

AWARD NUMBER: W81XWH-14-2-0144

TITLE: Evaluation of Spine Health and Spine Mechanics in Servicemembers with Traumatic Lower Extremity Amputation or Injury

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Fort Detrick, Maryland 21702-5012

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13. SUPPLEMENTARY NOTES					
14. ABSTRACT Low back pain (LBP) is an important secondary health condition following lower-extremity trauma, with an estimated prevalence as high as 52-89%, and reported as the condition most contributing to a reduced quality of life. During gait, alterations in trunk motion following lower-extremity trauma likely impose distinct demands on trunk muscles to maintain equilibrium and stability of the spine that, with repeated exposure, may increase risk for LBP. The overall objective of this research is to characterize features of trunk (spine) motion with lower-extremity trauma, thereby elucidating the relationship(s) between trunk motion and LBP risk via changes in spine mechanics and spine health. Using a novel set of clinical, experimental, and computational methods, we have demonstrated that altered trunk motions with lower-extremity trauma increase spinal loads by 17-95% relative to uninjured individuals. Moreover, we expect to show a positive association between these elevated spinal loads and poor spine health/history of LBP, which will support the need for trunk-specific rehabilitation procedures to reduce long-term incidence and recurrence of LBP. While we have been successful in disseminating these results thus far via scientific journals and conference presentations, within the remaining period of performance we will execute a strategic dissemination plan with several additional manuscripts that summarize key findings.					
15. SUBJECT TERMS Low Back Pain; Intervertebral Disc; Inter-Segmental Motion; Spine Load; Finite Element Model; Biomechanics					
16. SECURITY CLASSIFICATION OF:			17. LIMITATION OF ABSTRACT	18. NUMBER OF PAGES	19a. NAME OF RESPONSIBLE PERSON
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1. INTRODUCTION:

Linking lower-extremity trauma (i.e., amputation/injury) with low back pain (LBP) risk via biomechanical theory suggests that altered and asymmetric trunk motions and corresponding passive spinal tissue and trunk neuromuscular responses alter spine mechanics such that would, over time, adversely affect spine health. Therefore, the overall objective of this study is to investigate such relationships through cross-sectional evaluations of spine health and spine mechanics in persons with lower-extremity amputation/injury (with and without LBP) and uninjured controls.

KEYWORDS: Low Back Pain; Intervertebral Disc; Inter-Segmental Motion; Spine Load; Finite Element Model

2. ACCOMPLISHMENTS:

What were the major goals of the project?

This study has three main aims, as indicated below:

Specific Aim 1: Quantify lumbar spinal alignment and inter-segmental vertebral motions with traumatic lower-extremity amputation.

Major Task 1: Obtain IRB and HRPO approvals.

Target Date: by April 2015

Actual Date: April 24, 2015 (IRB approval) / June 26, 2015 (HRPO approval)

Major Task 2: Complete biomechanical data collections, analysis, and interpretations.

Target Dates: Months 6-24 (~81% complete)

Additional Milestones: One abstract presented and one manuscript submitted.

Specific Aim 2: Quantify alterations in spine mechanics (loading) with traumatic lower-extremity amputation.

Major Task 3: Estimate spinal loads using collected biomechanical data as inputs into the finite element model of the lumbar spine.

Target Dates: Months 6-24 (~81% complete)

Additional Milestones: One abstract presented and one manuscript published.

Specific Aim 3: Determine associations between spine loading and current spine health with traumatic lower-extremity amputation.

Major Task 4: Conduct physical spinal examinations.

Target Dates: Months 6-24 (~81% complete)

Major Task 5: Obtain magnetic resonance images of the lumbar spine for quantitative evaluation of lumbar disc health.

Target Dates: Months 6-24 (50% complete)

Major Task 6: Author manuscript on entire study.

Target Dates: Months 30-36 (50% complete)

Additional Milestones: One abstract presented and one manuscript submitted.

What was accomplished under these goals?

Within this reporting period, substantial work was again performed under major tasks 2-6. Specifically, prospective data collections continued in the areas of biomechanical and clinical assessments focused on the trunk and spine to identify potential relationships with LBP risk factors. Biomechanical assessments include overground gait analyses with a focus on the trunk and spine, as well as trunk muscle activity recorded using surface EMG. In addition, we are also capturing a more comprehensive understanding of current/recent history of (chronic) LBP and its impact on daily life and functional activities; including the NIH Task Force LBP Questionnaire and a legacy LBP questionnaire (Oswestry Disability Index). We have obtained data from 58 participants to date (Table 1):

Table 1. Sample breakdown by level of injury and mean (standard deviation) participant demographics, by low back pain (LBP) status using the NIH chronicity definition.

<i>Level of Injury</i>	No Current/Recent History of (chronic) LBP	(chronic) LBP
Uninjured Controls	17	2
Transtibial Limb Loss	8	18
Transfemoral Limb Loss	3	9
<i>Participant Demographics</i>		
Age (yr)	29.5 (8.3)	35.7 (7.0)
Body Mass (kg)	74.4 (11.9)	91.0 (15.2)
Stature (cm)	175.2 (5.8)	179.5 (6.6)
Time since limb loss (yr)	4.2 (4.5)	9.2 (6.3)

Several new key findings include:

- Evaluation of trunk muscle activities during gait identified differences in the motor control strategies underlying the observed trunk motion patterns. Specifically, persons with lower-extremity trauma demonstrated a second peak in erector spinae activation during mid-terminal swing (not observed in controls), and an overall longer duration of activation throughout the gait cycle (see Butowicz et al., 2018 in *Journal of Electromyography and Kinesiology*). Trunk neuromuscular control strategies secondary to lower-extremity trauma are seemingly driven by functional requirements to generate force proximally to help advance the (affected) lower limb during gait.
- Interestingly, spinal loads derived from our finite element simulations indicated differential increases with faster walking speeds among persons with vs. without lower-extremity trauma. At the fastest (vs. slowest) speed, increases in peak compressive and shear forces were respectively 24-84% and 29-77% larger among persons with lower-extremity trauma vs. uninjured controls (see Hendershot et al., 2018 in *Journal of Biomechanics*). Over time, repeated exposures to these increased loads, particularly at faster walking speeds, may contribute to the elevated risk for LBP among persons with lower-extremity trauma.

- When evaluating the influences of LBP on spinal loads, despite larger motions in the frontal and transverse planes, spinal loads were similar between persons with lower-extremity trauma presenting both with and without (chronic) LBP; though these were generally still larger relative to uninjured controls (see Acasio et al., 2018 in *Proceedings of the American Society of Biomechanics*). Nevertheless, it is certainly plausible that the presence or history of LBP have concurrently altered features of trunk-pelvic motion, as previously observed among non-limb loss individuals with and without LBP.
- Preliminary (and prior) analyses using a legacy measure for LBP disability (Oswestry Disability Index; ODI) had identified minimal disability (43/58 reported less than 20% disability). However, categorization using the NIH Research Task Force (RTF) definitions for chronicity of LBP, which utilize both duration and frequency, told a different story (Table 2). Additional psychosocial outcomes and subcategories are also preliminarily reported below.

Table 2. Mean (standard deviation) classification/disability scores and individual psychosocial outcomes, by low back pain (LBP) status using NIH definition.

	No Current/Recent History of (chronic) LBP	(chronic) LBP
RTF Classification	10.6 (2.4)	16.9 (6.3)
ODI % Disability	2.0 (3.1)	22.8 (21.1)
Pain Intensity (7 days)	0.2 (0.1)	3.2 (1.5)
Pain Interference	1.1 (0.3)	3.4 (3.1)
Functional Impact	1.1 (0.3)	1.7 (0.7)
Anxiety and Depression	5.9 (5.2)	9.2 (7.0)
Pain Catastrophizing	2.5 (3.1)	22.8 (21.8)
Kinesiophobia	19.2 (3.6)	26.8 (4.3)

Also of note, progress in Major Task 5 has jumped to 50%, due in large part to recent efforts to build a multidisciplinary team to extract relevant information from the electronic health record, with a particular focus on the health of tissues and muscle morphology within the low back. While this has unfortunately been slow to develop, we expect to have initial data/results by the end of the year, and will disseminate in appropriate journals shortly thereafter.

What opportunities for training and professional development has the project provided?

Under the subaward to the University of Kentucky, Dr. Bazrgari and I were providing mentorship to a PhD student, Iman Shojaei (now graduated). Beyond that, the project was not necessarily intended to provide training or professional development opportunities; however, the hiring of Dr. Butowicz as a post-doctoral researcher allows additional training and mentorship opportunities as part of this project.

How were the results disseminated to communities of interest?

Within this reporting period (Year 4), results were disseminated via 9 conference abstracts/presentations and 9 peer-reviewed scientific manuscripts (See “Section 6: Products” for a list with citation details). The team also participated in 3 additional presentations wherein information was disseminated to the clinical and research communities (Amputation System of Care Grand Rounds, State of the Science Symposium at USUHS, and AMSUS).

What do you plan to do during the next reporting period to accomplish the goals?

With the approved 90-day no cost extension (through 12/31/2018), we will continue collecting data, particularly for a non-limb loss control group with LBP to serve as a reference group (need about 10 more), that will facilitate comparisons across the full-factorial combination of with/without limb loss and with/without LBP. Beyond that, our large emphasis is on results dissemination. While we have published many papers thus far, we are now focusing on differences in outcomes with respect to LBP status/history (in contrast to our prior work comparing to non-limb loss controls, both without LBP). Interestingly, we have identified the historical definitions used to characterize (chronic) LBP are not necessarily the most appropriate for this population. In previous reports, we’ve noted a disproportionally large number of the TTA (vs. TFA) cohort had chronic LBP. However, a larger proportion of those with TFA also had what would be considered LBP if using different components of the NIH minimal dataset (that speaks both to duration and/or frequency). Psychosocial correlates (kinesiophobia, anxiety, and depression) also appear to play a substantial role in this population. As such, a critical, perhaps ground-breaking, publication for the near term will describe the LBP experience in these patients, while comparing/contrasting different legacy measures for evaluating LBP. Also, as previously noted we have not been able to obtain MR and other images prospectively; nevertheless, we have assembled a strong multi-disciplinary team of rehabilitation orthopaedic researchers to retrospectively evaluate features of the electronic health record for the purposes of more completely describing spine health across a larger cohort of persons with extremity trauma. This will include pre-injury data (if available), but notably LBP diagnoses/treatments and other clinical information in conjunction with imaging to evaluate spine health and muscle morphology. Thus, in total, we are planning the following 5 manuscripts in the next 3 months:

1. Quantify and compare the mechanical environment (spinal loads) during gait between persons with and without limb loss, both with and with low back pain. This will be first written as an abstract to the *3rd International Workshop on Spinal Loading and Deformation* (due in December) and, if accepted, automatically invited as a full article for publication in a special issue of the *Journal of Biomechanics* (as we have done in the past).
2. Quantify and compare trunk postural control strategies between persons with and without limb loss, both with and with low back pain. This will be written as an abstract for the *2019 Meeting of the International Society of Biomechanics / American Society of Biomechanics* (due in November), and subsequently a full article for either *Clinical Biomechanics* or *Gait & Posture*.

3. Quantify and compare trunk muscle activities between persons with and without limb loss, both with and without low back pain. This will be written as an abstract for the *2019 Meeting of the International Society of Biomechanics / American Society of Biomechanics* (due in November), and subsequently a full article for either *Clinical Biomechanics* or *Gait & Posture*.
 4. Describe the LBP experience in persons with limb loss, comparing different outcomes and questionnaires, with a secondary emphasis on psychosocial correlates. This will be written as a full article for publication in either *The Spine Journal* or *Pain Reports*.
 5. Characterize features of tissue health and muscle morphology within the low back among persons with limb loss. This will be written as a full article for publication in *The Spine Journal*.
4. **IMPACT:** Describe distinctive contributions, major accomplishments, innovations, successes, or any change in practice or behavior that has come about as a result of the project relative to:

What was the impact on the development of the principal discipline(s) of the project?

Nothing to report.

What was the impact on other disciplines?

Nothing to report.

What was the impact on technology transfer?

Nothing to report.

What was the impact on society beyond science and technology?

Our results to date support a prevailing model that altered trunk (spinal) motions among persons with lower-extremity trauma increase risk for the onset and/or recurrence of LBP. As we continue building evidence for this model, there is likely to be a strong case for interventional approaches aimed at controlling trunk motions and spinal loads during (and beyond) rehabilitation. While that is specific to one patient population, these relationships may advance overall public knowledge regarding such a common and impactful musculoskeletal disorder. Over time, this will reduce the substantial economic costs associated with its treatment and promote enhancements in psychological health and overall quality of life.

5. CHANGES/PROBLEMS:

Changes in approach and reasons for change

As briefly mentioned above, we had included additional clinically administered strength and endurance tests to bolster the biomechanical evaluations and improve eventual translation. These

were included as an amendment to the IRB-approved protocol, reviewed by HRPO as part of the CR and accepted on April 5, 2017. We have also initiated a separate (retrospective) protocol for purposes of evaluating other clinically relevant aspects of spine health and LBP history among service members with lower-extremity trauma, which we expect to compliment many aspects of the prospective protocol.

Actual or anticipated problems or delays and actions or plans to resolve them

Nothing to report

Changes that had a significant impact on expenditures

Nothing to report.

Significant changes in use or care of human subjects

No significant changes to report. As noted above, IRB/HRPO approval dates:

IRB approval granted on April 1, 2015 (formal approval documents were uploaded to IRBnet on April 24)

HRPO approval for WRNMMC was granted on June 26, 2015 (A-18549.1)

HRPO approval for University Kentucky was granted on June 29, 2015 (A-18549.2)

Walter Reed IRB official start date (permission to begin study): August 4, 2015

Walter Reed IRB continuing review date: March 30, 2019

6. PRODUCTS:

- **Publications, conference papers, and presentations**

Journal publications:

1. Golyski, P.R., Hendershot, B.D. (2018) Trunk and Pelvic Dynamics during Transient Turns among Persons with Unilateral Lower Limb Amputation. *Human Movement Science* 58: 41-54. Federal Support Acknowledged.
2. Hendershot, B.D., Shojaei, I., Acasio, J.C., Dearth, C.L., Bazrgari, B. (2018) Walking Speed Differentially Alters Spinal Loads among Persons with Traumatic Lower Limb Amputation. *Journal of Biomechanics* 70(21): 249-254. Federal Support Acknowledged.
3. Acasio, J.C., Butowicz, C.M., Golyski, P.R., Nussbaum, M.A., and Hendershot, B.D. (2018) Associations between trunk postural control in walking and unstable sitting at various levels of task demand. *Journal of Biomechanics* 75: 181-185. Federal Support Acknowledged.

4. Butowicz, C.M., Acasio, J.A., Dearth, C.L., Hendershot, B.D. (2018) Trunk Muscle Activation Patterns among Persons with Lower Limb Loss: Influences of Walking Speed. *Journal of Electromyography and Kinesiology* 40: 48-55. Federal Support Acknowledged.
5. Highsmith, M.J., Goff, L.M., Lewandowski, A.L., Farrokhi, S., Hendershot, B.D., Hill, O.T., Rabago, C.A., Russell-Espósito, E., Orriola, J.J., Mayer, J.M. (2018) Low Back Pain in Persons with Lower Extremity Amputation: A Systematic Review of the Literature. *The Spine Journal*, In Press. <https://doi.org/10.1016/j.spinee.2018.08.011>
6. Shojaei, I., Hendershot, B.D., Ballard, M., Acasio, J.C., Dearth, C.L., Bazrgari, B. Trunk Muscle Forces and Spinal Loads in Persons with Transfemoral Amputation during Sit-to-Stand and Stand-to-Sit Activities. *Clinical Biomechanics*, Under Review. Federal Support Acknowledged.
7. Butowicz, C.M., Krupenevich, R.L., Acasio, J.C., Dearth, C.L., Hendershot, B.D. Relationships between mediolateral trunk-pelvic motion, hip strength, and knee joint moments during gait among persons with lower limb loss. *Clinical Biomechanics*, Under Review. Federal Support Acknowledged.
8. Yoder, A., Silder, A., Farrokhi, S., Dearth, C.L., Hendershot, B.D. Lower Extremity Joint Contributions to Frontal Plane Trunk Dynamics in Persons with Transtibial Amputation. *Clinical Biomechanics*, Under Review. Federal Support Acknowledged.
9. Mahon, C.E., Butowicz, C.M., Dearth, C.L., Hendershot, B.D. Trunk-Pelvic Coordination with Lower-Limb Amputation: Longitudinal Changes in the First Year after Initial Ambulation. *Archives of Physical Medicine and Rehabilitation*, Under Review. Federal Support Acknowledged.

Books or other non-periodical, one-time publications.

Nothing to report.

Other publications, conference papers, and presentations.

1. Butowicz, C.M., Acasio, J.C., Dearth, C.L., Hendershot, B.D. Trunk Muscle Activation Patterns during Walking among Persons with Lower Limb Loss. *World Congress of Biomechanics* (WCB), Dublin, Ireland. Federal Support Acknowledged.
2. Butowicz, C.M., Krupenevich, R.L., Acasio, J.C., Hendershot, B.D. Relationships among Trunk-Pelvic Motions, Hip Strength, and Knee Joint Moments during Gait among Persons with Lower Limb Loss. *World Congress of Biomechanics* (WCB), Dublin, Ireland. Federal Support Acknowledged.
3. Mazzone, B., Farrokhi, S., Hendershot B.D., Watrous, J.R., McCabe, C.T. (2018) Prevalence and Relationship of Low Back Pain and Psychosocial Factors after Lower

Limb Amputation among Wounded Warrior Recovery Project Participants. *Military Health System Research Symposium* (MHSRS), Kissimmee, FL, USA. Federal Support Acknowledged.

4. Acasio, J.C., Butowicz, C.B., Dearth, C.L., Shojaei, I., Bazrgari, B., Hendershot, B.D. (2018) Trunk Muscle Activations, Motions, and Spinal Loads among Persons with Lower Limb Amputation: Influences of Chronic Low Back Pain. *Military Health System Research Symposium* (MHSRS), Kissimmee, FL, USA. Federal Support Acknowledged.
5. Acasio, J.C., Butowicz, C.B., Dearth, C.L., Shojaei, I., Bazrgari, B., Hendershot, B.D. (2018) Trunk Muscle Forces and Spinal Loads while Walking in Persons with Lower Limb Amputation both with and without Chronic Low Back Pain. *American Society of Biomechanics* (ASB), Rochester, MN, USA. Federal Support Acknowledged.
6. Acasio, J.C., Butowicz, C.B., Hendershot, B.D. (2018) Patterns of Erector Spinae Activation and Trunk-Pelvis Kinematics in Persons with Lower Limb Amputation: Influences of Low Back Pain. *American Society of Biomechanics* (ASB), Rochester, MN, USA. Federal Support Acknowledged.
7. Butowicz, C.B., Krupenevich, R.L., Acasio, J.C., Hendershot, B.D. (2018) Influences of Low Back Pain on the Energy Contributions of the Hip and Spine during Gait among Persons with Lower Limb Loss. *American Society of Biomechanics* (ASB), Rochester, MN, USA. Federal Support Acknowledged.
8. Hendershot, B.D., Butowicz, C.B., Krupenevich, R.L., Acasio, J.A., Pruziner, A.L., Miller, R.H., Goldman, S.G., Dearth, C.L. (2018) Toward Optimizing Long-Term Health after Limb Loss: Comprehensive Evaluations of Secondary Health Conditions. 10th *Annual Joint National Capital Region Research Competition*, Bethesda, MD, USA. Federal Support Acknowledged.
9. Shojaei, I., Hendershot, B.D., Ballard, M., Acasio, J.C., Dearth, C.L., Bazrgari, B. (2018) Trunk Muscle Forces and Spinal Loads during Sit-to-Stand and Stand-to-Sit Activities: Differences between Persons with and without Unilateral Transfemoral Amputation. 15th *International Symposium on Computer Methods in Biomechanics and Biomedical Engineering* (CMBBE), Lisbon, Portugal. Federal Support Acknowledged.

- **Website(s) or other Internet site(s)**

Nothing to report.

- **Technologies or techniques**

Nothing to report.

- **Inventions, patent applications, and/or licenses**

Nothing to report.

- **Other Products**

Nothing to report.

7. PARTICIPANTS & OTHER COLLABORATING ORGANIZATIONS

What individuals have worked on the project?

Name:	Bradford Hendershot, PhD
Project Role:	Principal Investigator, EACE/WRNMMC
Nearest person month worked:	2
Contribution to Project:	Provides overall project direction, including: tracking resources, ensuring regulatory compliance, coordinating data collections / analyses, and generating reports.
Funding Support:	Federal Employee

Name:	Babak Bazrgari, PhD
Project Role:	Co-Investigator, Site PI at University of Kentucky
Nearest person month worked:	1
Contribution to Project:	Continues to lead the finite element modeling for all biomechanical data
Funding Support:	

Name:	Courtney Butowicz, PhD
Project Role:	Post-Doctoral Researcher, HJF/WRNMMC
Nearest person month worked:	12
Contribution to Project:	Leads data collection, analysis, and interpretation with direction from the study PI.
Funding Support:	

Name:	Julian Acasio, MS
Project Role:	Research Engineer, HJF/WRNMMC
Nearest person month worked:	10
Contribution to Project:	Assists with data collection, analysis, and interpretation
Funding Support:	

Has there been a change in the active other support of the PD/PI(s) or senior/key personnel since the last reporting period?

Nothing to report.

What other organizations were involved as partners?

Nothing to report.

8. SPECIAL REPORTING REQUIREMENTS

COLLABORATIVE AWARDS: For collaborative awards, independent reports are required from BOTH the Initiating PI and the Collaborating/Partnering PI. A duplicative report is acceptable; however, tasks shall be clearly marked with the responsible PI and research site. A report shall be submitted to <https://ers.amedd.army.mil> for each unique award.

QUAD CHARTS: If applicable, the Quad Chart (available on <https://www.usamraa.army.mil>) should be updated and submitted with attachments.

- 9. APPENDICES:** Attach all appendices that contain information that supplements, clarifies or supports the text. Examples include original copies of journal articles, reprints of manuscripts and abstracts, a curriculum vitae, patent applications, study questionnaires, and surveys, etc.

- Appendix 1: Article published in *Human Movement Science*
- Appendix 2: Article published in *Journal of Biomechanics*
- Appendix 3: Article published in *Journal of Biomechanics*
- Appendix 4: Article published in *Journal of Electromyography and Kinesiology*
- Appendix 5: Article published in *The Spine Journal*
- Appendix 6: Article under review in *Clinical Biomechanics*
- Appendix 7: Article under review in *Clinical Biomechanics*
- Appendix 8: Article under review in *Clinical Biomechanics*
- Appendix 9: Article under review in *Archives of Physical Medicine and Rehabilitation*
- Appendix 10: Abstract presented at *World Congress of Biomechanics*
- Appendix 11: Abstract presented at *World Congress of Biomechanics*
- Appendix 12: Abstract presented at *MHSRS*
- Appendix 13: Abstract presented at *MHSRS*
- Appendix 14: Abstract presented at *ASB*
- Appendix 15: Abstract presented at *ASB*
- Appendix 16: Abstract presented at *ASB*
- Appendix 17: Abstract presented at *NCR Research Competition*
- Appendix 18: Abstract presented at *CMBBE*

Evaluation of Spine Health and Spine Mechanics in Service members with Traumatic Lower Extremity Amputation or Injury

OR130150 - Peer Reviewed Orthopaedic Research Program, Translational Research Award
W81XWH-14-2-0144

PI: Bradford Hendershot, PhD

Org: Henry M. Jackson Foundation

Award Amount: \$652,586

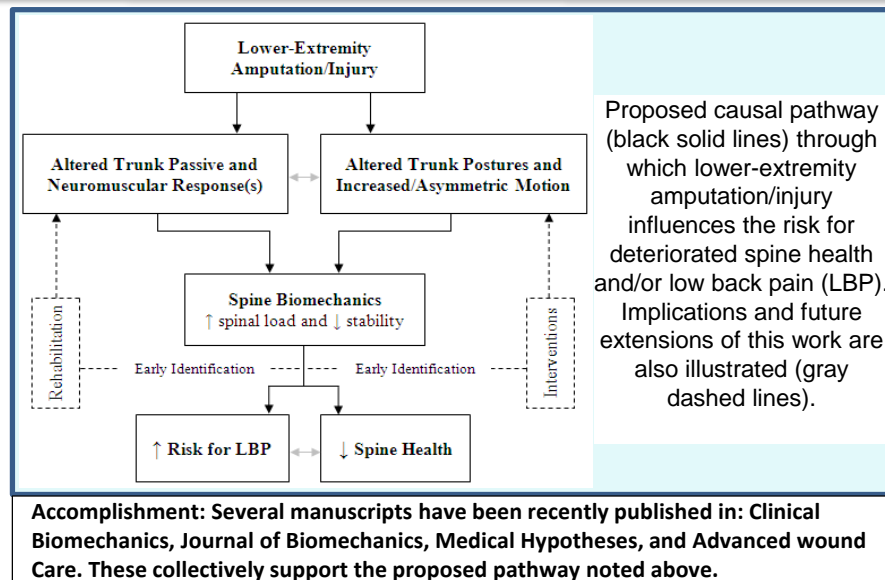


Study/Product Aim(s)

- Quantify alterations in lumbar spinal alignment and inter-segmental motions with traumatic unilateral lower-extremity amputation/injury
- Quantify alterations in spine mechanics (loading and stability) with traumatic unilateral lower-extremity amputation/injury
- Determine the association between spine mechanics (loading and stability) and current spine health with traumatic unilateral lower-extremity amputation/injury

Approach

Spine health and spine mechanics will be evaluated using a novel set of clinical, experimental, and modeling methods with up to four groups differing by the severity of lower-extremity trauma (and presence/severity of low back pain).



Timeline and Cost

Activities	CY	14	15	16	17	18
Submit for IRB approval						
Quantify spinal alignment/motion						
Quantify spine mechanics						
Evaluate spine health and correlate with spine mechanics						
Estimated Budget (\$K)		\$000	\$214	\$230	\$228	

Updated: 22 October, 2018

Goals/Milestones

CY15 Goal – Complete biomechanical collections

- ✓ Obtain IRB and HRPO approvals

CY16 Goal – Quantify spine mechanics

- ✓ Calculate spinal loads and stability using spine model

CY17 Goals – Complete clinical spine evaluations

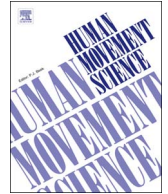
- ✓ Evaluate spine health
- ✓ Associate with spine mechanics (loads)
- ☐ Create a clinical metric for prognosis of spine health in this population
- ☐ Design and submit new research studies or clinical trials stemming from this work

Comments/Challenges/Issues/Concerns: 27 participants have completed data collections. Additional 90-day NCE approved.

