range of TAM/RAM between in-line walking and turns were most pronounced in the transverse plane, and were larger during transient turns by factors of 2 and 3 for TAM and RAM, respectively.

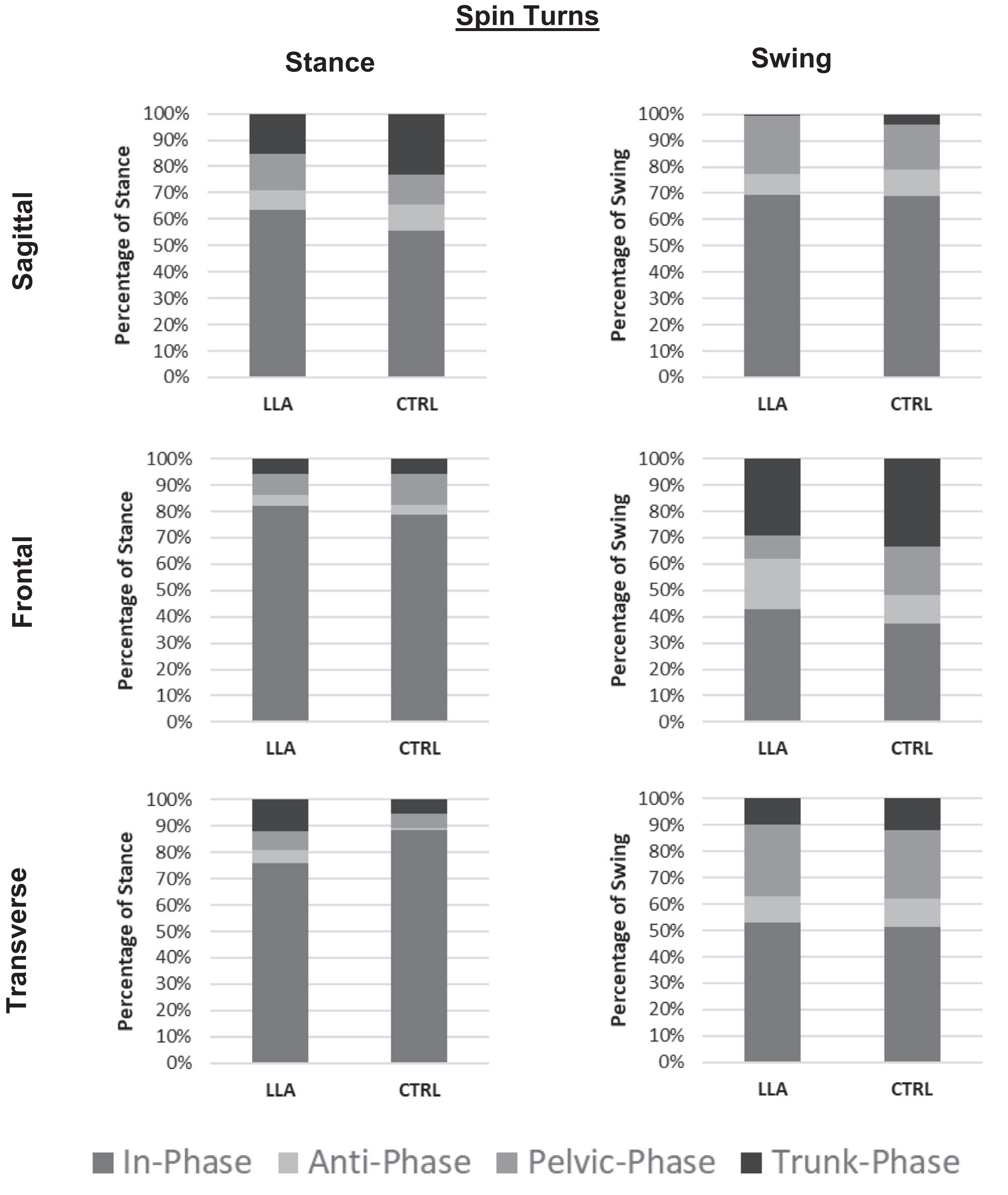

Fig. 3. Proportions of trunk-pelvis coordination modes during pivot stance and swing in each plane among persons with unilateral lower limb amputation (LLA) and uninjured controls (CTRL), by spin and step turns. Significant comparisons (*) between groups were at the α = 0.05 level only.

4.5. Limitations

Several limitations require attention when interpreting results of the current study. First, the generalizability of findings may be limited given persons with LLA were young, healthy, and otherwise uninjured members of the military who had sustained traumatic lower limb amputations. Second, the small sample sizes, combination of individuals with different levels of amputation into the LLA group, and many inherent levels of potential comparisons precluded additional analyses between pivot legs (i.e., prosthetic and intact). The five-person control group also may not provide an accurate statistical representation of the healthy able-bodied