Budget Expenditure to Date

Projected Expenditure: \$652,586

Actual Expenditure: \$587,585



Full Length Article

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

Pawel R. Golyski^a, Brad D. Hendershot^{a,b,c,*}

^a Research & Development Service, Department of Rehabilitation, Walter Reed National Military Medical Center, Bethesda, MD, USA

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ARTICLE INFO

Keywords:

Lower limb loss
Coordination
Momentum
Turns
Biomechanics

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. Secondly, 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:

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

where \mathbf{r}_i is the position vector of the segment's center of mass, \mathbf{r}_{Foot} is the position vector of the pivot foot, m_i is the mass of the segment, \mathbf{v}_i is the velocity vector of the segment's center of mass, and \mathbf{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:

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

where \mathbf{I}_i is the moment of inertia tensor for the segment of interest and \mathbf{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-pelvis 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

Spin Turns

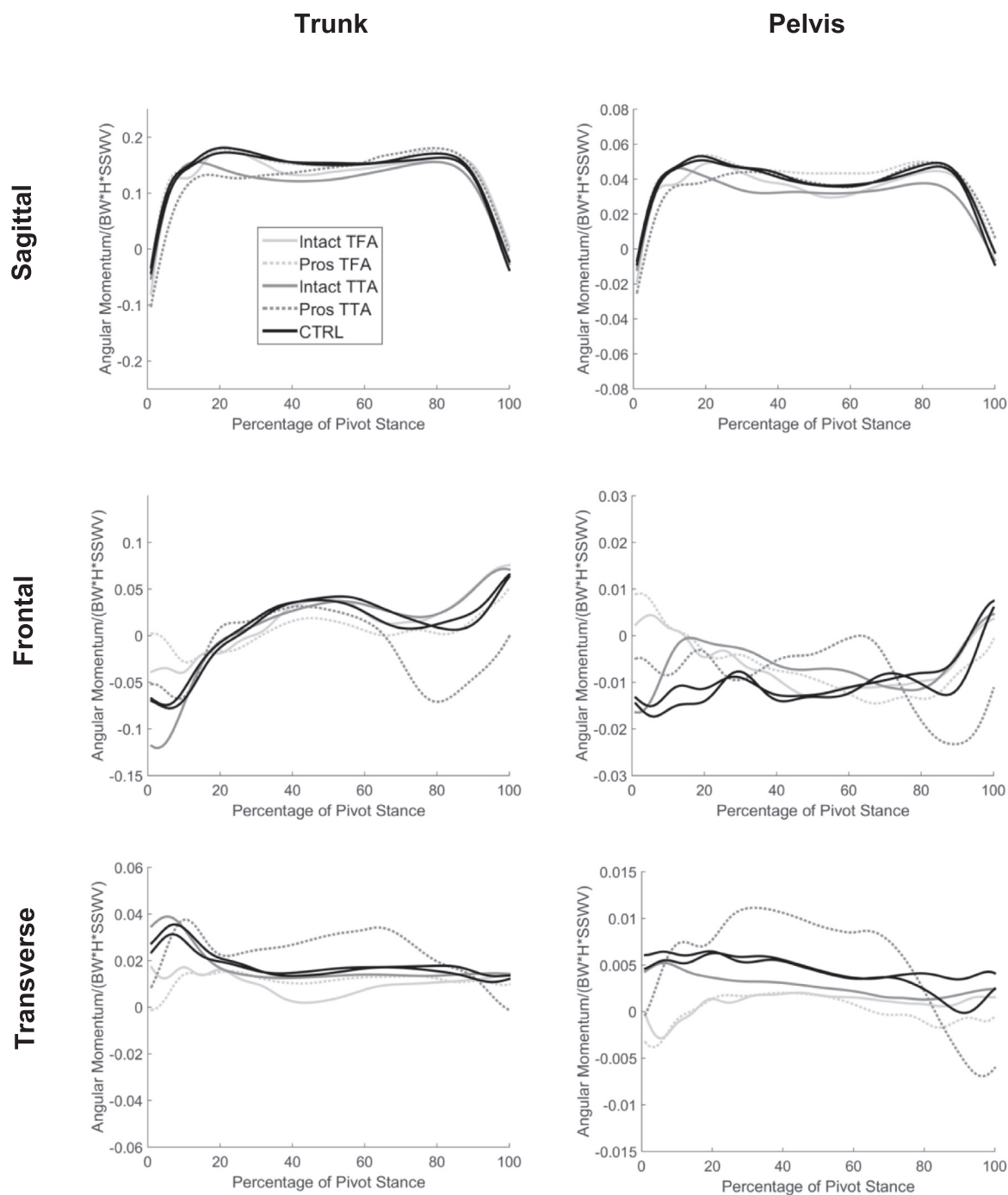


Fig. 1. Ensemble averages of trunk and pelvis translational angular momentum (TAM) in the sagittal, frontal, and transverse planes. Data are normalized by body weight (BW), height (H), and self-selected (in-line) walking velocity (SSWV). The two traces for each turn strategy executed by controls represent 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.

Step Turns

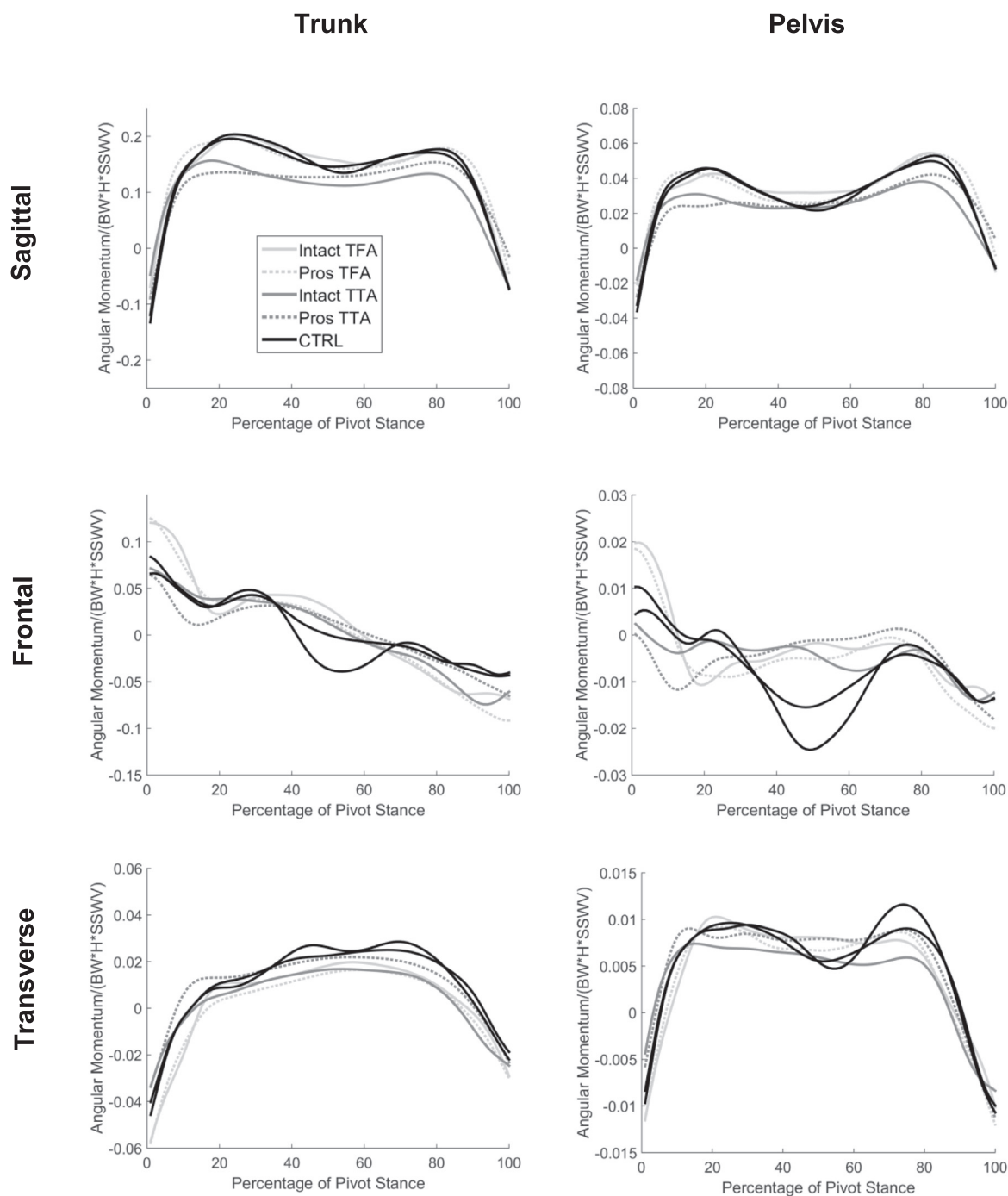


Fig. 1. (continued)

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 $\alpha = 0.05$ and $\alpha = 0.001$ 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-pelvis 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).

Spin Turns

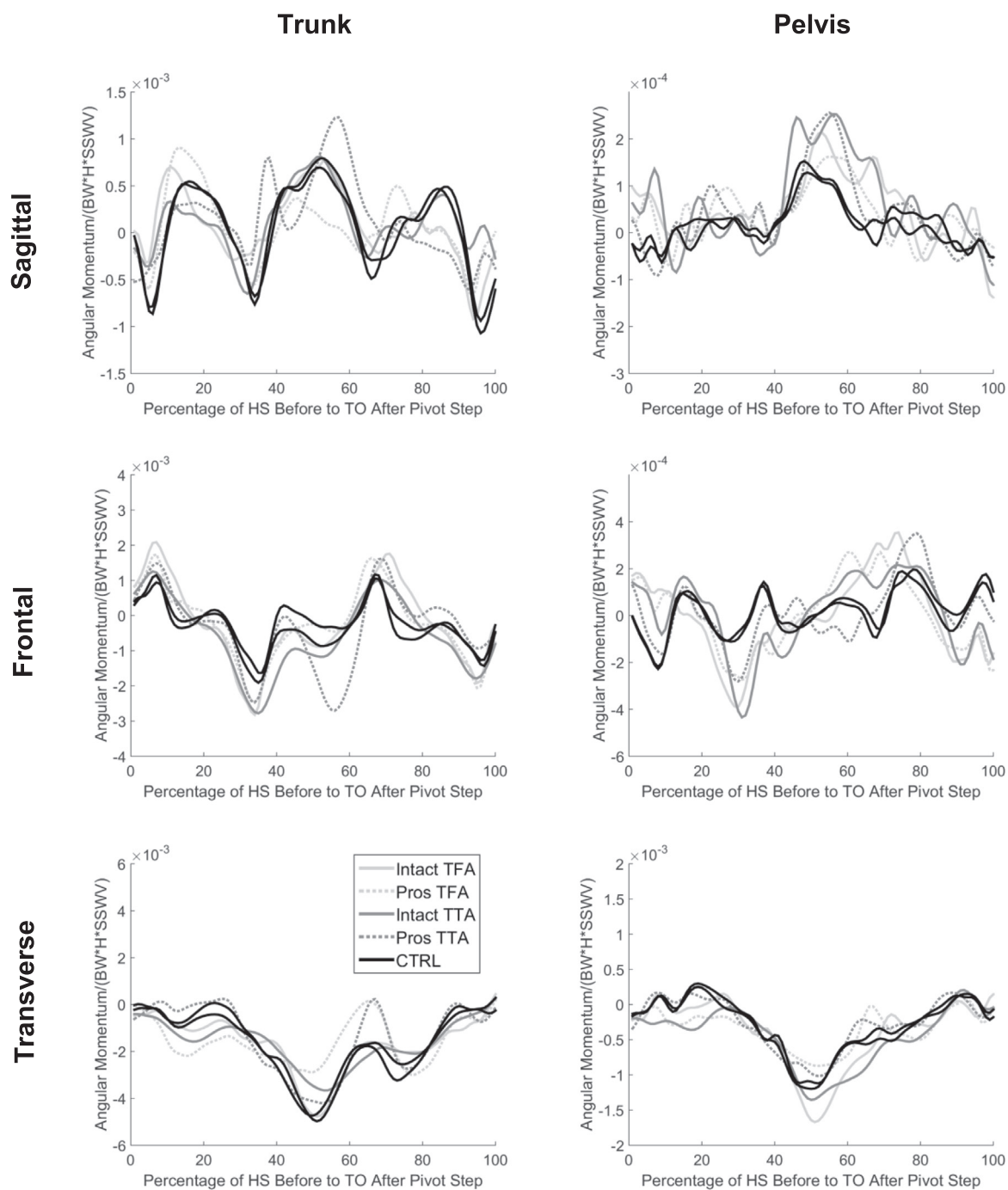


Fig. 2. Ensemble averages of trunk and pelvis rotational angular momentum (RAM) in the sagittal, frontal, and transverse planes during spin/step turns. Data are normalized by body weight (BW), height (H), and self-selected (in-line) walking velocity (SSWV). The two traces for each turn strategy executed 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.

Step Turns

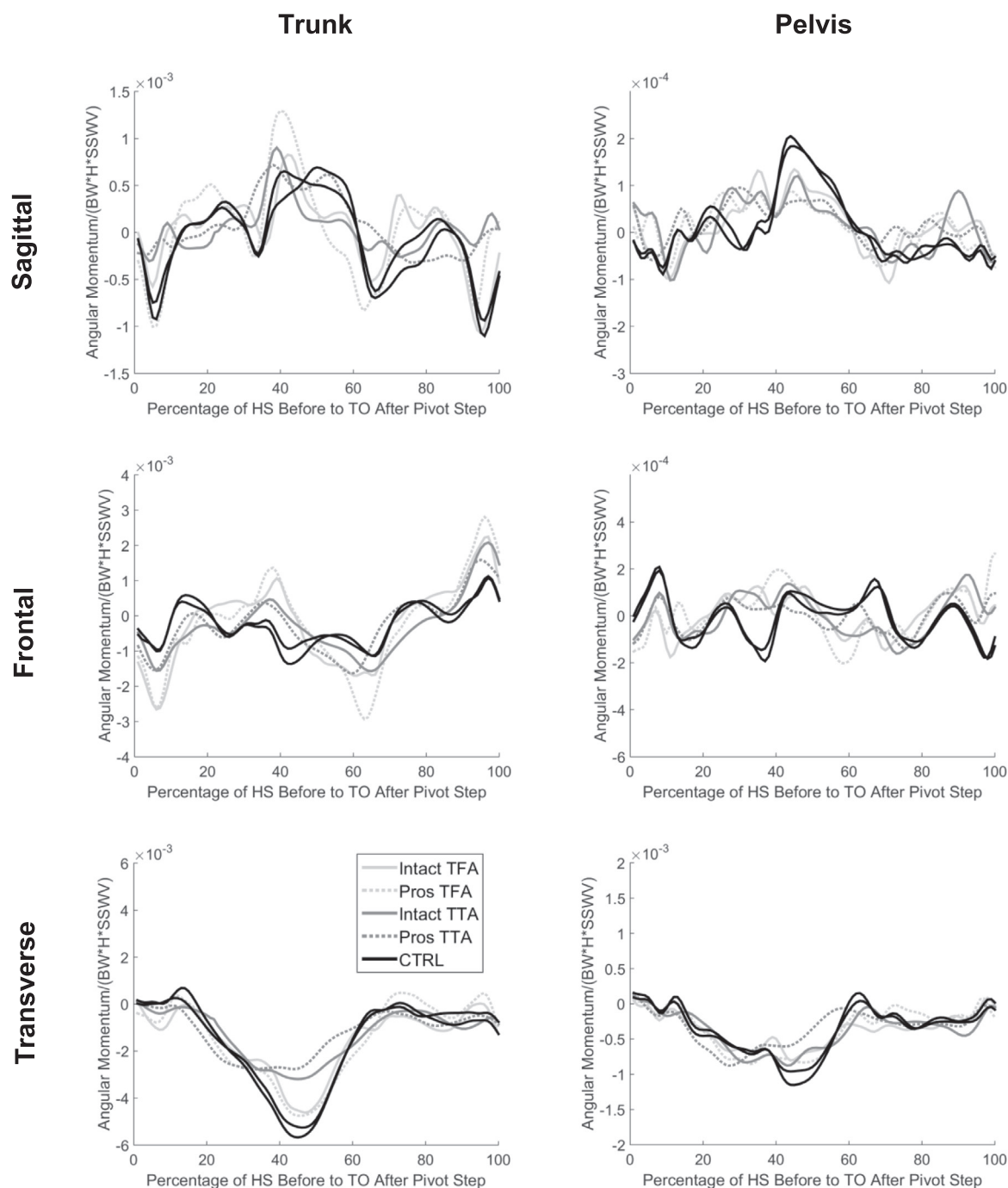


Fig. 2. (continued)

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 $\alpha = 0.05$ 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, Sværre, & Svensson, 1996; Lamothe 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.

Spin Turns

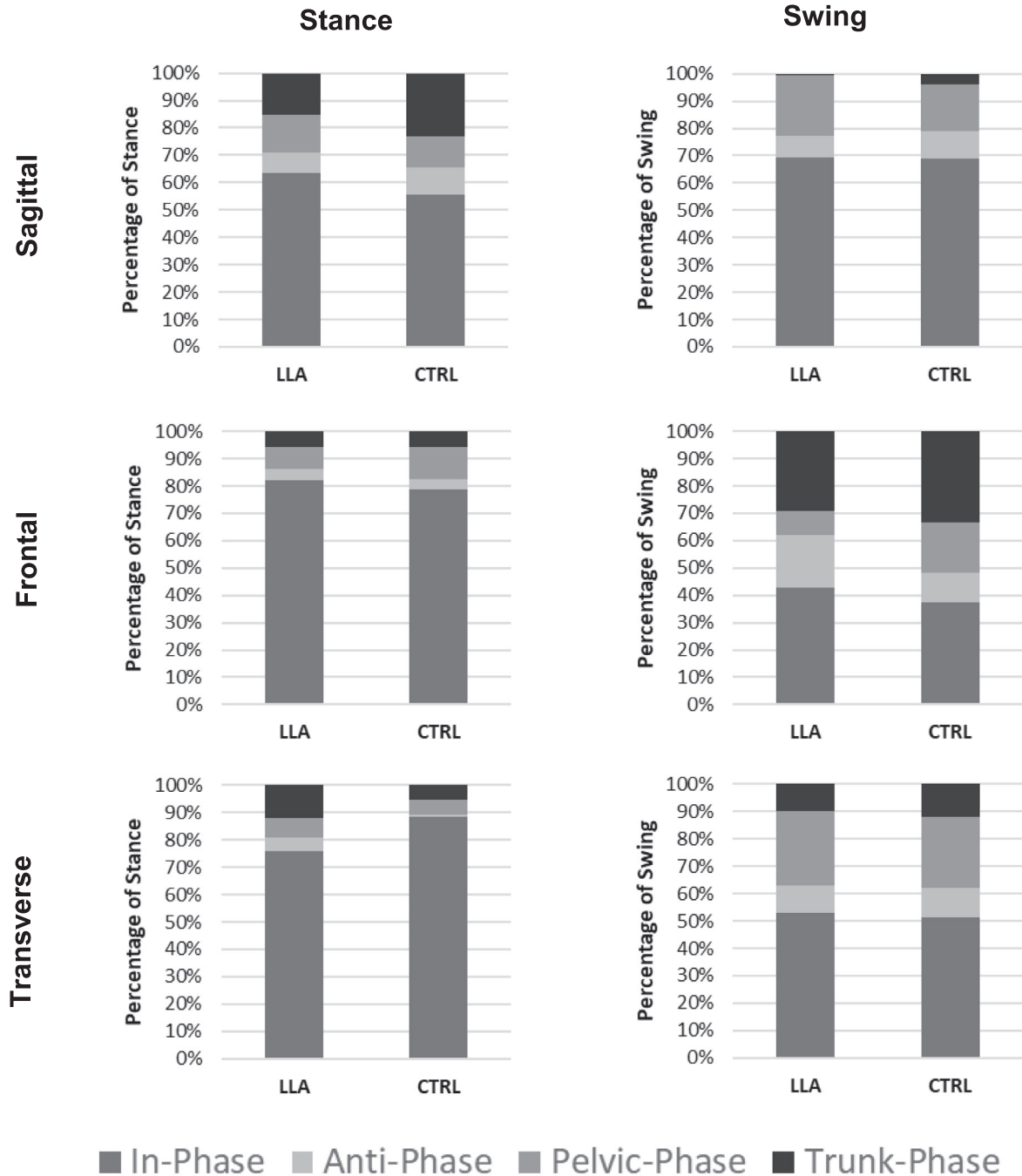


Fig. 3. Proportions of trunk-pelvis coordination modes during pivot stance and swing in each plane among persons with unilateral lower limb amputation (LLA) and uninjured controls (CTRL), by spin and step turns. Significant comparisons (*) between groups were at the $\alpha = 0.05$ level only.

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

Step Turns

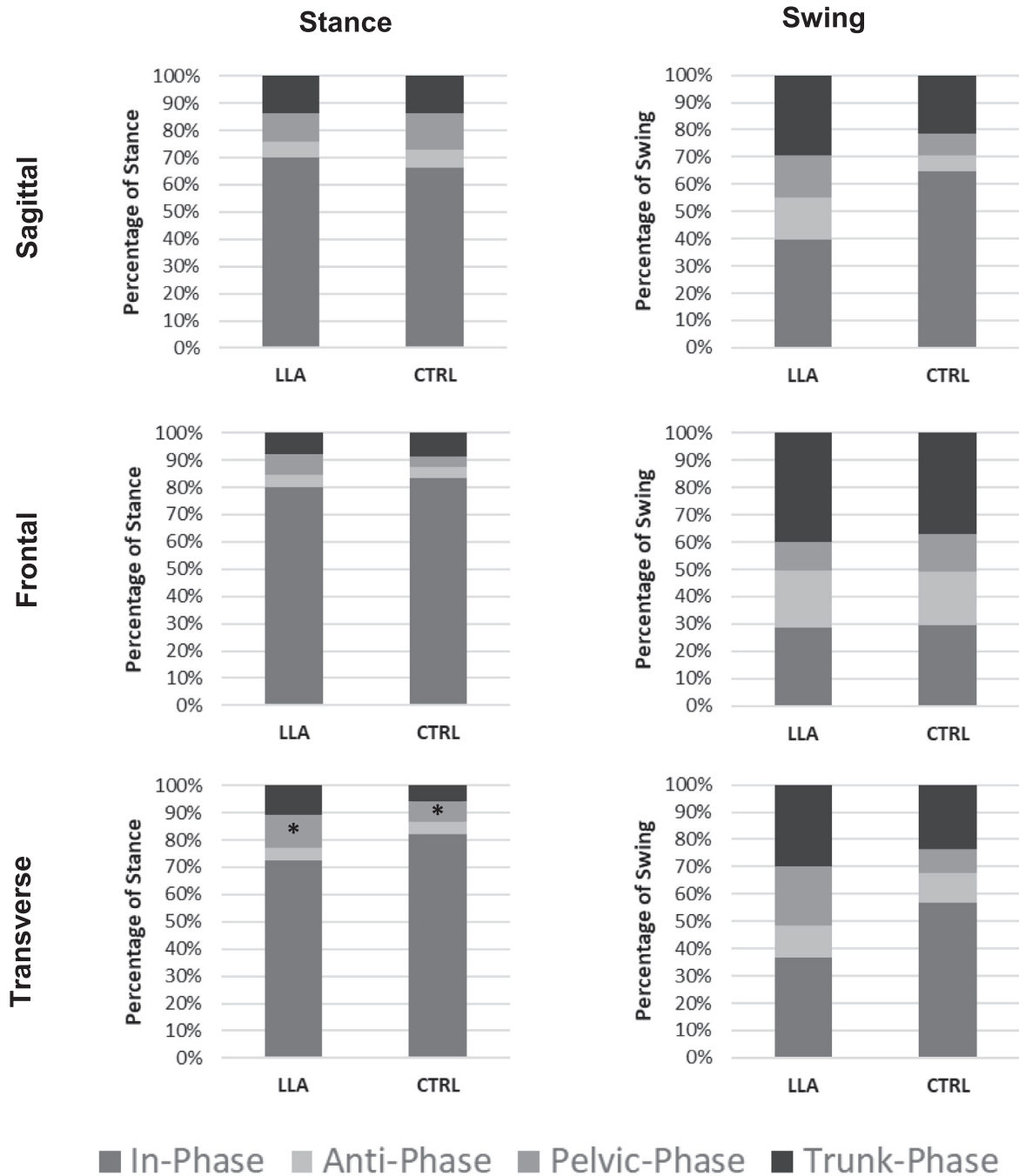


Fig. 3. (continued)

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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Walking speed differentially alters spinal loads in persons with traumatic lower limb amputation

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ABSTRACT

Persons with lower limb amputation (LLA) perceive altered motions of the trunk/pelvis during activities of daily living as contributing factors for low back pain. When walking (at a singular speed), larger trunk motions among persons with vs. without LLA are associated with larger spinal loads; however, modulating walking speed is necessary in daily life and thus understanding the influences of walking speed on spinal loads in persons with LLA is of particular interest here. Three-dimensional trunk-pelvic kinematics, collected during level-ground walking at self-selected (SSW) and two controlled speeds (~ 1.0 and ~ 1.4 m/s), were obtained for seventy-eight participants: 26 with transfemoral and 26 with transtibial amputation, and 26 uninjured controls (CTR). Using a kinematics-driven, non-linear finite element model of the lower back, the resultant compressive and mediolateral/anteroposterior shear loads at the L5/S1 spinal level were estimated. Peak values were extracted and compiled. Despite walking slower at SSW speeds (~ 0.21 m/s), spinal loads were 8–14% larger among persons with transfemoral amputation vs. CTR. Across all participants, peak compressive, mediolateral, and anteroposterior shear loads increased with increasing walking speed. At the fastest (vs. slowest) controlled speed, these increases were respectively 24–84% and 29–77% larger among persons with LLA relative to CTR. Over time, repeated exposures to these increased spinal loads, particularly at faster walking speeds, may contribute to the elevated risk for low back pain among persons with LLA. Future work should more completely characterize relative risk in daily life between persons with vs. without LLA by analyzing additional activities and tissue-level responses.

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1. Introduction

Persons with unilateral lower limb amputation (LLA) – both above and below the knee – commonly report low back pain (Hammarlund et al., 2011; Kulkarni et al., 2005) and perceive altered trunk motions/postures during activities of daily living as primary contributors to its onset and recurrence (Devan et al., 2015). Indeed, altered trunk motion can adversely influence the mechanical environment among spinal structures and tissues within the lower back, especially when the motion occurs in multiple planes simultaneously (Davis and Marras, 2000). Such alterations in the mechanical environment of the lower back may lead to pain if the associated changes in force and/or deformation

experienced in lower back tissues, instantaneously or cumulatively, exceed tolerances (Coenen et al., 2014; Kumar, 2001). The latter is of particular interest here given that many activities of daily living are highly repetitive and thus warrant consideration when assessing cumulative injury risk among persons with LLA.

Walking is a critically important activity of daily living. While not overly demanding on the lower back, walking nevertheless exposes the spine to a large number of loading cycles. For example, healthy adults with a moderately active lifestyle take approximately seven to thirteen thousand steps per day (Tudor-Locke et al., 2011). Although persons with LLA often take fewer steps (\sim half, though dependent on functional classification level; Halsne et al., 2013; Stepien et al., 2007), prior work has reported increases and asymmetries in trunk-pelvic motions during walking among persons with vs. without LLA (Goujon-Pillet et al., 2008; Jaegers et al., 1995). Recently, these differences were associated with larger mechanical demands on the lower back as well as lar-

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ger internal trunk muscle responses and resultant spinal loads (Hendershot and Wolf, 2014; Shojaei et al., 2016; Yoder et al., 2015). Repeated exposures to these elevated demands and loads may thus contribute to the higher prevalence and recurrence of low back pain among persons with LLA. However, these prior studies have predominantly focused on a singular (often self-selected) walking speed. Given that the amplitudes of trunk motion and acceleration increase among uninjured individuals with increasing walking speed (Kavanagh, 2009; Thorstensson et al., 1984), it is important to understand the influences of walking speed on trunk motions and spinal loads in persons with LLA.

Although the selection of an optimal walking speed is often governed by minimizing metabolic costs of transport (e.g., Ralston, 1958), the ability to increase/decrease walking speed remains important for many aspects of daily living (e.g., community ambulation and recreational activities). Modulation of walking speed can be achieved through a variety of temporal-spatial, kinematic, and kinetic mechanisms (Neptune et al., 2008), which are achieved primarily via the ankle plantarflexors during step-step transitions (Jonkers et al., 2009; Requião et al., 2005). Although persons with LLA lack active ankle function (on the prosthetic side), these individuals can typically compensate via other joints within the lower extremity (e.g., the knee or hip; Fey et al., 2010; Silverman et al., 2008). Of particular interest here, persons with LLA also employ a seemingly active trunk movement strategy (Hendershot and Wolf, 2015) that, given its relatively large mass, may differentially alter inertial demands of walking on the lower back and surrounding musculature with changing walking speed. Among uninjured individuals, increases in trunk motion at faster walking speeds have been associated with elevated demands/loads on the low back, albeit modest (Callaghan et al., 1999; Cheng et al., 1998); however, such a relationship has not been evaluated among persons with LLA, wherein there is an increased reliance on these proximal segments. The purpose of this study was therefore to quantify and compare trunk muscle responses and resultant spinal loads among persons with and without LLA across multiple walking speeds. It was hypothesized that, with increasing walking speed, persons with vs. without LLA increase their trunk muscle forces more, hence experiencing larger increases in spinal loads; secondarily, these increases would be largest among persons with more proximal levels of LLA (i.e., transfemoral).

2. Methods

2.1. Experimental procedures

This study retrospectively evaluated biomechanical data from seventy-eight male participants (Table 1) – 26 with unilateral transtibial (TTA), 26 with unilateral transfemoral (TFA) amputation, and 26 uninjured controls (CTR) – walking overground along a 15 m level walkway at one self-selected (SSW) and two additional (controlled) speeds (~1.0 and 1.4 m/s). All persons with LLA were independently ambulatory without the use of assistive

devices (e.g., canes, walkers). Additionally, all amputations were the result of traumatic injuries, and the participants reported no additional underlying musculoskeletal conditions. This retrospective study was approved by Institutional Review Boards of both the Walter Reed National Military Medical Center and University of Kentucky.

Three-dimensional kinematic data of the pelvis and thorax were collected by tracking (120 Hz) reflective markers positioned in the mid-sagittal plane over the S1, T10, and C7 spinous processes, sternal notch, and xiphoid; and bilaterally over the acromion, and the anterior/posterior superior iliac spines. All kinematic data (marker trajectories) were low-pass filtered using a fourth-order, bidirectional filter (cut-off frequency = 6 Hz). Controlled speeds were dictated using an auditory tone (“beep”) that sounded when the horizontal component of the velocity of the sternal notch marker was within 5% of the intended speed. Multiple passes were performed at each speed such that ~10 complete gait cycles could be obtained.

2.2. Dependent measures and analyses

Kinematic data was calculated and analyzed using Visual3D (C-Motion, Germantown, MD, USA) and custom MATLAB (Mathworks, Inc., Natick, MA, USA) scripts. Global trunk and pelvis angles, as well as pelvis center of mass position, were normalized and averaged over each stride. Relative trunk-pelvic angles were similarly calculated. Trunk-pelvic ranges of motion (ROM) were calculated as the difference between the maximum and minimum relative trunk-pelvic angles in all three planes.

To estimate trunk muscle responses and resultant spinal loads, these kinematic data were used as inputs to a non-linear finite element model of the spine with an optimization-based iterative procedure (Bazrgari et al., 2007), previously validated in a variety of dynamics tasks (Bazrgari et al., 2008a, 2008b, and 2009), covering a range of trunk motions and postures. The sagittally symmetric model is composed of six rigid elements representing the thorax and each lumbar vertebrae (L1–L5) along with six non-linear flexible beam elements representing the intervertebral discs/ligaments between T12 and S1. Mass and inertial properties were distributed along the spine according to reported ratios. Fifty-six muscles were represented in the model: 46 muscles connecting the individual lumbar vertebrae to the pelvis (i.e., local) and 10 muscles connecting the thoracic spine/rib cage to the pelvis (i.e., global).

Muscle forces are estimated via a heuristic optimization of equilibrium across the lumbar spine (via changing lumbar segmental kinematics) to satisfy a cost function that minimizes the sum of squared muscle stresses across all 56 muscles. A custom MATLAB (Mathworks, Inc., Natick, MA, USA) script was used to control the optimization procedure whereas a finite element software package (ABAQUS; version 6.13, Dassault Systemes Simulia, Providence, RI, USA) was used to estimate muscle forces and associated spinal loads within the non-linear FE model.

Rather than comparing the individual forces in each of the 56 muscles, the summation forces in all local and global muscles were calculated, hereby referred to as “local” and “global” muscle force. Similarly, rather than comparing spinal loads for all lumbar levels, loads (i.e., compression, as well as anteroposterior [A-P] and mediolateral shear [M-L]) were compiled from the L5/S1 spinal level (i.e., the level that usually experiences the maximum spinal loads). From all outcomes, peak values were extracted and evaluated using a linear mixed-model analysis of variance (between factor = group; within factor = speed). Participants were considered random effects with the correlation among repeated measures assumed to follow a compound symmetry model. Statistical signif-

Table 1
Mean (standard deviation) participant demographics by group: uninjured controls (CTR), persons with unilateral transtibial amputation (TTA), and persons with unilateral transfemoral amputation (TFA). The duration of time elapsed between injury and biomechanical testing is also indicated. Note, there were no significant ($P > .27$) group-level differences in these measures.

	CTR (n = 26)	TTA (n = 26)	TFA (n = 26)
Age (yr)	28.0 (4.7)	28.2 (6.6)	32.3 (8.8)
Stature (cm)	167.8 (6.6)	177.9 (6.1)	176.5 (6.5)
Body mass (kg)	85.7 (12.7)	88.7 (11.2)	84.0 (13.2)
Time (months)	N/A	13.6 (16.9)	36.0 (78.7)

icance was concluded at $P < .05$. All data are reported as means (standard deviations).

3. Results

Controlled walking speeds were not different ($P = .91$) between all three groups at 0.99 (0.05) m/s and 1.42 (0.09) m/s for the “slow” and “fast” conditions, respectively. However, SSW speeds differed between groups; CTR (1.41 (0.15) m/s) and persons with TTA (1.35 (0.14) m/s) were faster ($P < .001$) than persons with TFA (1.24 (0.14) m/s).

Overall, trunk-pelvic ROM were larger ($P < .001$) among persons with TFA and TTA vs. CTR (Table 2). With increasing speed, trunk ROM among persons with TFA increased ($P = .004$) in the sagittal plane; increases in the frontal and transverse planes were not different ($P > .27$) between groups.

Peak global muscle forces tended ($P = .07$) to increase with increasing speed, but these were not different ($P > .22$) between groups at each speed. However, there was a significant ($P = .035$) group \times speed interaction on peak local muscle forces; specifically, peak local muscle forces were larger among persons with TFA vs. TTA and CTR only at the fastest speed (Table 3/Fig. 1).

Peak A-P and M-L shear, as well as peak compression, all increased ($P < .001$) with increasing walking speed. There was a significant group \times speed interaction on both A-P ($P = .02$) and M-L ($P = .002$) shear forces; at the fastest speed, these were larger among persons with TFA and TTA vs. CTR. Similarly, there was a significant ($P = .003$) group \times speed interaction on peak compression; at the fastest speed, compression forces were larger among persons with TFA and TTA vs. CTR (Table 3/Fig. 1).

4. Discussion

This study assessed the influences of walking speed on trunk muscle responses and spinal loads in persons with and without LLA. As expected, both trunk muscle forces and spinal loads increased with increasing walking speed; however, these increases were generally larger among persons with LLA vs. CTR (supporting our primary hypothesis). Additionally, with the exception of lateral shear, spinal loads were larger among persons with TFA vs. TTA (partially supporting our secondary hypothesis).

Altered trunk muscle recruitment has been related to subjective (internal) factors, such as the presence of pain (Lamoth et al., 2006; van der Hulst et al., 2010), as well as changes in external demands, such as increasing walking speed (Anders et al., 2007). In activities

involving trunk motion around neutral postures, such as during walking, trunk muscle forces contribute substantially to spinal loads (due to minimal passive tissue contributions; Panjabi, 2003). The amplitudes of trunk motions and accelerations increase with increasing walking speed (Kavanagh, 2009; Thorstensson et al., 1984); associated alterations in inter-planar coupling suggest the importance of efficient neuromuscular control of global trunk motions. At faster speeds, trunk motions tend to become larger/faster and thus the demands on and resultant responses from trunk muscles generally increase as well (4.4–8.3% across speeds ranging from 0.4 to 1.5 m/s; Anders et al., 2007, Callaghan et al., 1999, van der Hulst et al., 2010). Moreover, these responses tend to differ slightly depending on the specific muscle of interest (i.e., global vs local stabilizer), whereby the local (vs. global) stabilizers are much lower in activation magnitude at slower speeds but increase more substantially at faster speeds (Anders et al., 2007). Although not different between groups, the global muscle forces reported herein tended to increase with increasing walking speed. However, increases in local muscle forces at the faster walking speeds among persons with TFA suggest a larger stabilizing response. Considering their respective anatomical and biomechanical differences, global trunk muscles (i.e., spanning the thorax and pelvis) best contribute to spine equilibrium (in response to external task demands) whereas local trunk muscles (i.e., spanning the lumbar vertebrae and pelvis) are better positioned to provide spine (segmental) stability. The similarities among global muscle forces with alterations in walking speed between person with and without LLA may be an indication of similar speed-related changes in spine equilibrium between the groups; larger increases in local muscle forces in person with LLA (TFA, specifically) at faster speeds suggest a larger stabilizing response.

The largest increases in spinal loads with increasing walking speed among persons with LLA were observed in the A-P direction. In the fastest (vs. slowest) controlled speed, A-P shear forces were respectively 77.1 (31.8), 84.8 (34.5), and 42.1 (24.3)% larger among persons with TFA, TTA, and CTR. Notwithstanding the often complex muscle responses that make direct associations between motion and spinal loads somewhat challenging, these larger increases among persons with LLA are likely due to an altered trunk flexion-extension movement pattern, particularly among persons with TFA (Table 2). This movement pattern likely assists with altering walking speed in the presence of altered lower limb anatomy and function. Such an observation is also consistent with more out-of-phase trunk-pelvic coordination in the sagittal plane as walking speed increases among persons with TFA (Russell Esposito and Wilken, 2014). Moreover, this altered movement pattern likely contributes to larger whole-body angular momentum commonly observed in persons with vs. without LLA at faster walking speeds (Silverman and Neptune, 2011). Previous work has suggested leg motion is the primary contributor to whole-body angular momentum (~60%) while trunk movement contributes little (<10%; Bruijn et al., 2008) in uninjured individuals. However, persons with LLA reduce propulsive forces from the prosthetic limb (Silverman and Neptune, 2011); they are thus unlikely to receive the same contribution to whole-body angular momentum from their legs as an uninjured individual and may have to rely on trunk motion to compensate. While this increased contribution of the trunk may help to regulate whole-body angular momentum and assist in fall prevention, the results herein suggest it may also be contributing to increased injury risk at the lower back.

Persons with LLA tend to self-select walking speeds that are slower than uninjured controls. Given the influences of walking speed on common biomechanical parameters, this presents challenges when designing a study or interpreting its results, particularly as it relates to ecological validity and clinical significance (Astefphen Wilson, 2012). Our prior work specifically selected par-

Table 2

Mean (standard deviation) trunk-pelvis range of motion (ROM) by group and walking speed (SSW = self-selected walking speed, speed 1 \approx 1.0 m/s, speed 2 \approx 1.4 m/s). # = significant difference between controlled speed 1 and 2; * = significant difference relative to CTR (in the same speed condition).

	Trunk-Pelvis ROM (degrees)		
	Sagittal	Frontal	Transverse
CTR			
SSW (1.41 m/s)	2.9 (0.9)	11.9 (3.5)	14.4 (3.9)
Controlled Speed 1	2.7 (0.8)	8.7 (2.9)	11.9 (2.7)
Controlled Speed 2	2.8 (0.9)	12.1 (3.3) [#]	15.3 (4.3) [#]
TTA			
SSW (1.35 m/s)	4.2 (1.4)	10.7 (3.2)	15.5 (2.8)
Controlled Speed 1	3.9 (1.3) [*]	7.8 (2.3)	12.1 (3.0)
Controlled Speed 2	4.4 (1.4) [*]	11.4 (3.1) [#]	16.7 (4.3) [#]
TFA			
SSW (1.24 m/s)	8.7 (2.9) [*]	9.3 (2.5)	15.3 (4.4)
Controlled Speed 1	7.1 (2.6) [*]	7.5 (2.8)	12.9 (3.8)
Controlled Speed 2	10.0 (3.1) ^{#*}	10.1 (3.6) [#]	16.4 (3.7) [#]

Table 3

Mean (standard deviation) muscle forces and spinal loads by group and walking speed (SSW = self-selected walking speed, speed1 \approx 1.0 m/s, speed2 \approx 1.4 m/s). All outcomes are normalized by total body mass (N/kg). # = significant interaction effect between group \times walking speed.

	Peak local muscle force	Peak global muscle force	Peak A-P shear force	Peak M-L shear force	Peak compression
CTR					
SSW (1.41 m/s)	9.3 (1.9)	11.4 (3.7)	5.1 (3.5)	8.8 (3.6)	23.6 (5.9)
Controlled Speed 1	8.2 (1.4)	8.7 (2.9)	3.4 (1.8)	5.2 (2.3)	19.2 (3.7)
Controlled Speed 2	9.7 (2.0)	11.8 (4.1)	4.9 (3.1)	8.7 (3.7)	23.5 (5.3)
TTA					
SSW (1.35 m/s)	9.2 (1.6)	11.9 (3.2)	5.0 (2.3)	8.9 (3.4)	23.7 (5.3)
Controlled Speed 1	8.2 (1.7)	9.2 (4.2)	3.1 (1.4)	6.2 (3.3)	19.9 (5.1)
Controlled Speed 2	10.3 (2.7)	13.9 (5.1)	5.7 (3.3) [#]	10.3 (5.2)	26.5 (7.6) [#]
TFA					
SSW (1.24 m/s)	10.0 (2.7)	13.2 (4.3)	5.7 (2.0)	9.5 (4.1)	25.5 (6.0)
Controlled Speed 1	8.9 (2.6)	11.2 (4.8)	3.7 (1.8)	7.9 (4.3)	22.7 (6.1)
Controlled Speed 2	12.1 (3.1) [#]	14.6 (4.9)	6.6 (4.0) [#]	9.6 (3.8)	29.1 (7.1) [#]

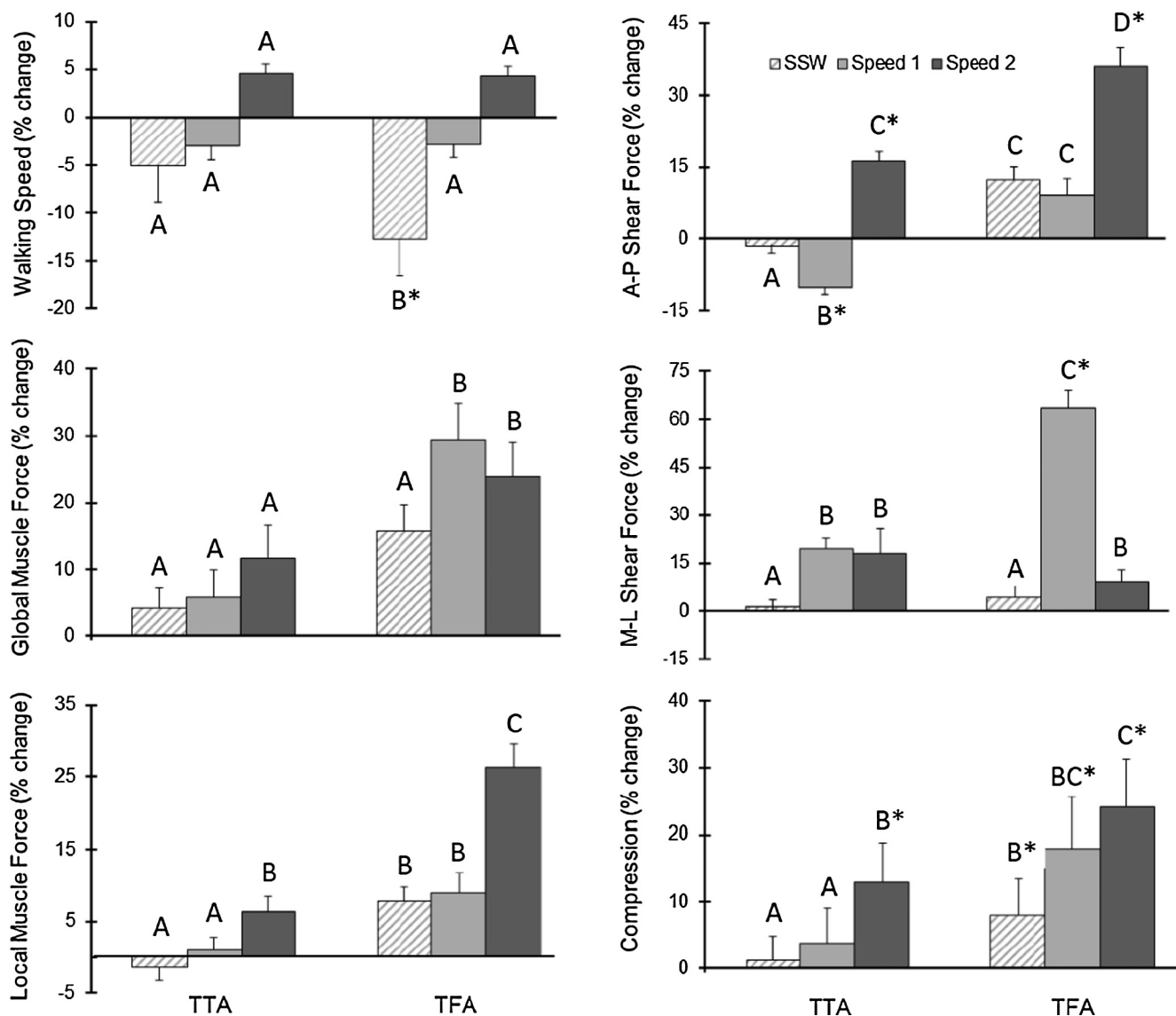


Fig. 1. Mean (standard deviation) percent change in each outcome for both the transtibial (TTA) and transfemoral (TFA) groups with respect to controls at self-selected (SSW) and controlled speeds ("Speed 1" = 1.0 m/s and "Speed 2" = 1.4 m/s). Letters indicate post hoc comparisons and asterisks indicate significant differences relative to controls.

participants by matching SSW speeds post hoc (within 5%; mean \approx 1.35 m/s), and identified 39–60% larger spinal loads in persons with TFA vs. uninjured CTR (Shojaei et al., 2016). In the current study, SSW speeds among persons with TFA were 0.21 (0.14) m/s slower

than uninjured CTR, suggesting smaller trunk inertial contributions to spinal loads in this group. However, larger spinal loads were observed in persons with TFA, despite slower self-selected walking speeds. This highlights the increased contribution of gravitational

demand to spinal loads among persons with TFA due to larger and more asymmetric trunk ROM. One might also presume the relative differences in the slow and fast vs. SSW speeds within each group may require more or less “effort” and thereby differentially affect the relationships reported herein; however, a sensitivity analysis revealed these differences in SSW did not influence any of the dependent measures.

Several limitations should be considered when interpreting the current results. Persons with LLA were young and generally active military personnel with injuries sustained due to trauma. Thus, the results may not be generalizable to all etiologies of amputation (e.g., older or as a result of dysvascular conditions). This study was cross-sectional and the durations of time since injury among persons with LLA were relatively short and highly variable (median = 23 months, range = 6–408 months). As such, it is possible that gait patterns may change (improve or decline) over time and the associated influences on spinal loads with changing walking speed may differ if assessed longitudinally. Moreover, the retrospective nature of this study limited the range of available walking speeds. Additional analyses at slower (i.e. <1.0 m/s) and faster (i.e. >1.4 m/s) walking speeds, or speeds more consistently spaced relative to each participant's SSW, may elucidate additional relationships among spinal loads and walking speed in persons with LLA. Although estimates of these model simulations are highly dependent on the accuracy and reliability of kinematic data, prior work suggests high intra-lab reliability and low standard errors of measurement (Kaufman et al., 2016). Additionally, trunk muscle responses and segmental kinematics were estimated using an optimization procedure assuming similar responses between persons with and without LLA; however, future work is needed with electromyography, imaging modalities, or other modeling techniques to understand these more directly. Such efforts would also support future tissue-level analyses incorporating physiological properties and biological responses (e.g., Lotz et al., 2013). Finally, we did not explicitly include contributions of arm swing in the model (Angelini et al., 2016), though participants were not instructed to alter arm swing and full body kinematics were collected and could be evaluated in subsequent analyses.

Walking is generally not a mechanically demanding activity for the lower back. For example, prior work in uninjured individuals has found peak compressive loads ranging from one to three times body weight when walking over level ground at varying speeds (Callaghan et al., 1999; Cheng et al., 1998). These magnitudes are substantially lower than during other activities (e.g., manual material handling or lifting tasks) and below injury thresholds. However, walking is a highly repetitive task with estimates of 1.5–4 million cycles per year depending on activity level. Thus, we posit that increases in spinal loads among persons with vs. without LLA warrant consideration when assessing injury risk. While tasks involving high physical demands on the lower back or which have a high rate/repetition have been traditionally considered high risk for low back pain (Putz-Anderson et al., 1997), a recent review paper suggests that repetition of low-force tasks seems to result in modest increases in risk; however, surprisingly rapid increases in risk are subsequently observed under high-force tasks (Gallagher and Heberger, 2013). Although not reported here, mean values of each component of spinal load across the entire gait cycle were similarly larger among persons with vs. without LLA, and also tended to increase more with increasing walking speed among persons with LLA, suggesting not just peak loads but the overall mechanical environment is elevated throughout.

In summary, the results presented herein indicate walking speed differentially alters trunk muscle responses and spinal loads among persons with vs. without LLA. Walking faster for persons with LLA was associated with larger increases in the estimated loads among tissues within the spine, regardless of SSW speed.

Over time, repeated exposure to these larger spinal loads during such a common and important activity of daily living may contribute to the elevated risk for low back pain after LLA, particularly due to fatigue failure of spinal tissues. Further work to more completely characterize spinal loads during other activities of daily living is warranted, thereby supporting future clinical recommendations for controlling risk over the longer term.

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Conflict of interest

All authors have no financial or personal relationships with other persons or organizations that might inappropriately influence our work presented therein.

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Short communication

Associations between trunk postural control in walking and unstable sitting at various levels of task demand

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ABSTRACT

Trunk postural control (TPC) has been investigated in several populations and tasks. Previous work observed targeted training of TPC via isolated trunk control tasks may improve performance in other activities (e.g., walking). However, the nature of this relationship remains unknown. We therefore investigated the relationship between TPC, at both the global (i.e., response to finite perturbations) and local (i.e., resistance to continuous perturbations) levels, during walking and unstable sitting, both at varying levels of task demand. Thirteen individuals (11 Male, 2 Female) with no recent history (past 12 months) of illness, injury, or musculoskeletal disorders walked on a dual-belt treadmill at four speeds (−20%, −10%, +10%, and +20% of self-selected walking speed) and completed an unstable sitting task at four levels of chair instability (100, 75, 60, and 45% of an individual's "neutral" stability as defined by the gravitational gradient). Three-dimensional trunk and pelvic kinematics were collected. Tri-planar Lyapunov exponents and sample entropy characterized local TPC. Global TPC was characterized by ranges of motion and, for seated trials, metrics derived from center-of-pressure time series (i.e., path length, 95% confidence ellipse area, mean velocity, and RMS position). No strong or significant correlations ($-0.057 < \rho < 0.206$) were observed between local TPC during walking and unstable sitting tasks. However, global TPC declined in both walking and unstable sitting as task demand increased, with a moderate inter-task relationship ($0.336 < \rho < 0.544$). While the mechanisms regulating local TPC are inherently different, global TPC may be similarly regulated across both tasks, supporting future translation of improvements in TPC between tasks.