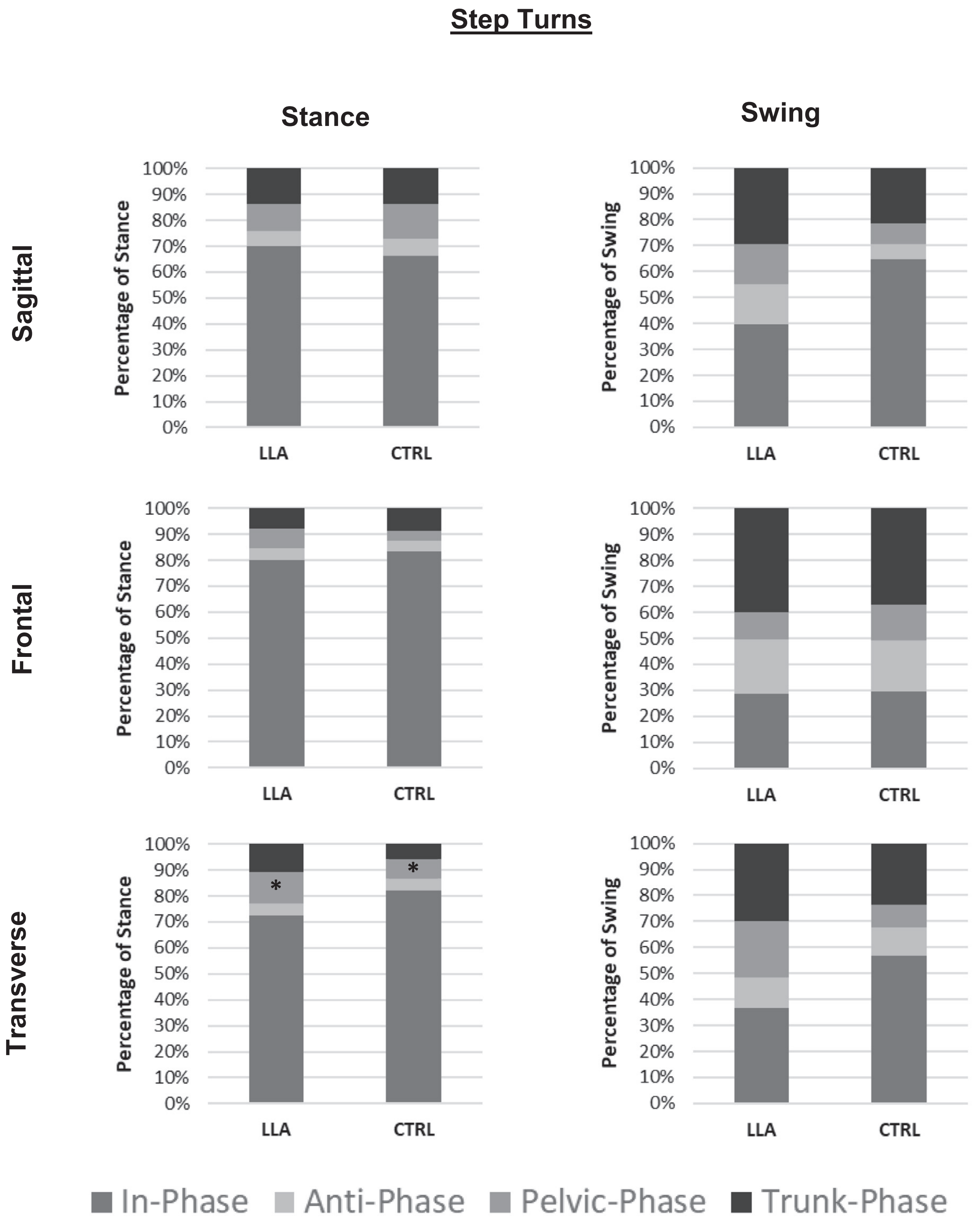

Fig. 3. (*continued*)

population at large, and future studies with larger sample sizes are warranted. Third, although we suggest that observed differences in trunk-pelvis movement patterns between persons with and without LLA may be associated with elevated risk of LBP onset or recurrence, we did not specifically control for its presence or prior/recent history, though no participants reported acute LBP during testing. Fourth, we did not specifically evaluate the influences of arm motion. While most likely to affect angular momentum in the transverse plane (Collins, Adamczyk, & Kuo, 2009; Herr & Popovic, 2008), general qualitative differences in arm swing strategies between groups were not observed. Finally, the turn cueing paradigm used was intended to represent transient changes in direction encountered in daily life, though the somewhat unpredictable, verbally-cued direction may have resulted in events that are more difficult to reproduce than turns in other studies wherein participants walked along a more consistent circular path (Segal et al., 2008; Ventura et al., 2011). Future work can control for such variability with alternative cueing methods (e.g., visual, compared to our auditory cues; Heitzmann et al., 2015), thereby also supporting explicit comparisons between step vs. spin turns, and potential interactions with the chosen pivot limb (i.e., prosthetic vs. intact).

4.6. Summary

We compared features of trunk-pelvic segmental motion and coordination between persons with and without LLA during 90-degree turns executed using self-selected step and spin strategies. We observed differences in the frequencies of inter-segmental coordination, trunk-pelvis ranges of motion, and segmental momenta across levels of amputation, depending on the plane and method of turn employed. Nevertheless, the identified compensatory adaptations used by persons with unilateral LLA to execute this common, but biomechanically challenging, task may be “maladaptive” and thus predispose these individuals to developing LBP (or its recurrence) with repeated exposure over the longer term.

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