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1. Introduction

Physical pathologies including stroke (Verheyden et al., 2006), lower limb loss (Hendershot and Nussbaum, 2013), and low back pain (Lamoth et al., 2006) can adversely influence trunk postural control (TPC). While TPC has been studied extensively, reported measures vary between tasks and specific features of dynamic systems (i.e., global and local). Here, we consider global TPC as the ability of a system to respond to finite ("global") perturbations (e.g., slip or trip), while local TPC is the ability to resist infinitesimal ("local") perturbations (e.g., natural gait fluctuations). During gait,

global TPC has been indirectly quantified by characterizing segmental motions, such as trunk position variability (Dingwell and Marin, 2006) and ranges of motion (ROM). Meanwhile, non-linear measures, including Lyapunov exponents (Asgari et al., 2015; Dingwell and Marin, 2006) and sample entropy (SampEn; Lamoth et al., 2010), have characterized local TPC. During unstable sitting, global TPC is often characterized using metrics derived from center-of-pressure (CoP) time series (Hendershot and Nussbaum, 2013; Radebold et al., 2001) and ROM (Larivière et al., 2015); while local TPC has also been characterized by non-linear analyses of CoP (Larivière et al., 2015; Van Dieën et al., 2010). In both walking and unstable sitting, TPC generally declines with increasing task demand as evidenced by larger values of TPC measures described previously (Dingwell and Marin, 2006; Radebold et al., 2001).

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Altered TPC can adversely influence performance in functional activities (e.g., walking), particularly given the relative mass and position of the trunk. Indeed, TPC deficits are associated with an increased risk of falls (Grimbergen et al., 2008, Tinetti et al., 1988) and musculoskeletal injury (Zazulak et al., 2007). Trunk-specific exercise regimens are therefore often proposed or utilized to help mitigate these risks and, in populations with impaired TPC, incorporated into rehabilitation efforts (e.g., Karthikbabu et al., 2011). Such isolated TPC tasks have been shown to reduce pain and functional disability scores in individuals with LBP (O'Sullivan et al., 1997, Carpes et al., 2008) and improve gait parameters in patients after stroke (Karthikbabu et al., 2011). These observations suggest that improvements to TPC may translate between tasks, but there remains a limited understanding of the effectiveness of such rehabilitation paradigms since the relationship between TPC mechanisms in isolated (e.g., unstable sitting) and functional (e.g., walking) activities has not been investigated thoroughly. Evidence comparing local TPC in two upright tasks (standing and walking) observed little-to-no correlation between them (Kang and Dingwell, 2006). However, only a single level of demand was investigated, and TPC during an isolated task (i.e., unstable sitting) was not determined. We thus explored the relationships between TPC during two distinct tasks, walking and unstable sitting, when both are performed at varying levels of task demand. As TPC has been observed to decrease with increasing demand in both tasks, we hypothesized that increases in respective task demands of walking and unstable sitting would be similarly reflected in decrements to TPC, as evidenced by strong inter-task correlations among TPC measures at each level of demand.

2. Methods

2.1. Study design and procedures

Thirteen participants with no current or recent history of illness, injury, or musculoskeletal disorders within the past 12 months (Table 1) completed walking and unstable sitting trials at varying demand levels. For walking trials, participants walked on an instrumented dual-belt treadmill (Bertec, Columbus, OH) at four speeds relative to self-selected walking speed (SSWS; Table 1), determined from the mean velocity of five over-ground trials across a 15 m walkway: -20% , -10% , $+10\%$, and $+20\%$ SSWS. Relative (vs. absolute) speeds were chosen to better normalize task demand across participants, with the expectation that faster speeds increase demand (Dingwell and Marin, 2006). At each speed, a 30-s acclimation period was provided before two minutes of data collection. For seated trials, participants sat on an unstable chair (Hendershot and Nussbaum, 2013) with eyes open at four levels of instability, relative to an individual's gravitational gradient (∇G): 100, 75, 60, and 45% ∇G (with instability increasing as % ∇G decreased). ∇G was calculated using previously established methods (Slota et al., 2008) and determined neutral seated stability. Participants completed four 60-s trials per condition. However, only the final (i.e., fourth) trial was used for data analyses; the prior three practice trials were used to attenuate learning effects (Van Daele et al., 2007). By the final trial, all participants successfully completed the unstable sitting task (i.e., the seat did not contact the base of support). Participants were asked to keep the chair level and arms crossed throughout trials.

An 18-camera motion capture system (Qualisys, Göteborg, Sweden) collected (120 Hz) 10 surface-marker locations to estimate three-dimensional trunk and pelvic kinematics for all tasks. Markers were placed over the T10 and C7 spinous processes, sternal notch, xiphoid, and bilaterally over the acromion, ASIS, and PSIS. During seated trials, kinetic data were collected (1200 Hz) using a force platform (AMTI, OR6-7-2000, Watertown, MA) mounted beneath the chair. Task and condition order were randomized and counterbalanced, respectively, with 60-s rests provided between trials. Prior to data collection, participants gave informed consent to protocols approved by the local Institutional Review Board.

2.2. Pre-processing

Data were analyzed using Visual3D (C-motion, Germantown, MD) and MATLAB (Mathworks, Natick, MA). Kinematic and kinetic data were low-pass filtered (Butterworth, 4th order, cut-off frequencies 6 and 10 Hz, respectively). Three-dimensional trunk angles (relative to pelvis) were determined using 6DOF inverse dynamics in Visual3D. For each walking trial, 75 strides of data were analyzed and resampled to 101 points per stride (i.e., 0–100% gait cycle). For unstable sitting trials, the first and last five seconds of data were removed to account for initial and anticipatory adjustments respectively.

2.3. Global TPC analyses

For both tasks, tri-planar trunk-pelvic ROM were determined. Though ROM does not directly quantify global TPC (i.e., response to a perturbation), increases in trunk ROM have been observed in populations with impaired TPC such as fall-prone populations (Tinetti et al., 1988, Grimbergen et al., 2008). Thus, though participants were not perturbed in the current protocol, ROM provided an indirect characterization of global TPC. For seated trials CoP path length, mean velocity, 95% confidence ellipse area (CEA), and RMS positions in the anteroposterior and mediolateral directions were also determined (Prieto et al., 1996).

2.4. Local TPC analysis

Maximum short-term Lyapunov exponents (λ_s ; Rosenstein et al., 1993) and SampEn (Richman and Moorman, 2000) were used to characterize local stability of trunk-pelvic angles. λ_s quantifies the rate of convergence/divergence of initially neighboring trajectories. Negative and positive λ_s values respectively indicate convergence (i.e., stability) and divergence (i.e., instability); larger positive values represent a decreased ability to resist local perturbations (i.e., decreased local TPC). Here, tri-planar λ_s were calculated via state spaces reconstructed from trunk-pelvic angles and their time-delayed copies (Dingwell et al., 2001). Global false nearest neighbor and mutual average information analyses respectively determined embedding dimensions ($m = 6$) and time delays ($\tau = 10$ and $\tau = 100$ samples for walking and seated conditions, respectively).

Unlike λ_s , SampEn does not directly characterize the response to local perturbations. Rather, it characterizes the prevalence of local perturbations within the system by quantifying its regularity (Richman and Moorman, 2000). Larger values of SampEn indicate

Table 1
Mean (standard deviation) participant demographic information and self-selected walking speeds (SSWS).

N	Age (years)	Stature (cm)	Mass (kg)	SSWS (m/s)
13 (11 M, 2 F)	28.7 (7.2)	177.1 (6.3)	74.6 (11.4)	1.46 (0.18)

low regularity (i.e., high prevalence of local perturbations) while lower values indicate high regularity (i.e., low prevalence of local perturbations). Similar to λ_s , SampEn was determined via state-spaces reconstructed from trunk-pelvic angles. For SampEn calculations, state-spaces were reconstructed with $m = 2$ (Yentes et al., 2013).

2.5. Statistical analyses

Single-factor, repeated-measures ANOVAs (SPSS Inc., Chicago, IL) assessed the effect of task demand (i.e., speed or % ∇G) on each outcome measure, with significance concluded when $P < 0.05$. Linear correlation analyses related local and global TPC measures between tasks (e.g., $m \lambda_s$, walking vs. λ_s , seated) using Spearman's rho (ρ) as data were not normally distributed. Correlation strength was assessed qualitatively (Portney and Watkins, 2009): 0–0.25 (little or no relationship), 0.25–0.50 (weak-moderate), 0.50–0.75 (moderate-strong), and >0.75 (strong-excellent).

3. Results

3.1. Walking

λ_s increased with increasing walking speed in all planes (Table 2). SampEn increased with speed in the sagittal and transverse planes. Although only approaching significance, SampEn also increased in the frontal plane. Sagittal and frontal plane trunk-pelvic ROM were similar between speeds, but transverse plane ROM increased with walking speed.

3.2. Unstable sitting

All CoP-based metrics were inversely related with % ∇G . In all planes, λ_s remained similar across % ∇G levels. While not statisti-

cally significant, SampEn tended to decrease with % ∇G in the transverse plane. Decreasing % ∇G led to increased sagittal and frontal plane ROM (Table 2).

3.3. Correlation analyses

No strong or significant inter-task correlations were observed in local TPC measures (i.e., SampEn and λ_s). However, measures of global TPC were weakly-to-moderately correlated (Fig. 1). Transverse plane ROM while walking was correlated with sagittal ($\rho = 0.424$, $P = 0.002$) and frontal plane ($\rho = 0.433$, $P = 0.001$) ROM, CEA ($\rho = 0.527$, $P < 0.001$), and both anteroposterior ($\rho = 0.470$, $P < 0.001$) and mediolateral ($\rho = 0.544$, $P < 0.001$) RMS positions while seated. Frontal plane ROM while walking was correlated with frontal plane ROM ($\rho = 0.345$, $P = 0.012$), CEA ($\rho = 0.336$, $P = 0.015$) and mediolateral RMS position ($\rho = 0.417$, $P = 0.002$) while seated. Although sagittal plane ROM while walking was not correlated with seated ROM in any plane, it was weakly correlated with mediolateral RMS position ($\rho = 0.382$, $P = 0.005$) while seated.

4. Discussion

Increases in λ_s , SampEn, and transverse plane trunk ROM with increased walking speed are consistent with previous work (Asgari et al., 2015; Dingwell and Marin, 2006; Lamothe et al., 2010; Van Emmerik et al., 2005), and suggest both local and global TPC declines with increasing task demand. Specifically, the increases in λ_s and SampEn suggest that as walking speed increased, participants became less able to resist local perturbations while simultaneously experiencing more of these perturbations. During unstable sitting trials, the increases in CoP-based measures with decreased chair stability are also consistent with prior reports (e.g., Radebold et al., 2001) and suggest that global TPC declines with increasing task demand during unstable sitting.

Table 2

Mean (standard deviation) ranges of motion (ROM), maximum short-term Lyapunov exponents (λ_s), sample entropy (SampEn), and CoP-based metrics for walking and unstable sitting conditions (SSWS = self-selected walking speed; ∇G = gravitational gradient, AP = anteroposterior, ML = mediolateral, VT = vertical). Asterisks (*) indicate a significant effect of task demand ($P < 0.05$).

Walking							
	–20% SSW	–10% SSW	+10% SSW	+20% SSW	$F_{(3,48)}$	P	η^2
ROM AP (degrees)	10.6 (4.5)	10.23 (3.5)	10.7 (3.1)	11.0 (3.4)	0.174	0.914	0.011
ROM ML (degrees)	16.3 (4.3)	16.82 (4.5)	18.8 (4.8)	18.7 (4.2)	1.448	0.241	0.083
ROM VT (degrees)	16.4 (4.5)	17.63 (5.8)	20.3 (5.4)	22.5 (7.9)	5.057	0.004*	0.240
λ_s AP	1.27 (0.09)	1.31 (0.10)	1.37 (0.15)	1.44 (0.09)	5.333	0.003*	0.250
λ_s ML	1.04 (0.11)	1.10 (0.13)	1.18 (0.16)	1.28 (0.20)	6.116	0.001*	0.278
λ_s VT	1.17 (0.15)	1.27 (0.15)	1.30 (0.12)	1.38 (0.14)	4.880	0.005*	0.234
SampEn AP	0.27 (0.06)	0.28 (0.06)	0.33 (0.08)	0.35 (0.08)	4.401	0.008*	0.216
SampEn ML	0.22 (0.04)	0.23 (0.04)	0.24 (0.04)	0.26 (0.04)	2.708	0.056	0.145
SampEn VT	0.17 (0.03)	0.18 (0.04)	0.21 (0.04)	0.23 (0.04)	7.349	<0.001*	0.315
Unstable Sitting							
	100% ∇G	75% ∇G	60% ∇G	45% ∇G	$F_{(3,48)}$	P	η^2
ROM AP (degrees)	3.8 (2.7)	5.5 (3.20)	5.6 (2.6)	8.4 (3.2)	5.127	0.004*	0.243
ROM ML (degrees)	1.8 (1.1)	2.0 (0.7)	2.4 (1.2)	4.5 (1.0)	19.457	<0.001*	0.549
ROM VT (degrees)	2.8 (1.9)	2.5 (0.6)	2.6 (1.0)	3.3 (1.3)	0.993	0.404	0.058
λ_s AP	0.12 (0.06)	0.10 (0.04)	0.10 (0.02)	0.09 (0.02)	1.235	0.307	0.072
λ_s ML	0.11 (0.04)	0.09 (0.04)	0.11 (0.04)	0.10 (0.03)	0.657	0.583	0.039
λ_s VT	0.13 (0.03)	0.12 (0.04)	0.10 (0.04)	0.11 (0.02)	1.987	0.128	0.110
SampEn AP	0.05 (0.03)	0.05 (0.02)	0.06 (0.02)	0.06 (0.04)	0.656	0.583	0.039
SampEn ML	0.04 (0.02)	0.04 (0.02)	0.05 (0.02)	0.04 (0.02)	0.783	0.784	0.220
SampEn VT	0.04 (0.02)	0.07 (0.02)	0.06 (0.04)	0.04 (0.02)	2.276	0.092	0.124
Path Length (cm)	43.66 (12.27)	45.49 (11.13)	61.90 (19.72)	84.74 (17.79)	15.498	<0.001*	0.569
Mean Velocity (cm/s)	0.84 (0.48)	1.21 (0.59)	2.84 (1.95)	5.26 (2.77)	9.051	<0.001*	0.492
95 %CEA (cm ²)	0.84 (0.26)	0.87 (0.19)	1.17 (0.35)	1.62 (0.34)	18.221	<0.001*	0.361
RMS AP (cm)	0.26 (0.07)	0.34 (0.12)	0.47 (0.15)	0.61 (0.14)	18.221	<0.001*	0.532
RMS ML (cm)	0.18 (0.09)	0.20 (0.06)	0.31 (0.11)	0.46 (0.15)	21.614	<0.001*	0.575

η^2 : small = 0.01, medium = 0.06, large = 0.14 (Cohen 1988).

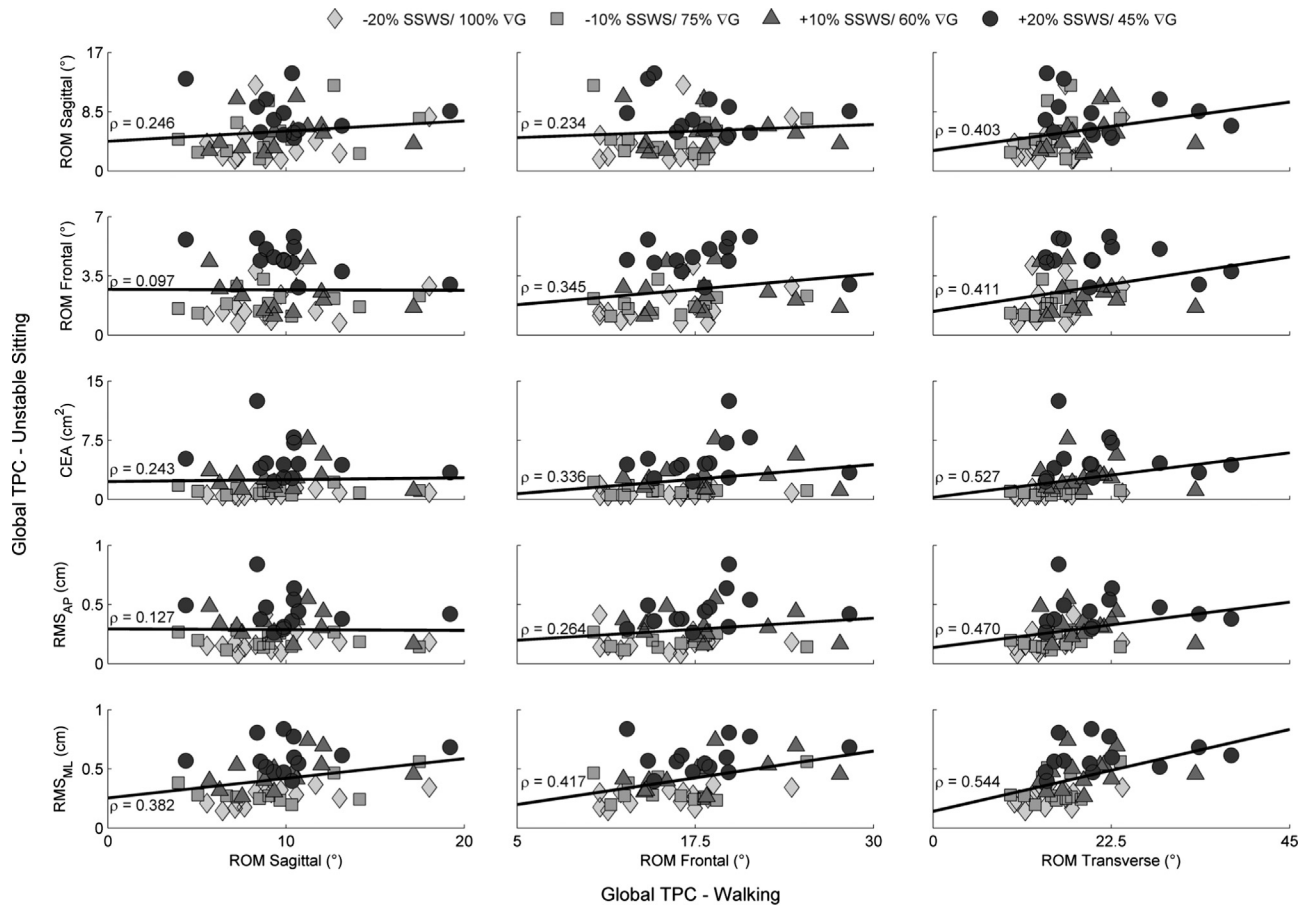


Fig. 1. Trunk-pelvic ranges of motion (ROM), 95% confidence ellipse area (CEA), and RMS positions for unstable sitting plotted against trunk-pelvic ROM while walking. Linear fits and corresponding correlation coefficients (ρ) are displayed. (SSWS = self-selected walking speed; ∇ G = gravitational gradient, AP = anteroposterior, ML = mediolateral).

However, no significant differences were observed in non-linear metrics between levels of instability in seated conditions suggesting local TPC was not affected by increases in task demand. Moreover, and contrary to our hypothesis, no strong correlations were observed between non-linear TPC measures of walking and unstable sitting, suggesting that local TPC mechanisms differ between seated and walking tasks. This is likely due to the relatively static nature of sitting (vs. walking), evidenced by smaller ROM. Furthermore, while the unstable sitting task required dynamic movements to correct for global perturbations, local perturbations and fluctuations of movement were less prominent given the ultimate goal to remain “still”, likely leading to increased local TPC (i.e., smaller λ_s and SampEn) regardless of demand (Table 2). Prior work observed similar results when comparing local stability in static and dynamic tasks (Kang and Dingwell, 2006).

Notably, non-linear metrics exhibited higher variance in seated versus walking tasks. Coefficients of variation for these metrics while walking were 6–24%, and in sitting were 23–64%; high inter-subject variability in the latter was perhaps due to task novelty. Participants may thus have adopted different strategies while adapting to the unstable sitting task, possibly contributing to poor inter-task correlations. Additionally, treadmill (vs. overground) walking can artificially reduce λ_s (Dingwell et al., 2001). Changes in gait parameters also persist for five minutes while acclimating to a dual-belt treadmill (Zeni and Higginson, 2010). Our relatively short acclimation period may therefore have influenced trunk kinematics, though all trials were performed under the same conditions and no order effects were observed ($P > 0.301$).

While transverse plane ROM during unstable sitting remained similar across task demands, this may be a result of the unstable

chair design. The springs mounted beneath the chair, while allowing for the control of instability level, also limit rotations about the vertical axis. Future work could therefore consider using an apparatus that allows for tri-axial rotations (Van Daele et al., 2009). Additionally, although moderate inter-task correlations were observed, future work could also investigate more “extreme” levels (or spacing) of task demand to further assess this relationship.

Despite little evidence relating local TPC in walking and unstable sitting, recent work suggests that a relationship between global TPC mechanisms exists between tasks. Persons with LBP reported decreased pain and functional disability scores after targeted TPC training (Carpes et al., 2008, O’Sullivan et al., 1997) with changes persisting in a 30-week follow-up (O’Sullivan et al., 1997). Trunk-specific training has improved gait parameters (e.g., gait speed, symmetry, etc.) and functional outcomes in patients post-stroke (Karthikbabu et al., 2011), with more pronounced improvements when trunk-specific exercises were performed on an unstable (versus stable) surface (Karthikbabu et al., 2011, Jung et al., 2016). These results, along with the positive correlations among global TPC measures in the present study, establish a tentative relationship by which improvements in TPC via unstable sitting may translate to other functional activities, though it is presently unclear if this relationship persists among individuals with impaired TPC.

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Conflict of Interest

The authors have no financial or personal relationships with other persons or organizations that might inappropriately influence our work presented herein.

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Trunk muscle activation patterns during walking among persons with lower limb loss: Influences of walking speed

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ABSTRACT

Persons with lower limb amputation (LLA) walk with altered trunk-pelvic motions. The underlying trunk muscle activation patterns associated with these motions may provide insight into neuromuscular control strategies post LLA and the increased incidence of low back pain (LBP). Eight males with unilateral LLA and ten able-bodied controls (CTR) walked over ground at 1.0 m/s, 1.3 m/s, 1.6 m/s, and self-selected speeds. Trunk muscle onsets/offsets were determined from electromyographic activity of bilateral thoracic (TES) and lumbar (LES) erector spinae. Trunk-pelvic kinematics were simultaneously recorded. There were no differences in TES onset times between groups; however, LLA demonstrated a second TES onset during mid-to-terminal swing (not seen in CTR), and activation for a larger percentage of the gait cycle. LLA (vs. CTR) demonstrated an earlier onset of LES and activation for a larger percentage of the gait cycle at most speeds. LLA walked with increased frontal plane trunk ROM, and a more in-phase inter-segmental coordination at all speeds. These data collectively suggest that trunk neuromuscular control strategies secondary to LLA are driven by functional needs to generate torque proximally to advance the affected limb during gait, though this strategy may have unintended deleterious consequences such as increasing LBP risk over time.

1. Introduction

Low back pain (LBP) is a common deleterious health condition secondary to lower limb amputation (LLA), with prevalence rates as high as 52–89% (Ehde et al., 2001; Ephraim et al., 2005; Kušljagić et al., 2006). Frequent incidences of LBP are linked to severe physical disability and performance limitations of activities of daily living (Ehde et al., 2001; Kulkarni et al., 2005). The etiology of LBP is typically multifactorial, with physical (i.e., biomechanical), psychological (i.e., anxiety) and social (i.e., support structure) risk factors considered in the holistic approach to understanding the disorder in both able-bodied and individuals with LLA (Farrokhi et al., 2017). Biomechanical factors, specifically, such as altered trunk-pelvic motion and coordination during repetitive/cyclical tasks (i.e., walking), are commonly posited to play a predominant role in LBP risk among persons with LLA (Devan et al., 2014; Esposito and Wilken, 2014; Hendershot and Wolf, 2014).

To assist balance and forward progression during walking, particularly as speed increases, persons with LLA laterally flex the trunk toward the prosthetic limb during ipsilateral stance and minimize relative motion between the trunk and pelvis in the axial plane (Esposito

and Wilken, 2014; Goujon-Pillet et al., 2008; Hendershot and Wolf, 2014; Jaegers et al., 1995). Able-bodied individuals demonstrate increases in trunk motion and muscle activity as walking speed increases (Anders et al., 2007; Callaghan et al., 1999; Saunders et al., 2005). Similarly, increased trunk motion as a function of walking speed is also observed in persons with LLA (Jaegers et al., 1995). As speed increases, axial trunk-pelvic coordination evolves from a synchronous in-phase pattern (i.e., rotations in the same direction) to a more asynchronous anti-phase pattern (i.e., rotation in opposite directions) (Lamoth et al., 2006a). This is comparable between persons with LLA (with and without LBP) and able-bodied individuals; however, persons with LLA demonstrate a more anti-phase coordination pattern in the sagittal plane and a more in-phase coordination pattern in the frontal plane (Esposito and Wilken, 2014). The frontal plane (in-phase) coordination pattern is suggested as a protective “guarding” of the trunk (Lamoth et al., 2002); a compensatory mechanism to increase stability (Esposito and Wilken, 2014). However, the trunk muscle activation patterns driving kinematic outcomes remain unknown.

Coordinated trunk muscle responses maintain equilibrium, maximize energy efficiency, and govern unexpected disturbances

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characteristic of normal environmental conditions (i.e., sudden changes in walking speed) (Lamoth et al., 2006b). For example, the thoracic erector spinae (contralateral to the stance limb) concentrically contract prior to lumbar thoracic spinae, thereby inverting the curvature of the spine toward the swing limb and moving the upper trunk. The subsequent contraction of the lumbar erector spinae then eccentrically controls the trunk while aiding pelvis and swing leg elevation, with the upper aspect of the trunk as the inertial reference (Anderson et al., 2003; Ceccato et al., 2009; Shiavi, 1990). Impaired trunk neuromuscular control may manifest as altered timing and activation of trunk musculature, which is associated with LBP (Arendt-Nielsen et al., 2010; van der Hulst et al., 2010; Vogt et al., 2003). For instance, during the swing phase of gait able-bodied individuals with vs. without LBP demonstrate increased lumbar and thoracic erector spinae activation, earlier onsets of lumbar erector spinae (LES), and increased co-contraction of trunk flexors and extensors (Arendt-Nielsen et al., 2010; Lamoth et al., 2006a; van der Hulst et al., 2010; Vogt et al., 2003). This increased and prolonged activation during swing, when paraspinal muscles are typically silent, suggests a protective mechanism to increase spinal stability. Moreover, increased activation of LES during swing is noted in LBP patients (vs. controls) when walking at faster speeds, suggesting an attempt to increase stiffness and thus spinal stability during speed-dependent perturbations (Lamoth et al., 2006b). While aberrant activation timing and magnitude of trunk musculature during gait are characteristic of persons with LBP, it is unknown whether persons with LLA exhibit similar changes.

Therefore, the first objective of this study was to determine trunk muscle activation patterns and corresponding trunk-pelvic segmental coordination in persons with LLA. We hypothesized that persons with LLA would demonstrate similar muscle activation and segmental coordination patterns to able-bodied individuals with LBP (e.g., earlier, prolonged activation and more in-phase segmental coordination) (Arendt-Nielsen et al., 2010; Lamoth et al., 2006a; van der Hulst et al., 2010; Vogt et al., 2003). The second objective was to determine how these patterns modulate with walking speed, hypothesizing that persons with LLA would demonstrate increased axial and frontal plane segmental motion with corresponding muscular activation patterns as speeds increase.

2. Methods

2.1. Participants

Eight males with unilateral LLA (three transfemoral, five transtibial) and ten able-bodied controls (CTR) participated in this study (Table 1). LLA participants wore energy storage and return feet, microprocessor knees (as relevant), and their prosthesis for 15 h per day (on average via self-report). LLA participants were at least one-year post traumatic injury and could ambulate without the use of an assistive device. LLA participants were excluded if they presented with any of the following: pre-existing spinal pathology or chronic LBP prior to traumatic amputation or injury, co-existing spinal trauma which occurred at the time of the traumatic amputation or injury, diagnosed neurologic deficit(s), including traumatic brain injury, any underlying musculoskeletal

disorders (not including amputation) resulting in functional impairment, upper-extremity amputations above the wrist, and/or pain or discomfort, regardless of cause ($> 3/10$ on a VAS for pain), with 100% weight bearing in socket or which interferes with performance of functional activities. All participants gave written informed consent to procedures approved by the local Institutional Review Board.

2.2. Experimental procedures

Participants walked along a 15 m walkway at four speeds (5 trials in each): 1.0, 1.3, 1.6 m/s, and self-selected walking (SSW) speeds. Non-SSW speeds were enforced within 5% of desired speed via auditory feedback using a custom LabVIEW VI (National Instruments, Austin, TX). Full-body kinematics were recorded by tracking (120 Hz) the locations of 51 surface-markers using an 18-camera motion capture system (Qualisys, Göteborg, Sweden). Electromyographic (EMG) activities of the erector spinae were simultaneously recorded (1200 Hz, Motion Lab Systems, Baton Rouge, LA), pre-amplified per channel with a 500 Hz anti-alias low pass filter, using rectangular bipolar Ag/AgCl surface electrodes. Electrodes were placed bilaterally at the thoracic longissimus (TES, 4 cm lateral to T9) and lumbar iliocostalis (LES, 6 cm lateral to L2) (Willigenburg et al., 2013), with reference electrode placed on the ulnar head. Prior to electrode application, skin was shaved, abraded, and cleaned with alcohol.

2.3. Data analysis

All data were analyzed using Visual3D (C-motion, Germantown, MD) and MATLAB (Mathworks, Natick, MA). Kinematic data were low-pass filtered (Butterworth, 6 Hz). Gait events were determined using previously published methods (Zeni et al., 2008). Three-dimensional trunk and pelvis angles, corresponding ranges of motion (ROM; relative to lab), and angular velocities were calculated. Tri-planar continuous relative phase (CRP) was calculated from angles and angular velocities during each stride (right heel strike to right heel strike) (Hamill et al., 1999). EMG data were normalized to the respective pre-amplification gains, high-pass filtered (Butterworth, cut-off frequency 20 Hz), and full-wave rectified. A root mean square envelope was then calculated using a 50 ms smoothing window (Anders et al., 2007). EMG signals were resampled to 1201 samples per stride and averaged across all strides and participants within each group. For all analyses, the right limb of CTR was used for comparison against the intact and affected limbs of LLA, as there were no differences ($p > 0.05$) between limbs among CTR.

EMG onsets and offsets were determined by visual inspection (Hodges and Bui, 1996a; Saunders et al., 2005); EMG onset was defined as the first upward deviation in EMG amplitude above baseline levels of activity; EMG offset was determined when the level of EMG activity returned to baseline and remained there for $> 5\%$ of the gait cycle (Hodges and Richardson et al., 1999; Saunders et al., 2005). Four reviewers independently analyzed each EMG signal and identified all perceived onsets and offsets of muscle activity within each time series. All occurrences of onset/offset were determined using the same criteria, and named sequentially (i.e., first/second) within the software once identified by each rater. A total of 32 EMG signals (four muscles \times four speeds \times two groups) were analyzed in a random order. Reviewers completed this analysis twice with at least 24 h between analyses and were blinded to analysis results to reduce rater bias (Portney and Watkins et al., 2009; Tenan et al., 2017). Intraclass Correlation Coefficients ($ICC_{3,1}$) were used to determine intra-rater reliability between analyses with values ranging from $ICC_{3,1} = 0.86$ – 0.98 . Analyses with ICC values greater than 0.75 are considered to have “good” reliability (Portney and Watkins, 2009); thus all data used met this criteria and were consistent with prior work (Tenan et al., 2017). The mean of the eight visual detections (two per reviewer) was used to evaluate onset and offset (Hodges and Bui, 1996a; Solnik et al., 2010; Tenan et al.,

Table 1

Mean (standard deviation) participant demographics. * Indicates significant difference ($p < 0.05$) between groups. Abbreviations: ODI: Oswestry Disability Index; CTR: able-bodied controls; LLA: persons with lower limb amputation.

Group	Age (yrs)	Stature (cm)	Mass (kg)	ODI	Time since injury (months)
CTR	29.1 (7.8)*	176.9 (7.0)	74.8 (14.9)	2.4 (4.9)	NA
LLA	37.9 (8.6)	177.9 (8.4)	88.2 (9.3)*	7.3 (11.9)	95.6 (51.4)

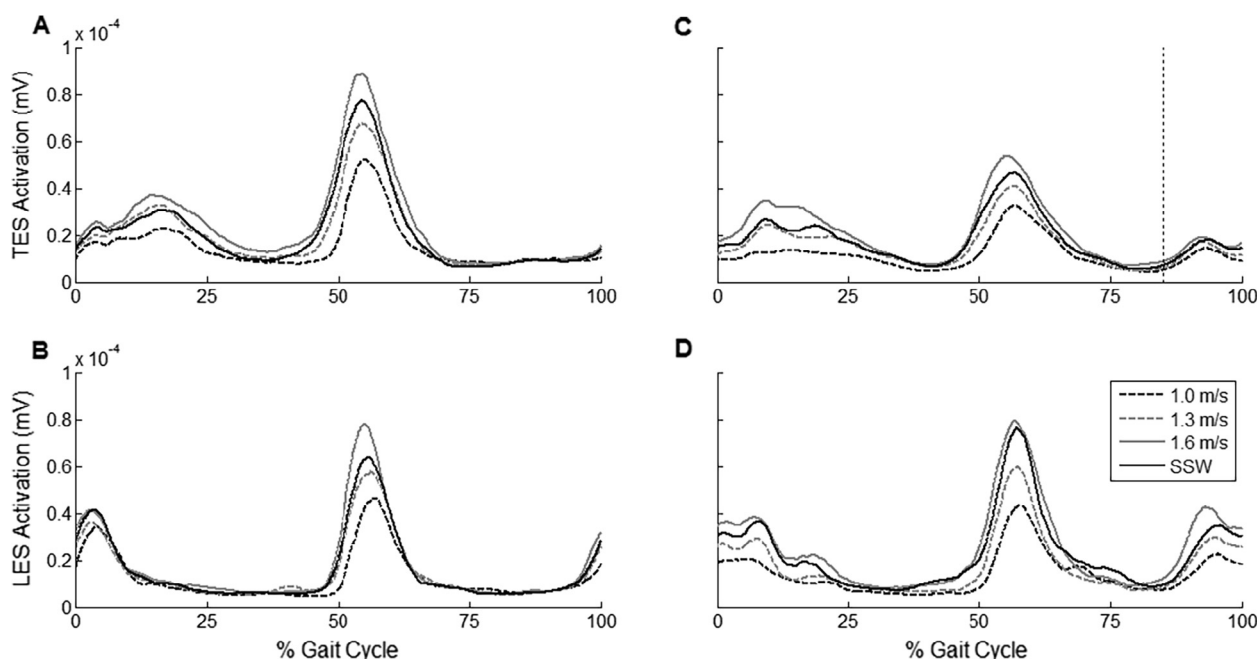


Fig. 1. (A) CTR thoracic erector spinae (TES), (B) CTR lumbar erector spinae (LES), (C) LLA TES, (D) LLA LES at 1.0 m/s, 1.3 m/s, 1.6 m/s, and SSW conditions. Displayed are raw EMG signals of group ensemble means. Vertical dashed line indicates second onset detected in LLA TES that was not demonstrated in CTR TES.

2017). Duration of muscle activation is reported as a percentage of the gait cycle between onset and offset.

Two-factor ANOVAs assessed the effects of group and speed on all outcome measures, with significance set at $p < 0.05$. When a main effect of group was observed, t -tests assessed differences between groups at each speed. Significance was adjusted to account for multiple comparisons ($p < 0.0125$). When a main effect of speed was observed, single-factor ANOVAs were used to assess the effect of speed within each group. If the main effect persisted, t -tests ($p < 0.008$) assessed speed-related differences within a group, with significance adjusted to account for multiple comparisons.

3. Results

3.1. Trunk muscle activation and speed dependency

There were no differences in the first onset of TES between LLA vs. CTR (46% vs. 45% gait cycle) during intact stance at any speed; however, LLA demonstrated a second onset during mid-terminal swing that was not observed in CTR (Fig. 1). A main effect of group was observed in TES activation (as a percentage of the gait cycle) in both intact and affected comparisons. During intact stance, LLA activated TES for a larger percentage of the gait cycle, compared to CTR ($F_{(1,56)} = 103.34$, $p < 0.0001$, $\eta^2 = 0.81$), with pairwise differences ($p < 0.0125$) at all speeds (Table 2). During affected stance, there was a main effect of group for initial TES onset ($F_{(1,56)} = 20.27$, $p < 0.0001$, $\eta^2 = 0.32$); however, LLA onsets were only earlier than CTR in the 1.3 m/s condition ($t = 6.00$, $p < 0.0001$, $d = 3.00$) (Fig. 2). In affected stance, LLA demonstrated TES activation for a larger percentage of the gait cycle ($F_{(1,56)} = 11.98$, $p = 0.001$, $\eta^2 = 0.47$) (Table 3). A main effect of group ($F_{(1,56)} = 34.83$, $p < .0001$, $\eta^2 = 0.68$) was observed during intact stance, with LLA delaying ($p < 0.0125$) the first TES offset during mid stance in all conditions (except SSW). Similar results were noted during affected stance ($F_{(1,56)} = 30.73$, $p < .0001$, $\eta^2 = 0.66$), with LLA exhibiting earlier ($p < 0.0125$) offsets at 1.0 m/s and 1.6 m/s compared to CTR during early stance. However, there were no group differences in the second offset of TES in this comparison.

There was an interaction ($F_{(1,56)} = 4.66$, $p = 0.006$, $\eta^2 = 0.45$) for the second LES onset during intact stance; simple effects contrast

Table 2

Mean (standard deviation) activation as a percentage of intact/right gait cycle. Abbreviations: CTR: controls; LLA: persons with lower limb amputation; TES: thoracic erector spinae; LES: lumbar erector spinae.

Group	Muscle	Speed	Activation (% Gait)	P-value
CTR	TES	1.0	40.8 ± 15.9	0.0001
LLA		1.0	87.0 ± 20.7	
CTR		1.3	48.2 ± 14.8	0.0007
LLA		1.3	86.3 ± 20.2	
CTR		1.6	48.5 ± 18.2	0.0006
LLA		1.6	79.3 ± 7.3	
CTR	LES	SSW	35.5 ± 12.0	< 0.0001
LLA		SSW	79.4 ± 11.1	
CTR	LES	1.0	39.8 ± 4.5	< 0.0001
LLA		1.0	67.5 ± 4.7	
CTR		1.3	44.4 ± 10.1	0.013
LLA		1.3	64.3 ± 17.1	
CTR		1.6	44.8 ± 12.9	0.002
LLA		1.6	65.1 ± 9.1	
CTR	LES	SSW	41.4 ± 8.0	0.001
LLA		SSW	60.1 ± 10.7	

revealed LLA demonstrated an earlier second onset of LES at all speeds. LES was active for a larger percentage of the gait cycle ($F_{(1,56)} = 69.54$, $p < 0.0001$, $\eta^2 = 0.74$) in LLA during intact stance (Table 2). Relative to affected stance, a main effect of group was observed for the first onset of LES ($F_{(1,56)} = 130.29$, $p < 0.0001$, $\eta^2 = 0.84$), with LLA demonstrating an earlier onset of LES compared to CTR at all speeds ($p < 0.0125$). LLA maintained LES activation for a larger percentage of the gait cycle compared to CTR in the affected comparison ($F_{(1,56)} = 38.40$, $p < 0.0001$, $\eta^2 = 0.64$) (Table 3). A main effect of group ($F_{(1,56)} = 20.20$, $p < 0.0001$, $\eta^2 = 0.52$) was noted in the first offset of LES in the intact comparison, with main effects of speed ($F_{(1,56)} = 7.15$, $p < 0.0001$, $\eta^2 = 0.52$) and group ($F_{(1,56)} = 36.45$, $p < 0.0001$, $\eta^2 = 0.63$) for the second offset. Pairwise within group differences ($p < 0.008$) were observed between 1.0 m/s and SSW conditions in CTR and 1.0 m/s vs. 1.6 m/s and 1.0 m/s vs. SSW conditions in LLA. Second offset group differences ($p < 0.0125$) were noted in 1.0 m/s and SSW conditions (Fig. 2). There were main effects of

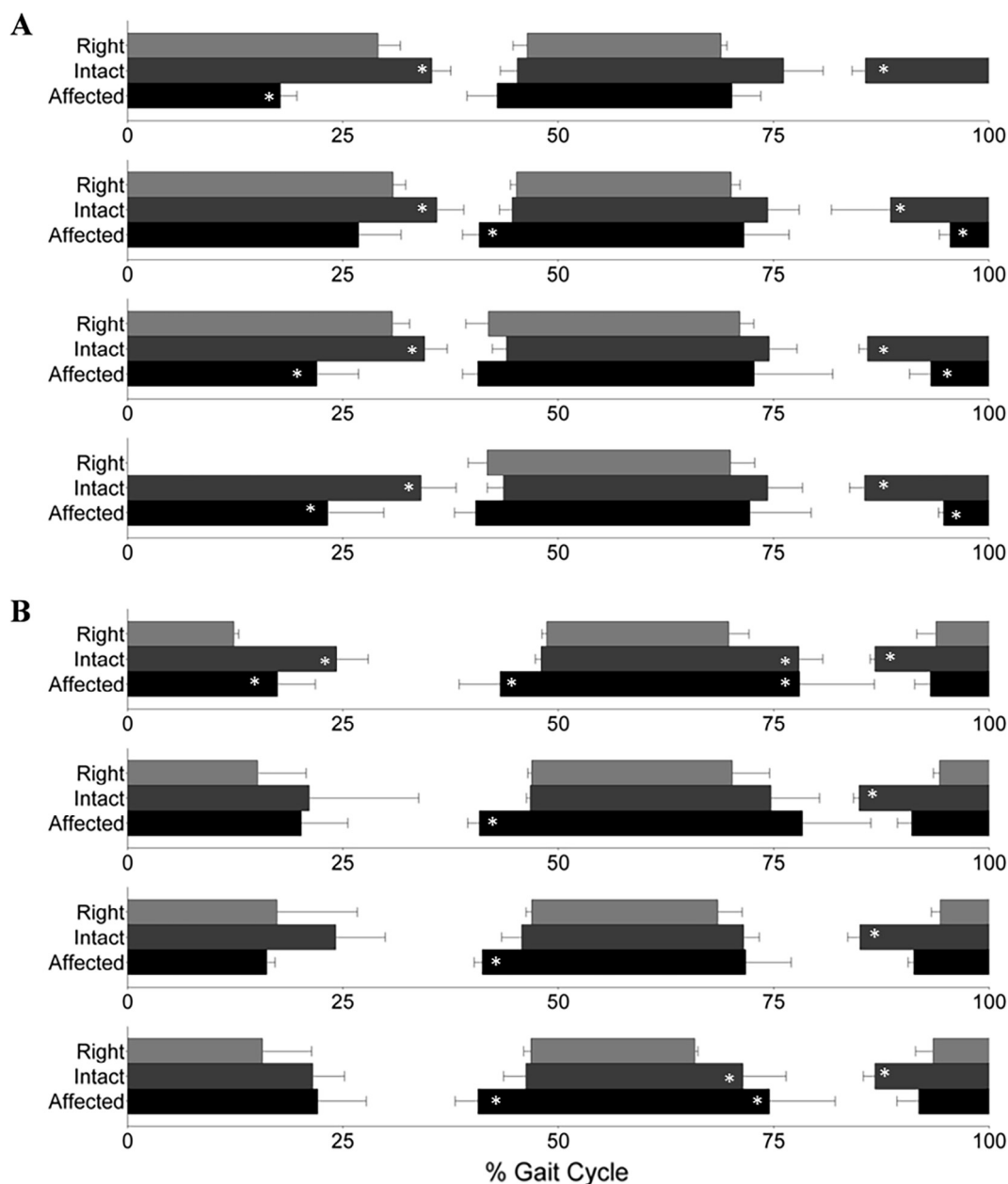


Fig. 2. (A) Thoracic erector spinae activation and (B) lumbar erector spinae activation in controls (right stance) and limb loss group (intact and affected stance). Asterisks (*) indicate significant differences ($p < 0.0125$) between LLA intact vs. CTR and LLA affected vs. CTR.

group for both the first ($F_{(1,56)} = 8.25$, $p = 0.006$, $\eta^2 = 0.36$) and second ($F_{(1,56)} = 24.88$, $p < 0.0001$, $\eta^2 = 0.56$) offset of LES during affected stance, with pairwise differences in the 1.0 m/s condition for the first offset and SSW condition for the second offset.

3.2. Trunk-pelvic motion and segmental coordination

Main effects of group were observed for global trunk ROM in sagittal ($F_{(1,64)} = 5.15$, $p = 0.027$, $\eta^2 = 0.27$), frontal ($F_{(1,64)} = 62.97$, $p < 0.0001$, $\eta^2 = 0.70$), and axial ($F_{(1,64)} = 26.63$, $p < 0.0001$, $\eta^2 = 0.54$) planes. Frontal plane trunk ROM was greater ($p < 0.0125$) among LLA vs. CTR across all speeds (Fig. 3). Axial plane trunk ROM was greater ($p < 0.0125$) in LLA vs. CTR during 1.3 m/s (13.0° vs. 9.6°) and SSW (11.9° vs. 8.5°) conditions (Fig. 3). There were no pairwise differences in the sagittal plane. While there was a main effect of group in the frontal plane ($F_{(1,56)} = 15.73$, $p < 0.0001$, $\eta^2 = 0.44$), there were no pairwise differences in segmental coordination (Fig. 4).

Although not statistically significant, LLA demonstrated a more in-phase coordination pattern than CTR in the frontal plane at all speeds.

4. Discussion

This study aimed to determine trunk muscle activation patterns, corresponding trunk-pelvic inter-segmental coordination, and speed-dependent pattern modulations in persons with LLA. In general, there were no differences in TES onset times between groups; however, LLA demonstrated a second TES onset (not seen in CTR) during mid-to-terminal swing, and TES activation for a larger percentage of the gait cycle. Also, LLA demonstrated an earlier onset of LES and activation for a larger percentage of the gait cycle at most speeds. As expected, persons with LLA consistently walked with increased frontal plane trunk ROM compared to CTR at all speeds. Corresponding CRP in the frontal plane was more in-phase in LLA (vs. CTR), supporting our hypothesis, and consistent with LBP patients and previous work (Esposito and

Table 3

Mean (standard deviation) activation as a percentage of affected/right gait cycle. Abbreviations: CTR: controls; LLA: persons with lower limb amputation; TES: thoracic erector spinae; LES: lumbar erector spinae.

Group	Muscle	Speed	Activation (% Gait)	P-value
CTR	TES	1.0	40.8 ± 15.9	0.771
LLA		1.0	38.8 ± 9.8	
CTR		1.3	48.2 ± 14.8	0.050
LLA		1.3	62.2 ± 11.3	
CTR		1.6	48.5 ± 18.2	0.167
LLA		1.6	61.1 ± 16.5	
CTR	SSW	SSW	35.5 ± 12.0	0.002
LLA		SSW	60.6 ± 14.8	
CTR	LES	1.0	39.8 ± 4.5	0.007
LLA		1.0	59.2 ± 16.9	
CTR		1.3	44.4 ± 10.1	0.003
LLA		1.3	66.8 ± 14.3	
CTR		1.6	44.8 ± 12.9	0.047
LLA		1.6	55.7 ± 5.8	
CTR	SSW	SSW	41.4 ± 8.0	0.005
LLA		SSW	64.2 ± 17.6	

Wilken, 2014; Seay et al., 2011).

4.1. Trunk muscle activation, motion, and speed dependency

Trunk muscles provide segmental stability while controlling trunk motion. As walking speed increases, activation magnitudes of the LES and TES respectively increase at heel strike and/or toe-off (Anders et al., 2007), which increase lumbar stability against braking forces at heel strike and eccentrically control the trunk prior to single-limb stance. In both CTR and LLA groups here, activation patterns were generally similar with increasing walking speed (Fig. 1), consistent with prior work in able-bodied controls (Anders et al., 2007).

Altered trunk muscular activation patterns were also observed in LLA vs. CTR, particularly during intact stance. Here, TES and LES were active at initial contact, remained active longer through early stance in LLA vs. CTR, and corresponded to increased lateral trunk flexion among persons with LLA during this phase of gait. Moving through terminal stance to mid swing, LLA significantly delayed the deactivation of TES compared to CTR; this delayed deactivation was coupled with an increase in lateral trunk flexion toward the affected (opposite) limb and axial rotation (shoulder opposite the intact limb is more forward than CTR and subsequently rotates backwards through stance). In preparation for the next intact heel strike, LLA then reactivate TES (i.e., second onset) during terminal swing, moving the center of mass back toward the intact limb. Of note, although temporal aspects of both stance and swing phases could influence trunk muscle activation patterns and kinematics, there were no significant differences observed between LLA and CTR groups in the current study, thereby mitigating this potential confounder. Contrary to our hypothesis, trunk ROM remained similar across walking speeds. The general invariance of trunk ROM across walking speeds in persons with LLA is not surprising considering global muscle (i.e., TES) activation was not different between speeds (Hendershot et al., 2018). Therefore, as the lumbar iliocostalis stabilizes the spine and thoracic longissimus laterally flexes the trunk, the lack of speed-dependent changes in activation would produce similar lateral trunk motions across speeds. The observed differences in trunk motion between groups are greater than reported minimal detectable change values previously reported for these variables (Wilken et al., 2012), with secondary analyses identifying a significant ($p = 0.01$) difference in step width between LLA and CTR at all speeds. The larger step width in LLA vs. CTR may be an adaptive control strategy to counteract greater lateral trunk motion, thereby increasing lateral stability. The increased activation of global trunk muscles (i.e., larger trunk muscles that span multiple segments to control trunk movement), paired with

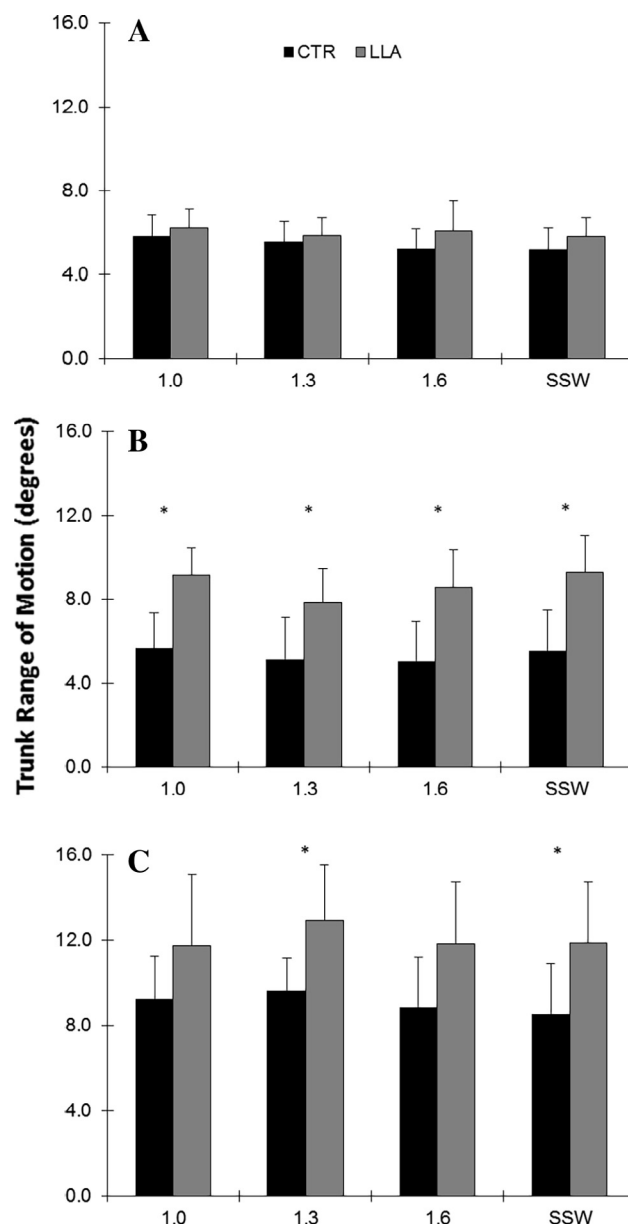


Fig. 3. Sagittal (A), frontal (B), and axial (C) trunk ranges of motion (ROM) at each speed. Compared to CTR, LLA walked with significantly larger trunk ROM at all speeds in the frontal plane, and at 1.3 m/s and 1.6 m/s in the axial plane. Asterisks (*) indicate significant differences ($p < 0.0125$) between groups.

increased lateral trunk flexion and axial rotation, is therefore potentially an adopted control strategy driven by generation of torque proximally to advance the affected limb.

During affected stance, LLA activated LES earlier than CTR at all speeds and maintained activation for a larger percentage of the gait cycle in all but the 1.6 m/s walking speed. The lack of a difference at 1.6 m/s may be due to an increased reliance on momentum to propel the body forward at a speed faster than their normal comfortable pace. These results are consistent with typical lumbar activation patterns in patients with LBP (Lamoth et al., 2006a; Vogt et al., 2003) and characterize the asymmetric gait mechanics and altered control strategies utilized by LLA. These activation patterns support the suggestion that LLA utilize greater TES activation than CTR during intact stance as a compensatory strategy to generate proximally generate torque to account for absent or altered torque generating capabilities distally, which is consistent with previous work (Hendershot and Wolf, 2014). Furthermore, the combination of increased ratio of LES to TES and

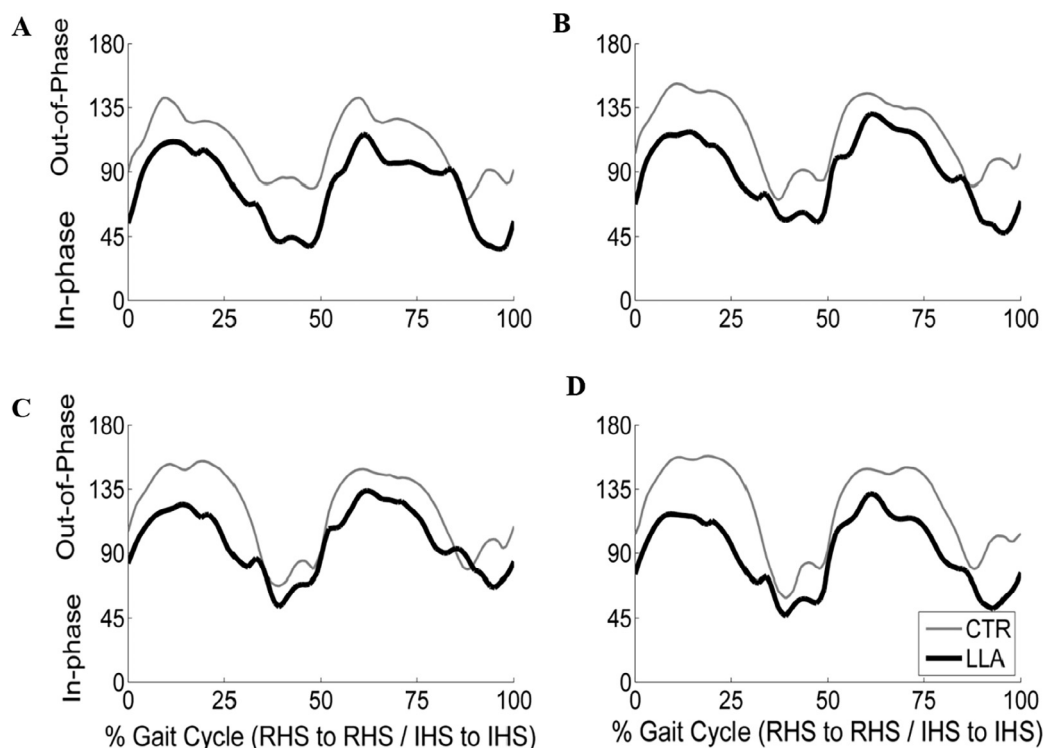


Fig. 4. Frontal plane segmental coordination among LLA (intact-intact foot strike) and CTR (right- right foot strike). (A) 1.0 m/s condition; (B) 1.3 m/s condition; (C) 1.6 m/s condition; (D) Self-selected walking velocity condition. Curves represent ensemble group averages.

overall occurrences of TES activations that are demonstrated by LLA and not CTR explain the more in-phase segmental coordination observed in LLA. Thus, it appears that LLA recruit global trunk musculature, increase global trunk ROM, and adopt a more rigid trunk-pelvic coordination pattern to aid forward propulsion of the affected limb. This strategy generates an increased rotational torque about the lumbar spine that, with repeated exposure, may offer a mechanistic explanation to LBP among persons with LLA.

4.2. Relationship to LBP

This is the first study to characterize trunk muscle activation patterns and corresponding trunk-pelvic motions in LLA during walking. These results suggest that these individuals adopt muscular activation patterns similar to able-bodied patients with LBP. The earlier activation of LES may stabilize the lumbar spine prior to affected limb heel contact and prepare the trunk to move towards the affected limb. As speed-related differences in LES patterns among LLA were most prominent when compared to the 1.0 m/s speed, the pattern observed at 1.0 m/s may be due to decreased muscular demand to advance the affected limb at slower (vs. faster) speeds.

During intact stance, there were no differences between groups in the second activation of TES; however, LLA activated TES significantly earlier than CTR during affected stance in the 1.3 m/s speed. While this is the first study to examine TES at the T9 vertebral level, the results are similar to previous work which found no thoracolumbar muscle activation (T12 vertebral level) of able-bodied controls during walking at a self-selected velocity (Vogt et al., 2003). Interestingly, these participants with chronic LBP demonstrated almost identical thoracolumbar activation patterns to controls (Vogt et al., 2003), which is inconsistent with the activation of TES presented here. These differences in activation patterns during gait could be attributed to the level of erector spinae analyzed, signal analysis differences (i.e., onset determined via computer-based algorithm vs. visual inspection), sample-specific characteristics, and/or functional walking demands of persons with LLA.

Thus, it is possible that persons with LLA adopt unique neuromuscular control strategies that are mediated by functional requirements of locomotion and not by LBP (no/minimal disability; Table 1).

Trunk muscle activation and cyclical motions of the trunk and pelvis during the gait cycle generate loads on the lumbar spine. As walking speed increases, trunk muscle activation, lumbar motion, shear joint reaction forces and moments at the L4/L5 joint increase (Callaghan and McGill et al., 2001). Activation of trunk musculature increases stiffness and joint forces (McGill et al., 2003) and, when coupled with increases in trunk ROM and more in-phase segmental coordination patterns characteristic of LBP and LLA patient populations (Esposito and Wilken, 2014; Lamothe et al., 2006a) may elucidate pathways for LBP development (Hendershot et al., 2018). Of note, muscle activation magnitude was not an objective of the current study; therefore, EMG signals were not normalized to a reference signal (i.e., maximal voluntary contraction) as normalization is not required for temporal-based analyses of EMG data (Di Fabio, 1987; Hodges and Richardson, 1996b). However, the increased relative activation of LES to TES musculature, corresponding trunk “stiffening” strategy, and increased motion in the current study may be associated with an increase in intervertebral joint loads in the lumbar spine among persons with LLA. This control strategy could provide a mechanistic explanation for LBP development among persons with LLA.

4.3. Methodological considerations

The use of both persons with (traumatic) transtibial and/or transfemoral LLA is novel and allows for the generalization of the results to both of these populations; although caution is needed as the results of the current study may not be generalizable to individuals with LLA due to other causes. Previous reports suggest gait mechanics differ between persons with transfemoral and transtibial LLA; however, there were no statistically significant trunk-pelvic kinematic differences between individuals in the current study. While visual inspection is accepted as the “gold standard” of EMG onset detection (Hodges and Bui, 1996a; Solnik

et al., 2010; Tenan et al., 2017), this method is inherently variable and susceptible to human error. Reviewers in the current study demonstrated good test-retest reliability that is consistent with previous work, mitigating this concern. The general lack of significant post hoc differences between groups at each speed may be a function of small sample size and thus type II error. Nevertheless, the large effect sizes observed as group main effects support our hypotheses as well as confidence in the presence of group differences in trunk muscle activation patterns. Future work should also consider assessing anterior trunk muscular activation to determine flexor/extensor co-activation strategies as well as a reference criterion that allows for the comparison of activation magnitudes between groups.

4.4. Conclusions

Persons with LLA demonstrate altered activation of posterior trunk muscles (i.e., earlier onsets and delayed offsets) compared to able-bodied controls during walking. While prior work in able-bodied individuals with LBP has suggested that altered LES activation patterns are a function of poor neuromuscular control and efforts to increase lumbar stability, it appears persons with LLA adopt proximal strategies to advance the affected limb during over-ground walking. However, the differential patterns of muscular activation and trunk-pelvic motions may influence spinal loads and subsequently increase LBP risk. Further work is needed to explicitly relate muscular activation patterns (and magnitudes) with spinal loading during walking in persons with LLA.

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Conflict of interest

The authors declare that there is no conflict of interest.

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Review Article

Low back pain in persons with lower extremity amputation: a systematic review of the literature

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Abstract

BACKGROUND CONTEXT: Lower extremity amputation (LEA) is associated with an elevated risk for development and progression of secondary health conditions. Low back pain (LBP) is one such condition adversely affecting function, independence, and quality of life.

PURPOSE: The purpose of this study was to systematically review the literature to determine the strength of evidence relating the presence and severity of LBP secondary to LEA, thereby supporting the formulation of empirical evidence statements (EESs) to guide practice and future research.

STUDY DESIGN/SETTING: Systematic review of the literature.

METHODS: A systematic review of five databases was conducted followed by evaluation of evidence and synthesis of EESs.

FDA device/drug status: Not applicable

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RESULTS: Seventeen manuscripts were included. From these, eight EESs were synthesized within the following categories: epidemiology, amputation level, function, disability, leg length, posture, spinal kinematics, and osseointegrated prostheses. Only the EES on epidemiology was supported by evidence at the moderate confidence level given support by eight moderate quality studies. The four EESs for amputation level, leg length, posture, and spinal kinematics were supported by evidence at the low confidence level given that each of these statements had some evidence not supporting the statement but ultimately more evidence (and of higher quality) currently supporting the statement. The remaining three EESs that addressed function, disability and osseointegrated prosthetic use were all supported by single studies or had comparable evidence that disagreed with study findings rendering insufficient evidence to support the respective EES.

CONCLUSIONS: Based on the state of the current evidence, appropriate preventative and, particularly, treatment strategies to manage LBP in persons with LEA remain a knowledge gap and an area of future study. © 2018 Elsevier Inc. All rights reserved.

Keywords:

Amputee; Limb loss; Lumbago; Rehabilitation; Spinal pain; Transfemoral; Transtibial.

Introduction

Common musculoskeletal derangements of the spine that contribute to low back pain (LBP) include discogenic dysfunction, facet joint syndrome, sacroiliac joint syndrome, spinal instability, and postural syndrome [1]. There are many factors related to spinal derangements including behavioral, congenital, traumatic, disease processes, and others. These derangements and factors can co-exist, leading to varying levels of disability attributed to LBP. Severe lower extremity trauma, including lower extremity amputation (LEA), can further confound and complicate the clinical presentation and management of LBP [2].

The prevalence of LBP is much higher (52%–89%) among persons with LEA as compared to the general non-amputee population (12%–45%) [3–6]. Low back pain has been considered more bothersome than residual and phantom limb pain [4]. In a cross-sectional survey of persons with LEA ($n=255$), 52% rated their pain as persistent and 25% described their pain as frequent and severely interfering with daily activities [6]. Performance of daily activities with altered anatomy and biomechanics may be related to the development of LBP following LEA [7–10]. Persons with LEA present unique challenges to rehabilitation clinicians managing their LBP. Clinical practice guidelines highlighting efficacious interventions to manage LBP in this group are not available. However, sparse evidence regarding the underlying mechanisms, prevalence, intensity, and management of LBP among those with LEA is available. A systematic review and synthesis of evidence in these areas may inform the development of targeted interventions and lead to improved rehabilitation in this population. Therefore, the purpose of this project was to systematically review and evaluate the literature, and to formulate empirical evidence statements (EESs) regarding the etiology, epidemiology, and management of patients with LEA and LBP.

Materials and methods

Search strategy

A search strategy used in several previous prosthesis- and amputation-related systematic reviews was implemented [11,12]. Five medical literature databases (Medline/Pubmed, CINAHL, EMBASE Elsevier, Web of Science, and Cochrane Clinical Trials Register) were searched on January 1, 2016 based on the following terms (Table 1):

- Primary search terms (target population): *transtibial*, *transfemoral*, *lower extremity*, and *amputee*.
- Secondary search terms (target comorbidity): *low back pain*, *sciatica*, *lumbago*, *back pain*, *back disorder*, *spinal disease*, and *backache*.

Searches were prelimited using the following criteria: English language, abstract available, and peer reviewed. A manual search of included articles' reference lists was also conducted in the event very recent publications or keywords missed important publications in the electronic automated search.

Screening

Resulting references were exported to EndNote (vX7, Thompson, CA, USA) bibliographic citation software. Two reviewers independently screened resulting references' titles, then abstracts, and finally full text articles according to inclusion/exclusion criteria (listed below). Articles were then classified as either (i) pertinent, (ii) not pertinent, or (iii) uncertain pertinence. Full-text articles were then reviewed for all manuscripts classified as pertinent or uncertain pertinence. Disagreements regarding citations of uncertain pertinence were resolved by having a third reviewer independently review full-text articles, discuss, and reach agreement on ultimate inclusion or exclusion.

Inclusion criteria were as follows: (1) peer-reviewed publication; (2) English language; (3) published within the previous 10 years (2006–2016); and (4) study

Table 1

Selected sample search term sets and from the Medline and CINAHL databases

Database	Medline	CINAHL
General search term set	Dorsalgia[tiab] OR exp Back Pain OR backache[tiab] OR (lumbar pain)[tiab] OR coccyx[tiab] OR coccydynia[tiab] OR sciatica[tiab] OR sciatic neuropathy/ OR spondylosis [tiab] OR lumbago[tiab]	"lumbago" OR (MH "Spondylolisthesis") OR (MH "Spondylolysis") OR (MH "Thoracic Vertebrae") OR lumbar N2 vertebra OR (MH "Lumbar Vertebrae") OR "coccydynia" OR "coccyx" OR "sciatica" OR (MH "Sciatica") OR (MH "Coccyx") OR lumbar N5 pain OR lumbar W1 pain OR "backache" OR (MH "Low Back Pain") OR (MH "Back Pain+") OR "dorsalgia"
Amput* string	<p>(((((Lower Extremity"[Mesh] OR lower extrem*[TIAB] OR lower extrem*[OT] OR lower limb*[TIAB] OR lower limb*[OT] OR leg[TIAB] OR leg[OT] OR legs[TIAB] OR legs [OT] OR hip[TIAB] OR hip[OT] OR hips[TIAB] OR hips [OT] OR thigh*[TIAB] OR thigh*[OT] OR foot[TIAB] OR foot[OT] OR feet[TIAB] OR feet[OT] OR "Knee Joint"[Mesh] OR knee[TIAB] OR knee[OT] OR knees[TIAB] OR knees[OT] OR "Ankle Joint"[Mesh] OR ankle*[TIAB] OR ankle*[OT] OR "Femur"[Mesh] OR femur*[TIAB] OR femur*[OT] OR transfemoral[TIAB] OR transfemoral[OT] OR trans-femoral[TIAB] OR trans-femoral[OT] OR "Tibia"[Mesh] OR tibia*[TIAB] OR tibia*[OT] OR transtibial[TIAB] OR transtibial[OT] OR trans-tibial[TIAB] OR trans-tibial [OT] OR transpelvic[TIAB] OR transpelvic[OT] OR trans-pelvic[TIAB] OR trans-pelvic[OT] OR syme's[TIAB] OR syme's[OT] OR symes[TIAB] OR symes[OT]))) AND</p> <p>((("Amputation"[Mesh] OR amput*[TIAB] OR amput*[OT] OR disarticulat*[TIAB] OR disarticulat*[OT] OR hemipelvectom*[TIAB] OR hemipelvectom*[OT] OR "Amputees"[Mesh] OR "Amputation Stumps"[Mesh] OR "Artificial Limbs"[Mesh] OR artificial limb*[TIAB] OR artificial limb*[OT] OR "Amputation, Traumatic"[Mesh] OR "Prostheses and Implants"[Mesh:noexp] OR residual limb*[TIAB] OR residual limb*[OT] OR limb loss*[TIAB] OR limb loss*[OT] OR prosth*[TIAB] OR prosth*[OT] OR stump*[TIAB] OR stump*[OT])))</p>	<p>((MH "Lower Extremity+") OR (TI lower extrem* OR AB lower extrem*) OR (TI lower limb* OR AB lower limb*) OR (TI leg OR AB leg) OR (TI legs OR AB legs) OR (TI hip OR AB hip) OR (TI hips OR AB hips) OR (TI foot OR AB foot) OR (TI feet OR AB feet) OR (MH "Knee Joint+") OR (TI knee OR AB knee) OR (TI knees OR AB knees) OR (MH "Ankle Joint") OR (TI ankle* OR AB ankle*) OR (MH "Femur+") OR (TI femur* OR AB femur*) OR (TI transfemoral OR AB transfemoral) OR (TI trans-femoral OR AB trans-femoral) OR (MH "Tibia") OR (TI tibia* OR AB tibia*) OR (TI transtibial OR AB transtibial) OR (TI trans-tibial OR AB trans-tibial) OR (TI transpelvic OR AB transpelvic) OR (TI trans-pelvic OR AB trans-pelvic) OR (TI syme's OR AB syme's) OR (TI symes OR AB symes) OR (TI thigh* OR AB thigh*))</p> <p>AND</p> <p>((MH "Amputation+") OR (TI amput* OR AB amput*) OR (TI disarticulat* OR AB disarticulat*) OR (TI hemipelvectom* OR AB hemipelvectom*) OR (MH "Amputees") OR (MH "Amputation, Traumatic") OR (MH "Limb Prosthesis") OR (TI prosth* OR AB prosth*) OR (TI artificial limb* OR AB artificial limb*) OR (TI limb loss OR AB limb loss) OR (TI residual limb* OR AB residual limb*) OR (TI stump* OR AB stump*) OR (MH "Prostheses and Implants"))</p>

included subjects with both lower extremity amputation and low back pain;

Exclusion criteria were as follows: publication date outside of the 10-year search window; nonhuman subject research; non-English language; pediatric studies; studies of patients with bilateral lower extremity amputations; case report or case series methodology; studies of digit or partial foot amputation; hypothesis, editorial, classification, or taxonomy papers; thesis, dissertation, and preliminary or pilot level research; and duplicate publication.

Study data

Data from each article including demographic, anthropometric, dependent and independent variables, quantifiable outcomes, and conclusions were entered into an Excel database (Microsoft Corporation, Redmond, WA, USA). These data were verified by a multidisciplinary team (ie, physical therapists, prosthetists, chiropractors, and biomechanists) for completeness and accuracy. Data were assessed for the ability to aggregate for descriptive characteristics (ie, anthropometrics) as well as outcomes (ie kinematic data and pain)

and to calculate effect sizes (Cohen *D*) [13]. To prevent double counting of subject data, data from systematic reviews were not included in the extraction and aggregation.

Quality assessment

The study design and methodologic quality of those publications meeting eligibility criteria were independently assessed by two reviewers according to the American Academy of Orthotists and Prosthetists (AAOP) State-of-the-Science Evidence Report Guideline Protocol [14]. Prior to assessment, the two raters participated in a prelaunch reliability procedure. Test articles were assigned to the two reviewers for assessment. The process was repeated until 90% agreement was attained regarding use of the AAOP rating tool as scored by a third rater. Reviewers discussed pertinent issues until consensus on study design and methodological quality was obtained for the included publications. Each reviewer rated each study according to the AAOP Study Design Classification Scale that describes the type of study design [14]. The State of the Science Conference Quality Assessment Form [14] was used to rate

methodologic quality of studies classified as experimental (E1–E5) or observational (O1–O6). The form identifies 18 potential threats to internal validity and eight potential threats to external validity. In accordance with the guidelines, examples of criteria are provided and described as not applicable for certain study designs; however, guidelines indicate that provided examples are not exhaustive and that reviewers should use their judgment in determining which criteria are not applicable for certain study designs [14]. Threats were evaluated and tabulated.

The internal and external validity of each study was then subjectively rated as “high,” “moderate,” or “low” based on the quantity and importance of threats present. As a guide for rating the internal and external validity separately, studies achieving $\geq 80\%$ of applicable criteria were classified as “high.” If studies achieved $< 80\%$ but $> 50\%$ of applicable criteria, they were classified as “moderate.” Studies achieving $\leq 50\%$ of applicable criteria received a “low” classification. Each study was then given an overall quality of evidence rating of either “high,” “moderate,” or “low” by combining the ratings of internal and external validity as outlined by the AAOP State-of-the-Science Evidence Report Guidelines [14]. The overall ratings from the AAOP State-of-the-Science Evidence Report Guidelines were used in assigning confidence to the developed EESs described in the *Results* section.

Empirical evidence statements

Based on results from the included publications, EESs were developed describing collective findings from included research regarding LBP in persons with LEA. Reviewers rated the level of confidence of each EES as “high,” “moderate,” “low,” or “insufficient,” based on the number of publications contributing to the statement; the methodologic quality of those studies and whether the contributing findings were confirmatory or conflicting [14].

Results

In total, 302 articles were identified from the search (Fig. 1). Ten articles required eligibility determination by the third rater. In most cases, articles requiring the additional review were studies of the spine in a sample of individuals with amputation but the subjects did not have a history of LBP and thus were excluded.

Ultimately, 17 of the original 302 articles met inclusion criteria. Four articles were published in 2009. Between 2006 and 2016, the mean (standard deviation [SD]) number of articles published per year on the subject of LBP in LEA was 1.5 (1.1) (Fig. 2). Study designs included 13 cross-sectional studies, one controlled trial, and three systematic reviews (Table 2). Manuscripts were published predominantly in physical medicine, rehabilitative, and

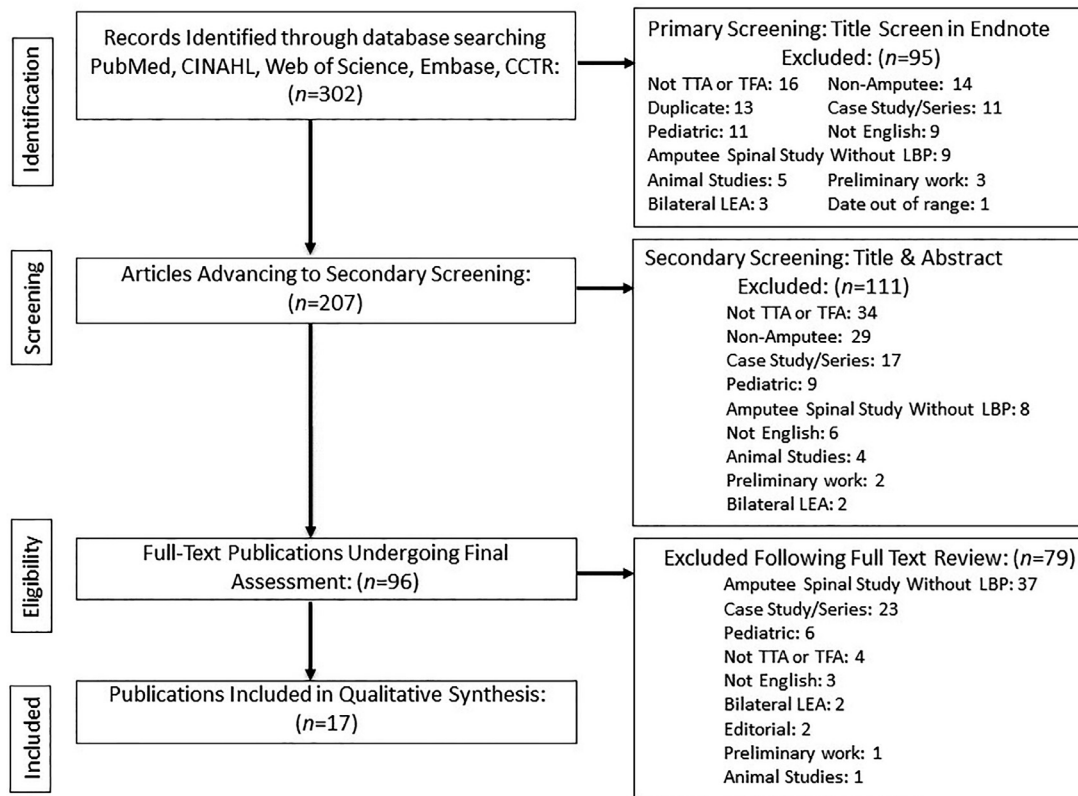


Fig. 1. Results of the literature search and application. TTA, transtibial amputee; TFA, transfemoral amputee; LBP, low back pain; LEA, lower extremity amputation.

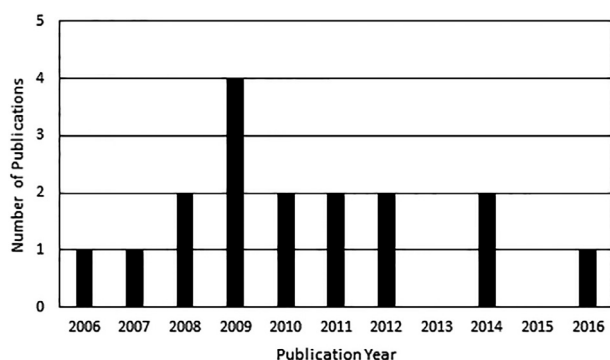


Fig. 2. Included studies by publication year.

Table 2
Distribution of included studies by study design

Study design	Number of publications
Meta-analysis (S ₁)	0
Systematic review (S ₂)	3
Randomized control trial (E ₁)	0
Controlled trial (E ₂)	1
Interrupted time series trial (E ₃)	0
Single subject trial (E ₄)	0
Controlled before and after trial (E ₅)	0
Cohort study (O ₁)	0
Case-control study (O ₂)	0
Cross-sectional study (O ₃)	13
Qualitative study (O ₄)	0
Case series (O ₅)	0
Case study (O ₆)	0
Group consensus (X ₁)	0
Expert opinion (X ₂)	0
Total	17

Table 3
Distribution of the studies per journal

Journal	Number of publications
American Journal of Physical Medicine & Rehabilitation	3
Archives of Physical Medicine & Rehabilitation	1
Bosnian Journal of Basic Medical Sciences	1
British Journal of Surgery	1
Disability & Rehabilitation	1
Gait and Posture	1
Irish Journal of Medical Science	1
Journal of Foot & Ankle Surgery	1
Journal of Orthopaedic Trauma	1
Journal of Rehabilitation Research & Development	4
Military Medicine	1
Prosthetics Orthotics International	1
Total	17

biomechanical journals (Table 3). Due to heterogeneity in sample size and demography, methods, accommodation periods, outcome measures and design, the calculation of effect sizes and meta-analysis was not possible (Table 4).

Subjects

The clinical, patient-oriented studies included a total of 1,260 experimental subjects with a mean (SD; range) sample size of $n = 79$ (94; 8–298). These were subjects with the combination of LEA and LBP. The interquartile mean (IQM) (interquartile range [IQR]) age for experimental study subjects (ie, those with LEA and BP) included in the clinical, patient-oriented studies with adequate data to aggregate was 47.2 years (8.2). The absolute age range of experimental subjects was 16 to 93 years. The height and weight of subjects were only reported in one of the 17 studies. Body mass index was reported or could be calculated in three [15–17] of the 17 studies with an IQM (IQR) of 27.1 m/kg^2 (0.4), which is considered “overweight” according to the Centers for Disease Control and Prevention.

Of studies sufficiently describing subjects for analysis of amputation level and etiology, the majority of amputee subjects (48.3%) were transfemoral level and 36.7% were trans-tibial level. The remaining 15% included partial foot amputation and disarticulations of the ankle, knee, and hip. In terms of amputation etiology, when sufficiently described for detailed analysis, the majority of amputations (89.5%) were caused by trauma. Malignancy, vascular disease, illness, and congenital limb difference were the causes for limb loss in the remaining cases.

Internal validity

Prior to rating, the prelaunch reliability procedure required three test ratings for the two raters to achieve <90% agreement. The most prevalent threats to internal validity in this body of literature include a lack of blinding, lack of use of a control group, no reported consideration for fatigue, learning, accommodation and washout, no reporting of effect size, and lack of random allocation (Table 5). Considering all included studies, the overall assessment favored moderate level internal validity (13/17 studies). Two of 17 had high internal validity [15,18] and two [19,20] had low internal validity. Additionally, seven studies had attrition greater than 20%.

External validity

The greatest threat to external validity was inadequate descriptions of the study samples. Specifically, amputation levels, sociodemography, and anthropometry were not clearly described. Thus, it is difficult to know whether findings are generalizable to the larger population of persons with LEA and LBP. Nevertheless, the majority of the studies (12/17) had high external validity and five had moderate external validity (Table 5).

Table 4
Participant characteristics, primary outcomes and conclusions extracted from included studies

Author (year)	Population (etiology)	Amputation level (sample size)	Mean (range/SD) age (y)	Mean (range/SD) time since amputation (y)	Primary outcome measure(s)	Conclusions
Kusljagic (2006)	Civilian/military (traumatic)	LEA (37)	46 (11)	Not reported	Pain prevalence, psychosocial factors	89% report chronic LBP. Higher levels of social function among civilian versus military
Ebrahimzadeh (2007)	Military (traumatic)	LEA (27)	21 (16–54)*	17 (15–22)	Pain prevalence, psychosocial factors	74% report long-term pain and discomfort
Smith (2008)	Civilian (trauma, PVD, cong, tumor)	TTA (57), TFA (32) KD (4), HD (2), BLEA (10), AD (2)	51 (16–83)	17 (15)	LBP, RLP (periodicity, Frequency, intensity, ADL interference)	48% had LBP w/ 5/10 intensity and reported activity interference of 3.4–3.8/10
Morgenroth (2009)	Civilian/veterans (traumatic)	TFA w/ (9) and w/out (9) LBP	51 (12)	23 (15)	Static and dynamic leg length in single- and double-limb support	Static and dynamic leg length discrepancy not different b/t groups
Taghipour (2009)	Military (traumatic)	LEA (141)	45 (36–63)	22 (20–27)	Pain prevalence, health-related quality of life	LBP most impactful physical condition reducing quality of life
Ebrahimzadeh (2009)	Military (traumatic)	TTA (200)	23 (14–60)*	17 (15–22)	Pain prevalence, psychosocial factors	At long-term follow-up (~17 y), 44% reported LBP and 54% had persistent psychological problems
Morgenroth (2010)	Civilian/veterans (traumatic)	TFA w/ (9) and w/out (8) LBP, CTR (6)	50 (30–77)	23 (3–57)	Lumbar spine kinematics	Larger transverse rotations among LBP group
Reiber (2010)	Military (traumatic)	Vietnam (298), OIF/OEF (283)	61/29	39 (4)/3 (1)	Pain prevalence and psychosocial factors	36%–42% report chronic LBP, 37%–59% w/ PTSD symptoms
Behr (2011)	(Traumatic, vascular)	TFA (14), KD (14), TTA (14)	55 (36–85)	12 (0.6–56)	Pain prevalence and activity level	57% reported LBP that was “troublesome”
Hammarlund (2011)	Not specified (traumatic/tumor)	TFA (19), KD (9), TTA (18)	48 (19–79)	23 (3–58)	Pain prevalence, health-related quality of life (RMDQ, SF36)	87% reported LBP after amputation (vs 20% before); not different by amputation level. Lower quality of life versus normative data
Devan (2012)	Civilian (traumatic)	TFA (145)	57 (18–93)	27 (1–66)	LBP prevalence, physical activity questionnaires	64% reported LBP and 39% reported activity restriction due to LBP
Esposito (2014)	Military (traumatic)	TFA w/ (9) and w/out (7) LBP, CTR (12)	28 (22–39)	2.7 (0.4–5.9)	Trunk-pelvic segmental coordination	Increased coronal in-phase coordination (segmental rigidity) w/ LBP
Hagberg (2014)	Not specified (trauma, tumor, other)	TFA (39)	44 (12)	Not reported	Health-related quality of life (Q-TFA, SF36)	Improved quality of life, prosthesis use, and physical activity 2 y after OI
Fatone (2016)	civilian	TFA w/ (12) and w/out (11) LBP	47 (20–67)	16 (2–41)	Pelvic and spinal kinematics	Reversal of motion pattern in sagittal/transverse plane w/ and w/out LBP

W/, with; w/out, without; Y, year(s); SD, standard deviation; LEA, lower extremity amputation; TTA, transtibial amputation; TFA, transfemoral amputation; LBP, low back pain; OEF, Operation Enduring Freedom; OIF, Operation Iraqi Freedom; RLP, residual limb pain; ADL, activity(ies) of daily living; AD, ankle disarticulation; KD, knee disarticulation; HD, hip disarticulation; BLEA, bilateral lower extremity amputee; OI, osseointegration; CTR, control (subjects); RMDQ, Roland Morris disability questionnaire; SF36, short form 36 health survey; PTSD, post-traumatic stress disorder; Q-TFA, Questionnaire for persons with transfemoral amputation.

*At the time of injury.

Table 5
Internal and external validity of included studies

Author (year)	Study design	Internal validity																		External validity									
		1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	Total	1	2	3	4	5	6	7	8	Total
Kusljagic (2006) ¹	O3	n/a	n/a	n/a	n/a	n/a			n/a	n/a	•	n/a	•	•			•		•	Mod			•	•		•	•	•	Mod
Ebrahimzadeh (2007) ²	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a		n/a	•	•						Low		•	•	•	•	•	•	•	High
Smith (2008) ³	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	•	n/a	•	•			•	•		Mod		•	•	•	•	•	•	•	High
Gailey (2008) ⁴	S2	n/a	n/a	n/a	n/a	n/a	n/a	•	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	Mod		•	n/a	n/a	n/a	•	•	•	High
Morgenroth (2009) ⁵	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	•	•	•	•	•	•	•	•	•	High		•	•	•	•	•	•	•	High
Robbins (2009) ⁶	S2	n/a	n/a	n/a	n/a	n/a	n/a	•	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	Low		•	n/a	n/a	n/a	•	•	•	High
Taghipour (2009) ⁷	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	n/a	•	•	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Ebrahimzadeh (2009) ⁸	O3	n/a	n/a	n/a	n/a	n/a	n/a	•	n/a	n/a	n/a	n/a	n/a	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Morgenroth (2010) ⁹	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	•	n/a	•	•	•	•	•	•	•	High			•	•	•	•	•	•	Mod
Reiber (2010) ¹⁰	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	n/a	•	•	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Behr (2011) ¹¹	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	•	•	•	•	•	•	•	•	•	Mod			•	•	•	•	•	•	Mod
Hammarlund (2011) ¹²	O3	n/a	n/a	n/a	n/a	n/a		•	•	•	•	•	•	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Perkins (2012) ¹³	S2	n/a	n/a	n/a	n/a	n/a	n/a	•	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	n/a	Mod			n/a	n/a	n/a	•	•	•	Mod
Devan (2012) ¹⁴	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	n/a	n/a	n/a	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Russell Esposito (2014) ¹⁵	O3	n/a	n/a	n/a	n/a	n/a			n/a	n/a	•	•	•	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Hagberg (2014) ¹⁶	E2			•	•	n/a		•		n/a	•	•	•	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	High
Fatone (2016) ¹⁷	O3	n/a	n/a	n/a	n/a	n/a		•	n/a	n/a	n/a	•	•	•	•	•	•	•	•	Mod		•	•	•	•	•	•	•	Mod

Scoring: a dot (•) indicates that the criterion was met. A blank space indicates that the criterion was not met. n/a means the item was not applicable in that particular study design. See Table 1 for study design key.

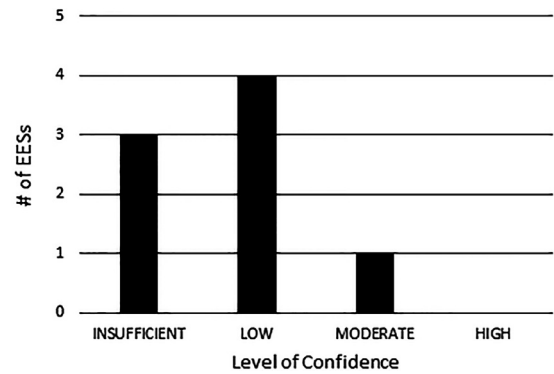


Fig. 3. Distribution of level of confidence for empirical evidence statements (EESs).

Funding analysis

Funding was declared in seven of 17 manuscripts (41%). One manuscript indicated private foundation support, two indicated funding through an academic institution, and four manuscripts indicated funding by way of a governmental sponsor. Within the four US government funded studies, three were supported by the US Department of Veterans Affairs and one by the National Institutes of Health. Three of the studies were funded outside of the United States. Support was not declared in nine manuscripts (53%), whereas one manuscript specifically declared that it was unfunded.

Empirical evidence statements

Eight empirical evidence statements were synthesized. The rate of EES production in this body of evidence was eight EES's per 10 years or a crude rate of 0.8 EES/y. Lack of evidence resulted in insufficient confidence in three of the statements. Evidence supported low confidence in four of the statements and moderate confidence in one EES (Fig. 3). The topical areas covered included epidemiology, amputation level, function, disability, leg length, posture, spinal kinematics, and osseointegrated prosthetic use. Only the epidemiology EES was supported by evidence at the moderate confidence level given that it was supported by eight moderate quality studies. The four EESs for amputation level, leg length, posture, and spinal kinematics were supported by evidence at the low confidence level given that each of these statements had some evidence not supporting the statement but ultimately more evidence (and of higher quality) supporting the statement at the present time. Finally, the remaining three EESs that addressed function, disability, and osseointegrated prosthetic use were all supported by single studies or had comparable evidence (quantity and quality) that disagreed with study findings rendering insufficient evidence to support the respective EES (Table 6).

Discussion

With regard to study design, 13 of the included studies were observational, one was experimental, and three were

Table 6

Empirical evidence statements, indicating level of confidence and category

	Empirical evidence statement (EES)	Supporting studies	Level of confidence	Category
1	Back pain increases following lower extremity amputation	8 × Mod ^{*1, 2, 6, 8, 10, 12–14}	Moderate	Epidemiology
2	Back pain is affected by level of amputation	Support: 4 × Mod ^{*4, 6, 7, 13} Does not support: 3 × Mod ^{3, 11, 12}	Low	Amputation level
3	In persons with lower extremity amputation, function is affected by back pain	Support: 1 × Mod ¹² Does not support: 1 × Mod ¹⁴	Insufficient	Function
4	Frequent bouts of back pain in persons with lower extremity amputation are associated with increased disability	1 × Mod ¹²	Insufficient	Disability
5	Leg length discrepancy is associated with back pain in persons with lower extremity amputation	Support: 1 × Mod ^{*4} Does not support: 1 × High ⁵	Low	Leg length
6	Postural asymmetries and postural control issues are associated with back pain in patients with lower extremity amputation	Support: 1 × Mod ^{*4} Does not support: 1 × High ⁵	Low	Posture
7	Spinal and pelvic kinematics are influenced by low back pain in persons with lower extremity amputation	Support: 2 × Mod ^{9, 15} Does not support: 1 × Mod ¹⁷	Low	Spinal kinematics
8	Back pain is not affected by the use of osseointegrated prosthesis	1 × Mod ¹⁶	Insufficient	Osseointegrated prosthetic use

*Indicates that the supporting reference is or includes a systematic review.

systematic reviews. Although this is a somewhat heterogeneous blend of study designs, a more optimal body of literature inclusive of prospective, randomized controlled intervention trials may have enabled meta-analyses. Internal validity could have been strengthened in the included studies with minor reporting changes as described by standardized criteria [21,22]. For instance, had the included samples been better described (ie, more uniform reporting of anthropometry and demography), effect sizes been reported, and learning/accommodation and fatigue reported, more of the studies would have likely improved their internal validity ratings from low to moderate or moderate to high. Conversely, external validity was generally high in the selected studies that provide confidence that results have clinical importance despite some methodological weaknesses (ie, threats to internal validity).

In this study, the rate of EES production regarding subjects having LEA and LBP was eight EES's per 10 years (crude rate of 0.8 EES/y). This rate of EES production is considerably low compared to other areas of prosthetic literature. For example, in a previous study of lower extremity prosthetic componentry for persons with transtibial amputation [23], the EES production rate was 1.4 EES/y. More problematic is that in the componentry review, this EES production rate was based upon the use of high-quality evidence, whereas the present review of LBP in LEA is low but is based upon all available quality of evidence. Further, although key sponsors, such as NIH, were notably absent as research supporters in the componentry review, all of the studies included were funded (ie, industry, other governmental departments, nonprofit sponsors, etc.). In the present review, the majority of research available, 53%, was unfunded. This identifies numerous potential issues. For

instance, high-quality research can become more difficult to accomplish without adequate funding, which could also decrease interest among researchers in this area. More funding from key research sponsors is needed in this area if the quality and quantity of available research are to become available to fill knowledge gaps related to the care of persons with LEA who suffer from LBP.

Because the majority of this body of evidence was observational by study design, the EESs tended to describe factors that affect or are affected by LBP in persons with LEA. For example, EESs described LBP as increasing following LEA, differences by amputation level, decreased function, increased disability, and altered gait mechanics associated with LBP in persons with LEA. Again, these EESs are predominated by descriptions of LBP and its effects in persons with LEA. Thus, the number of experimental studies was limited to one, minimizing the ability to determine optimal therapeutic intervention choices or their effects in managing persons with LEA who have LBP. Therefore, efficacy of interventions to manage LBP in persons with LEA remains a considerable knowledge gap and an area of future study.

The first EES indicates that LBP increases following lower extremity amputation. Eight moderate quality studies support the statement ultimately providing moderate confidence in the EES [3,5,19,20,24–27]. The reported prevalence of LBP in the included studies ranged broadly from a minimum of 36% to a maximum of 89% with an interquartile range of 34% [3,25]. The interquartile mean(SD) of the reported prevalence was 62(19)%. These minimum and mean prevalence rates of LBP in persons with LEA are considerably higher than the 15% to 25% prevalence values of LBP reported in the nonamputee general population [25].

One study reported significantly increased LBP after amputation as opposed to before amputation [5]. Furthermore, the characteristics and consequences of this pain in persons with LEA have been described as progressive, disabling, and contributing to limitations in occupation, recreation, and socialization [19,20]. With regard to function, LBP in LEAs has been associated with problems sitting, sleeping, and traveling [3]. Finally, LBP in this population has been associated with decreased health-related quality of life [5].

The second EES states that back pain is affected by level of amputation. Four moderate quality studies support the statement whereas three moderate quality studies do not support it. Ultimately, this yields a low level of confidence in the statement. Importantly, two of the studies supporting the statement were systematic literature reviews [20,28]. Both concluded that persons with transfemoral level amputation reported LBP with a higher prevalence than their transtibial counterparts, which was consistent with two additional clinical studies [27,29]. Perkins et al. suggest that the increased susceptibility to LBP at the higher amputation level may in part be the result of myofascial changes following transfemoral amputation along with gait pattern alterations [27]. Confirming these proposed causes for LBP following LEA through further research could lead to improvements in prevention and management.

Adverse effects of function related to LBP are the subject of the third EES. Hammarlund et al. used the Roland Morris Disability Questionnaire (RMDQ), a valid and reliable measure of functional capacity relative to perceived back pain in a sample of 46 nondysvascular lower extremity amputees [5]. They concluded that nearly all participants with LBP daily or several times per week reported severe or moderate disability on the RMDQ. Devan et al. assessed the relationship of back pain on function in terms of physical activity [26]. Overall, they concluded that there was no relationship between physical activity of LEAs with or without LBP and that there was an equal distribution of persons with LBP in low, medium, and high physical activity groups. They did however find that those reporting activity limitations due to LBP had lower physical activity scores than those with LBP who did not have physical activity limitations. It is important to note that this difference in function related to LBP is potentially confounded by the use of two different outcome approaches and further by the fact that Devan et al. studied those with traumatic transfemoral amputation, whereas the Hammarlund et al. sample was more heterogeneous by amputation level [5,26]. Nonetheless, further evidence is needed to understand which elements of function may potentially be impaired by LBP in persons with LEA.

Increasing frequency of LBP episodes and associated disability is the subject of EES four. A single, moderate-quality study supports EES four with a significant association ($p=.003$) between LEAs who reported LBP daily or several times per week and those reporting moderate or severe disability [5]. Devan et al. studied the relationship

between LBP and physical activity [26]. Their findings create further ambiguity in understanding disability as it relates to LBP in LEAs. That is, they found no association between physical activity in LEAs with or without LBP. One additional systematic review concluded that the majority of LEAs with LBP report minimal to no impact on social, recreational or work activities [20]. Conversely, approximately 25% described their LBP as severely interfering with these activities. These are important findings but do not directly relate to the issue of bout frequency of LBP. Ultimately, although the association identified by Hammarlund et al. was significant, the fact that only a single study supports the conclusion is presently insufficient to confidently support the statement at this time [5].

In EES five, association is made between leg length discrepancy and the presence of LBP in persons with LEA. One clinical study used the RMDQ to identify LEAs with LBP and those without [15]. Motion analysis was then used to determine leg length differences during static standing, dynamically during single and double limb support in gait and with either the prosthetic or sound foot leading. This single, high-quality clinical study did not find a relationship between leg length discrepancy and LBP. Conversely, a systematic literature review [28] indicates that leg length discrepancy among lower extremity prosthetic users is among many contributors to LBP. Further, the review states that those using prostheses that are of the same length as the sound limb have significantly fewer pain symptoms compared to those with length asymmetries between the intact and prosthetic limbs. Postural asymmetries reportedly result from these disparities. For instance, leg length differences of 12.5 mm have been associated with as much as 4° of lateral sacral tilt. It has been further reported that only 15% of LEAs use prostheses of equal length to the sound limb, whereas 34% of prosthesis users have prosthetic leg length differences greater than 20 mm, and that in 79% of cases, the prosthesis is the shorter limb. Given this disagreement between a single clinical study [15] and a systematic literature review [28], there is low confidence in the evidence supporting EES five. This statement indicates an association between leg length discrepancy and back pain in persons with lower extremity amputation. One additional clarifying point is that Morgenroth et al. studied LEAs with chronic LBP as opposed to acute onset cases. Thus, it is not currently possible to determine causation of LBP as a result of leg length discrepancy using these findings. Rather, their study is more useful in assisting to determine whether leg length discrepancy has a role in altering symptoms in chronic LBP cases among those with LEA [15].

Empirical evidence statement six is somewhat related to EES five. Though EES five directly addresses leg length discrepancy, EES six indicates that postural asymmetries and postural control issues are associated with LBP in patients with LEA. Gailey et al. report that persons with LEA tend to stand with increased sway and with increased weight bearing on the sound limb and that this may be

related to the lack of proprioception from the prosthesis [28]. Postural abnormalities observed in those with LEA are numerous including coronal and sagittal compensatory pelvic tilt, increased lumbar lordosis, involved-side hip flexion contracture, lateral trunk asymmetry and more [28]. Morgenroth et al. state that LBP is a common secondary disabling condition affecting TFA and that it is common clinical practice to assess for and correct postural asymmetry in the form of leg length discrepancy [15]. In their sample of subjects with longstanding transfemoral amputation and moderate, persistent LBP, leg length, discrepancies were not different relative to a similar population without LBP. Morgenroth et al. concluded that in longstanding transfemoral amputees with chronic symptoms, their LBP was unlikely to be related to their postural asymmetry [15]. Confidence in EES six is low given support from a systematic review but a lack of support from one clinical study, both of moderate quality.

Relative to EES seven, there is limited ($n=3$ studies) evidence reporting alterations in trunk, spinal, and pelvic motions among persons with LEA with LBP (EES seven). Although LBP is multifactorial, repeated exposures to altered trunk-pelvic motions is a purported risk factor for the onset or recurrence of LBP secondary to LEA [7,9,30–33]. The presence of LBP among persons with (transfemoral) LEA is associated with larger axial rotations of the lumbar spine [18], more rigid (in-phase) trunk-pelvic coordination strategies [16], and an apparent (albeit underpowered) trend toward a reversal in patterns of trunk-pelvic motion in the sagittal and transverse planes [17]. Such findings begin suggesting linkage of specific trunk/spinal and pelvic kinematic patterns with LBP secondary to LEA. However, the presence and magnitude of LBP has been inconsistently characterized using a variety of approaches, including binary yes/no, visual analog scale (0–10), question(s) within the Prosthesis Evaluation Questionnaire, and the Grade Questionnaire. Thus, there is a clear need for more consistent and comprehensive quantification of LBP in future work. For example, using the NIH task force for chronic LBP questionnaire that aims to classify LBP by its impact (ie, intensity, interference, and physical function), or using a minimal dataset to describe participants and reporting responder analyses in addition to mean outcomes could be useful [34]. Moreover, considerable prior work among non-amputation individuals have identified substantial influences of LBP on trunk and pelvic motions [35–37], begging the question of the relative contributions of LBP and LEA on the observed movement patterns. To that end, additional biomechanical metrics are needed to understand the underlying factors driving the movement patterns.

A single moderate quality study of 39 subjects supports the final EES [38]. This statement indicates that back pain is not affected by use of an osseointegrated prosthesis. Interfacing the prosthesis with a socket has been associated with adverse effects to skin, comfort, and function [23]. For example, skin erosions of varying

degrees, pain, and unreliable suspension can all potentially emerge related to socket use [39]. Anchoring the prosthesis to the residual limb via osseointegration purportedly mitigates some of the aforementioned complications. Another issue potentially associated with socket use is that gait pattern alterations due to pain or instability could also lead to LBP. Hagberg et al. surveyed LBP in a single item from the Questionnaire for Persons with Transfemoral Amputation [38] finding that at 2 years following osseointegration, approximately 40% of subjects reported reduced LBP, nearly 40% were unchanged, and nearly 20% reported an increase in their LBP symptoms. Compared with baseline, these differences were not statistically significant. Of note, the authors indicated the small sample size and reliance solely upon subjective outcomes limited the strength of evidence. Findings were also confounded by the fact that prosthetic components were changed throughout the 2-year follow-up period. More research is needed to identify and characterize the relationships between osseointegrated prosthetic use and LBP.

Clinical practice guidelines (CPGs) for primary care management of LBP in the general population usually recommend focused history and examination, limited use of diagnostic imaging, self-care, brief education, nonsteroidal anti-inflammatory drugs, manual therapy, and exercise [40]. The US Department of Defense (DoD) and the US Department of Veterans Affairs (VA) similarly have CPGs for persons with LEA and for those with LBP [41,42]. None of the articles uncovered in this systematic review assessed the appropriateness of these recommendations for patients with LEA suffering from LBP. Whether the LBP CPGs can be applied to LEAs or if modifications in the treatment approach are needed is unknown. Further, the recommendations in the VA/DoD include measuring the intensity of LBP but also to initiate a strengthening program for the upper and lower extremities as well as the core to prevent the development of LBP [42]. These recommendations in the second version of the VA/DoD CPG were forwarded from the original CPG and were largely based in expert opinion. These recommendations remain untested in this population. Future research is needed to clarify clinical decision-making processes for management of LBP in lower extremity amputees.

Limitations

This body of literature only included a single experimental study [38] and a single study with high internal and external validity [15]. The majority of the included studies were observational, of moderate overall quality, and unfunded. Viewed in aggregate, the subjects studied were somewhat heterogeneous with regard to age, LEA etiology, time since LEA, and included both military and civilian sectors; the methodological quality could be improved with standardized reporting in most cases [21,22]. An example

may include more thorough sample descriptions. Additionally, incorporating blinding (ie, raters, statisticians) would also improve internal validity. Further, important factors are missed due to reporting omissions in many cases such as gender, race or ethnic considerations. Finally, this review uncovered three potential etiologies of LBP in LEA, namely, leg length discrepancy (ESS five), postural asymmetries and control issues (ESS six), and altered spinal kinematics. However, a causal relationship between these potential etiologies and LBP in LEA has not been established and requires further research.

Conclusions

Because the majority of this body of evidence was observational instead of experimental, the EESs produced tended to describe factors affecting or that are affected by LBP in persons with LEA. More specifically, the EESs supported observationally have concluded that back pain in LEAs has relationships with the following phenomena: increased experiences, level of LEA, leg length differences, postural issues as well as spinal and pelvic kinematics. With only a single experimental study, the ability to determine optimal therapeutic intervention choices or their effects in managing LEAs who have LBP is greatly limited. Therefore, efficacy of interventions to manage LBP in persons with LEA remains a considerable knowledge gap and an area of future study.

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Abstract

Background: Alterations and asymmetries in trunk motions during activities of daily living are suggested to cause higher spinal loads in persons with unilateral lower limb amputation (LLA). Given the repetitive nature of most activities of daily living, knowledge of the amount of increase in spinal loads among persons with LLA is important for designing interventions aimed at prevention of secondary low back pain due to potential fatigue failure of spinal tissues. The objective of this study was to determine differences in trunk muscle forces and spinal loads between persons with and without LLA when performing a common activity of daily living, sit-to-stand and stand-to-sit tasks.

Methods: Three-dimensional kinematics of the pelvis and thorax, obtained from ten males with unilateral (transfemoral) LLA and 10 male uninjured controls when performing five repetitions of sit-to-stand and stand-to-sit activities, were used within a non-linear finite element model of the spine to estimate trunk muscle forces and resultant spinal loads.

Findings: The peak compression force, medio-lateral (only during stand-to-sit), and antero-posterior shear forces were respectively 348N, 269N, and 217N larger in person with vs. without LLA. Persons with LLA also experienced on average 171N and 53N larger mean compression force and medio-lateral shear force, respectively.

Interpretation: The spinal loads for both groups were generally smaller than the reported threshold of spinal tissues injury. However, tasks like sit-to-stand and stand-to-sit, with a peak compression force of ~ 2.6kN in persons with LLA, if performed following a highly repetitive activity like walking will impose >50% risk of fatigue failure for spinal tissues.

Keywords: Rising and sitting; Trunk muscle forces; Spinal loads; Low back pain; Limb loss; Biomechanics

Abstract

Background: Repeated exposures to larger lateral trunk-pelvic motion and features of knee joint loading likely influence the onset of low back pain and knee osteoarthritis among persons with lower-limb amputation. Decreased hip abductor strength can also influence frontal plane trunk-pelvic motion and knee moments; however, it is unclear how these are inter-related post-amputation.

Methods: Twenty-four participants with unilateral lower-limb amputation (14 transtibial; 10 transfemoral) and eight uninjured controls walked at 1.3 m/s while full-body biomechanical data were captured. Multiple linear regression and Cohen's f^2 predicted ($P < 0.05$) the influences of mediolateral trunk and pelvic ranges of motion, angular accelerations, and bilateral isometric hip abductor strength on peak (intact) knee adduction moment and loading rate.

Findings: There were no group differences in hip strength, peak knee adduction moment or pelvis acceleration ($p > 0.06$). The combination of hip strength, and mediolateral trunk and pelvic motion did not predict ($F_{(5,29)} = 2.53$, $p = 0.06$, adjusted $R^2 = 0.27$, $f^2 = 0.08$) peak knee adduction moment. However, the combination of hip strength and trunk and pelvis acceleration predicted knee adduction moment loading rate ($F_{(7,29)} = 3.59$, $p = 0.008$, adjusted $R^2 = 0.45$, $f^2 = 0.25$), with peak trunk acceleration ($\beta = 0.72$, $p = 0.008$) and intact hip strength ($\beta = 0.78$, $p = 0.008$) significantly contributing to the model.

Interpretation: These data suggest increased hip abductor strength counteracts increased lateral trunk acceleration, concomitantly influencing the rate at which the ground reaction force vector loads the intact knee joint. Persons with lower-limb amputation perhaps compensate for increased intact limb loading by increasing trunk motion, increasing demand on hip abductors to attenuate this preferential loading.

1. Abstract

Background: Persons with a unilateral, transtibial amputation (TTA) often exhibit abnormal trunk movement deviations during walking relative to uninjured persons. Prior work has shown that kinetics throughout the whole body have potential to contribute to trunk control. The aim of this study was to characterize how gait compensations of persons with a unilateral TTA contribute to altered, angular trunk dynamics at a whole-body level during walking.

Methods: Overground motion capture data were collected for 10 persons with a unilateral TTA and 10 uninjured persons walking at a self-selected speed. An induced acceleration analysis was used to decompose experimentally measured trunk angular accelerations into constituent accelerations caused by actions of all net joint moments in the body.

Findings: Several deviations in joint moments were found to correspond with altered trunk accelerations for the TTA group. The primary finding was that the prosthetic ankle plantarflexor moment and affected limb knee extensor moment imparted different accelerations on the trunk in both the frontal and sagittal planes. Knee-induced differences appeared to correspond with deficits in knee moment magnitude, while ankle-induced differences appeared associated with body postural factors.

Interpretation: Our findings highlighted that maladaptive mechanical compensations throughout the body may contribute to abnormal trunk angular movements in persons with a unilateral TTA. Interventional strategies such as movement training to alter foot placement or adjusting prosthetic device mechanical properties may be a useful supplement to traditional treatment methods to correct faulty trunk motion.

Keywords (max 6)

Human locomotion, below knee amputation, induced acceleration analysis, musculoskeletal modeling, movement re-training

ABSTRACT

Objective: To investigate trunk-pelvic kinematic outcomes with time from initial ambulation with a prosthesis, and amputation level, among persons with unilateral lower limb loss. It was hypothesized the magnitudes of trunk-pelvic range of motion (ROM) will increase and pelvic-trunk coordination will increase (become more out-of-phase) with increasing time of ambulation. Secondly, persons with more proximal limb loss will initially exhibit less trunk and pelvic ROM, and more in-phase trunk-pelvic coordination.

Design: Inception cohort with up to five repeated biomechanical evaluations during a one-year period (0, 2, 4, 6, and 12 months) after initial ambulation with a prosthesis.

Setting: Biomechanics laboratory within Military Treatment Facility

Participants: Thirty-two males with unilateral lower limb loss (twenty-two with transtibial limb loss and ten with transfemoral limb loss).

Interventions: Not applicable.

Main Outcome Measures: Triplanar trunk-pelvic ROM, and intersegmental coordination (continuous relative phase; CRP), were computed as participants walked overground at a self-selected (~1.30 m/s) and controlled (~1.20 m/s) speed.

Results: With increasing time after initial ambulation, trunk ROM generally decreased, most notably for persons with transfemoral limb loss, while pelvic ROM generally remained consistent. Mean CRP became more out-of-phase over time, and frontal CRP was more in-phase for persons with transfemoral vs. transtibial limb loss.

Conclusions: Temporal relationships in the features of trunk-pelvic motions within the first year of ambulation after limb loss have longer-term implications for the surveillance of LBP onset and recurrence, and may help identify important biomechanical factors in its causation. Future work should therefore continue longitudinal evaluations of trunk-pelvic motions, as well as injury rate and pain level.

Key Words: Extremity Trauma; Extremities; Wounds and Injuries; Biomechanics; Locomotion; Rehabilitation; Torso

Trunk muscle activation patterns during walking among persons with lower limb loss

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Introduction: Persons with vs. without unilateral lower limb amputation (LLA) walk with larger trunk and pelvis motions, presumably to assist with balance and forward progression; however, these potentially play a role in the elevated risk for low back pain (LBP) [1]. Uninjured individuals with vs. without LBP increase lumbar muscle activation when walking, often posited as a means to increase spinal stiffness and stability to avoid pain [2]. While aberrant activation magnitude of trunk musculature during walking is characteristic of persons with LBP, it is unknown whether similar alterations are observed in persons with LLA, and may provide insight into neuromuscular control strategies after LLA. Thus, the objective of this study is to determine trunk muscle activation patterns among persons with LLA during walking.

Methods: Fifteen participants with unilateral LLA [5 transfemoral (TFA; 38.4±6.3yrs, 174.3±4.6cm, 78.0±3.6kg); 10 transtibial (TTA; 33.3±8.4yrs, 178.9±8.1cm, 91.7±15.4kg)] and eleven uninjured controls (CTR; 30.6±8.9yrs, 176.8±8.7cm, 75.1±14.2kg) walked along a 15m walkway at 1.3m/s. Trunk electromyographic (EMG) data were obtained bilaterally from thoracic (TES) and lumbar (LES) erector spinae, high-pass filtered (Butterworth, cut-off frequency=20 Hz), and full-wave rectified. A root mean square envelope was calculated using a 50ms smoothing window. EMG activation magnitudes for each participant were normalized to the ensemble mean amplitude of each stride [3], maximum (“peak”) activations (as a percentage of mean activity) extracted during the gait cycle, and activation onset/offsets (relative to % gait cycle) determined via visual inspection. Single-factor ANOVAs ($p<0.05$) assessed the effects of group on peak activation and onset/offset of peak activation, with post hoc t-tests, and Cohen’s d assessing differences between groups ($p<0.0125$).

Results: While there were no differences in peak TES ($F_{(2,23)}=2.83$, $p=0.08$, $\eta^2=0.44$) and LES ($F_{(2,23)}=2.87$, $p=0.077$, $\eta^2=0.45$) between groups; TFA (vs. TTA, CTR) activated TES and LES earlier and maintained activation for longer durations (Figure 1).

Discussion: Earlier onset and delayed offset of trunk musculature among persons with TFA suggest these individuals adopt functional strategies to generate force to advance the affected limb. Among persons with TFA, reduced activations of global musculature, in the presence of larger motions, may thereby increase demands of local musculature to support and control motions of the spine. These patterns potentially elucidate altered neuromuscular control strategies associated with LBP development among persons with TFA, and therefore may guide trunk-specific motor control training paradigms.

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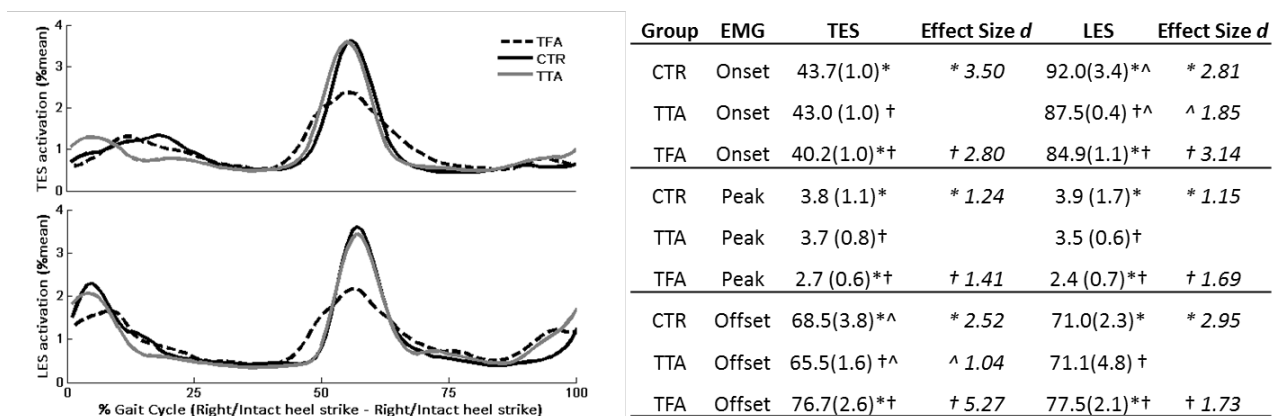


Figure 1. A) TES and LES activation as a percentage of the mean activation; B) Group and speed means (standard deviations) and effect sizes (d) for post hoc comparisons. *Indicate effect sizes of differences between CTR and TFA. †Indicate effect sizes of differences between TTA and TFA. ^Indicate effect sizes of differences between CTR and TTA.

Relationships among trunk-pelvic motion, hip strength, and knee joint moments during gait among persons with limb loss

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Introduction: Altered gait mechanics in persons with unilateral lower limb amputation (LLA) are associated with increased risk for low back pain (LBP) and knee osteoarthritis (OA) [1,2]. Specifically, repeated exposures to larger lateral trunk-pelvic motion and features of knee joint loading (e.g., rate of the knee adduction moment [KAM]) compared to uninjured controls likely contribute to such conditions over time. Decreased hip abductor strength can also influence frontal plane trunk-pelvic motion and knee moments; however, it is unclear how these are inter-related post-LLA [3]. Here, we report the relationships among features of trunk-pelvic motion, hip abductor strength, and frontal plane joint moment of the intact limb during gait among persons with LLA.

Methods: Fifteen participants with LLA [6 transfemoral (38.7±6.7yrs, 176.0±3.6cm, 80.8±9.8kg) and 9 transtibial (32.7±7.1yrs, 179.3±8.6cm, 92.3±16.7kg)] and six uninjured controls (25.5±4.2yrs, 175.8±9.6cm, 67.7±7.8kg) walked along a 15m walkway at 1.3(±10%) m/s. Full-body biomechanical data were captured using an 18-camera system and six force platforms embedded within the walkway. KAM was calculated by inverse dynamics using Visual3D; KAM_LR represents the slope between 20-80% of the period from minimum to first peak. Trunk and pelvic ranges of motion (ROM) were calculated with respect to the global coordinate system; peak trunk relative to pelvis angles (Trunk_Rel) were also computed. Trunk (TrunkAccel) and pelvis (PelvisAccel) angular accelerations were calculated as the slope of the angular velocity during the same period as KAM_LR. Intact/right limb eccentric hip abductor strength (HIP; (%BW*Ht)) was measured using a hand-held dynamometer. Multiple linear regression and Cohen's *d* were calculated to determine frontal plane predictors of KAM and KAM_LR ($p < 0.10$).

Results: After controlling for participant groups, neither HIP, trunk-pelvic ROM, Trunk_Rel, or Trunk/PelvisAccel significantly predicted peak KAM. However, the combination of HIP, PelvisAccel, TrunkAccel, trunk ROM, pelvis ROM, and Trunk_Rel significantly predicted KAM_LR ($F_{(4,13)}=4.077$, $p=0.014$, $R^2=0.69$) with HIP ($p=0.02$, $d=6.49$), PelvisAccel ($p=0.03$, $d=1.35$), and TrunkAccel ($p=0.091$, $d=1.30$) significantly contributing to the model.

Discussion: These data suggest that increased HIP allows for faster correction of contralateral pelvic drop during stance; however, faster and larger trunk rotations among persons with LLA compensate for decreased HIP, concomitantly influencing the rate at which the ground reaction force vector loads the knee joint (i.e., increasing KAM_LR). As such, considering interactions between proximal and distal segments is likely important to comprehensively characterizing mechanistic pathways for LBP and OA in persons with LLA.

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Table 1. Mean±SD for dependent and independent variables

	KAM_LR (%BW*Ht)	HIP (%BW*Ht)	PelvisAccel (deg/s)	TrunkAccel (deg/s)	Pelvis ROM (deg)	Trunk ROM (deg)	Trunk_Rel (deg)
Control (n=6)	37.4±12.8	14.5±2.0	652.0±282.3	217.2±69.7	7.5±2.7	3.5±1.1	6.0±2.9
Transtibial (n=9)	26.9±9.4	9.45±1.7*	527.5±170.7	318.2±118.2	5.2±1.7	7.0±1.6*	7.7±3.4
Transfemoral (n=6)	38.7±9.1	8.3±1.3*	576.1±541.0	554.2±285.0*	6.1±2.3	9.2±2.9*	5.2±4.0

*different from controls, < 0.05

Prevalence and Relationship of Low Back Pain and Psychosocial Factors after Lower Limb Amputation among Wounded Warrior Recovery Project Participants

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Background: Low back pain (LBP) is a common secondary health condition after limb amputation with important implications related to functional capabilities and quality of life. To date, however, the majority of data regarding the prevalence of LBP after amputation have come from studies of older military veterans or civilians with limb loss. As such, there is limited information on the prevalence of LBP after limb amputation in younger Service members from recent conflicts. Additionally, a growing body of evidence suggests that psychosocial factors, such as depression symptoms, significantly influence the experience of LBP in patients without amputation. However, there is currently a dearth of information available regarding the association of psychosocial factors and LBP after limb amputation. The purpose of this study was to assess the prevalence and potential association of LBP with psychosocial factors in Service members with deployment-related lower limb amputations.

Methods: Data on psychosocial factors, including quality of life, post-traumatic stress disorder (PTSD), and depression, comes from the Wounded Warrior Recovery Project (WWRP). The WWRP is an ongoing, web-based, longitudinal study that aims to gather patient-reported outcomes of deployment-related injured Service members. The Military Health System Data Repository was utilized to extract medical record data. Diagnostic codes were queried for at least one instance of coding related to LBP. The population of interest was individuals with deployment-related amputations. Of the current WWRP sample of 4,974 individuals who were injured on deployment between June 2004 and May 2013 and completed a baseline WWRP assessment between 2012 and 2017, 81 individuals had lower limb amputations. The majority of the sample of Service members with amputations were male (99%), enlisted (79%), Army (78%), and blast-related injuries (95%). General linear models were utilized to analyze associations between LBP and psychosocial factors, while controlling for injury severity and time since amputation.

Results: In this sample, 58% of individuals with amputations had been diagnosed with LBP by a medical provider; 31% screened positive for PTSD using the PTSD Checklist and 32% screened positive for depression using the Center for Epidemiological Studies Depression Scale. Among individuals with amputations, those with LBP reported lower quality of life (0.415, standard error [SE] = .014) compared to those with amputations without LBP (0.470, SE = .016) ($B = .055$, $p = .01$; $\eta^2 = .085$). Similarly, individuals with amputations and LBP reported higher PTSD scores (40.95, SE = 2.11) compared to those without LBP (33.91, SE = 2.43) ($B = -7.039$, p

=.033; $\eta^2 = .059$). There was no significant difference between depression scores in individuals with amputations with (15.17, SE = 1.50) or without LBP (11.51, SE = 1.72) ($p = .113$).

Conclusions: Presence of LBP after limb amputation appears to be associated with greater PTSD symptoms and lower quality of life. Given the cross-sectional nature of the current data, determination of a cause-and-effect relationship was not possible. Further research is needed to assess the efficacy of addressing psychosocial factors as part of a multi-disciplinary approach for improving pain and function in Service members with amputations and LBP.

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Trunk Muscle Activations, Motions, and Spinal Loads among Persons with Lower Limb Amputation: Influences of Chronic Low Back Pain

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INTRODUCTION: Low back pain (LBP) is a prevalent and costly musculoskeletal disability, particularly within the military, where LBP is among the leading causes of medical visits and lost duty days [1]. Although most LBP remains idiopathic, physical (biomechanical) factors likely contribute more substantially in certain populations (e.g., persons with lower limb amputation; LLA) [2]. During activities of daily living, altered trunk-pelvic motion in persons with (vs. without) LLA impose greater mechanical loads on spinal tissues, and thus have been suggested as a risk factor for the development of LBP [3]. Moreover, in the presence of LBP, further alterations in trunk-pelvic motion have been characterized [4-6]; yet, features of trunk muscle activations underlying these altered motions with LBP and effects on spinal loads remain unclear. The purpose of this study was therefore to evaluate trunk-pelvic motion, muscle activities, and spinal loads among persons with LLA both with and without LBP (and a control group without LBP for reference). Among persons with (vs. without) LLA, we hypothesized trunk muscle activations and corresponding trunk-pelvic motions with vs. without LBP would be associated with larger spinal loads, supporting a pathway wherein repeated exposure to abnormal lumbopelvic mechanics can adversely affect spine health.

METHODS: Eighteen persons with LLA – 8 with LBP (“LLA-P”) and 10 without LBP (“LLA-NP”) – and 10 uninjured controls without LBP (“CTR-NP”) participated in this cross-sectional, IRB-approved study. The LLA-P group reported chronic LBP (n=7 every day or nearly every day in the most recent 6 months; n=1 at least half of days in the most recent 6 months [7]). Mean (standard deviation) pain in the past seven days = 3.8 (1.3). Participants walked overground across a 15m walkway at 1.3m/s while an 18-camera motion capture system (Qualisys, Göteborg, Sweden) tracked (120Hz) trunk and pelvis kinematics. Simultaneously, electromyographic (EMG) activities were sampled (1200Hz) bilaterally at two levels of the erector spinae (T9 [TES] and L2 [LES]). Kinematic data were low-pass filtered (Butterworth, 6 Hz), EMG data were high-pass filtered (Butterworth, 20 Hz), full-wave rectified, and smoothed with a 50ms RMS window. Tri-planar trunk and pelvis angles and angular velocities were calculated in Visual3D (C-motion, Germantown, MD). Global trunk and pelvic ranges of motion (ROM) were determined and trunk-pelvic continuous relative phases (CRP) were calculated in MATLAB (Mathworks, Natick, MA). EMG data were normalized to the signal mean and down-sampled to match kinematic capture rate, and processed bilaterally for individuals with LLA (i.e., on both the intact and affected side) and for the right side of CTR-NP as no differences were observed between left and right sides. Cross-correlations related EMG and trunk and pelvic rotation time series during both intact and affected strides. Finally, trunk-pelvic kinematic data were input to a non-linear finite element model of the spine [8], wherein a heuristic optimization procedure estimated trunk muscle forces and spinal loads by minimizing the sum of squared muscle stress (i.e., the cost function) across 56 muscles. Individual muscle forces were summed across local (i.e., connecting individual lumbar vertebrae to the pelvis) and global muscles (i.e., connecting the thorax/rib cage to the pelvis). Spinal loads were compiled from the intervertebral level at which maximum spinal loads occurred (i.e., L5/S1). Peak spinal loads and muscle forces were determined and normalized to body mass. Separate one-way repeated-measure ANOVAs assessed the effect of group (LLA-P, LLA-NP, CTR-NP) on ROM, mean CRP, magnitudes

of cross-correlation coefficients (R), and spinal loads ($P < 0.05$). Bonferroni-corrected t-tests assessed pairwise differences ($P < 0.0167$).

RESULTS: Larger frontal plane trunk ROM were observed in both LLA-P and LLA-NP vs. CTR-NP ($P < 0.008$). Transverse plane trunk ROM were larger in LLA-P vs. CTR-NP ($P < 0.001$) but similar to LLA-NP ($P = 0.032$). A main effect was observed in frontal plane mean CRP, with LLA-NP exhibiting smaller (i.e., more in-phase) CRP than CTR-NP ($P = 0.013$). Main effects were observed in the R-values of both intact and affected side LES and sagittal trunk angles. Affected-side LES activations more strongly correlated with sagittal trunk angles in LLA-P and LLA-NP vs. CTR ($P < 0.003$). Intact-side LES more strongly correlated with sagittal trunk angles in LLA-NP vs. CTR-NP ($P = 0.005$), and with transverse trunk angles in LLA-NP vs. LLA-P ($P = 0.010$). R-values of affected-side LES and sagittal pelvis angles were greater in LLA-NP vs. CTR-NP ($P = 0.011$) but similar between LLA-P vs. CTR-NP ($P = 0.023$). R-values of trunk and pelvis angles and TES did not differ between groups. No main effects were observed in peak spinal loads ($P > 0.078$) or global muscle forces ($P = 0.076$). However, peak local muscle forces differed between groups ($P = 0.017$); local muscle forces were larger in CTR-NP vs. both LLA-P ($P = 0.012$) and LLA-NP ($P = 0.016$).

CONCLUSION: Though sagittal plane kinematics were similar between groups, R-values of LES activation patterns and sagittal plane trunk and pelvis angles were larger in both LLA-P and LLA-NP vs. CTR, suggesting an active LES control strategy that may be especially important in the LLA-P group as they tended to walk with more anterior trunk lean than CTR. In the transverse plane, smaller trunk ROM and stronger correlations between intact LES and trunk angles were observed in LLA-NP vs. LLA-P, suggesting LLA-NP are better able to control transverse plane trunk rotations. Despite larger trunk ROM in both the frontal and transverse planes, a lack of differences in spinal loads among LLA-P and LLA-NP vs. CTR are contrary to both our hypotheses (LLA-P \neq LLA-NP) and prior work [LLA-NP $>$ CTR; 9]. Nevertheless, larger transverse motions with vs. without LBP are consistent with prior work in persons with (transfemoral) LLA [4]. Interestingly, all persons in the LLA-P group had transtibial LLA while those in the LLA-NP group comprised a combination of both transtibial ($n = 7$) and transfemoral ($n = 3$) LLA. In the absence of LBP, alterations in the characteristics of trunk-pelvic motion are typically larger in persons with transfemoral vs. transtibial LLA [2]. It is therefore possible that presence of chronic LBP has concurrently increased trunk-pelvic motions in a group that is otherwise more similar to uninjured CTR. Also, while we identified group differences in local muscle forces, further consideration may be warranted for the model/optimization assumptions regarding muscle recruitment strategies with vs. without LBP [10]. Of note, despite categorization of chronic LBP, participants in the LLA-P group at the time of testing reported mean (standard deviation) numerical pain scores of 2.2 (1.3).

In summary, although prior work has identified larger spinal loads in persons with vs. without LLA, the current (cross-sectional) results do not necessarily support the notion that larger spinal loads during walking influence the persistence of LBP. It is however possible that individuals in the LLA-P group experienced larger spinal loads at some point prior to developing LBP and, thus, future work is needed to longitudinally characterize the temporal relationships in these outcomes with time since LLA to better elucidate the causal relationships.

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Learning Objectives:

1. Describe risk factors for low back pain secondary to LLA
2. Describe differences in trunk-specific outcomes between those with LLA with and without low back pain
3. Describe clinical considerations to mitigate the impact of LBP secondary to LLA

TRUNK MUSCLE FORCES AND SPINAL LOADS WHILE WALKING IN PERSONS WITH LOWER LIMB AMPUTATION BOTH WITH AND WITHOUT CHRONIC LOW BACK PAIN

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INTRODUCTION

Low back pain (LBP) is a common musculoskeletal impairment among persons with lower limb amputation (LLA), capable of substantially reducing longer-term quality of life [1]. During activities of daily living, such as walking, altered trunk-pelvic motion with (vs. without) LLA impose greater mechanical loads on spinal tissues, and thus have been suggested as a risk factor for the development of LBP [2]. Moreover, in the presence of LBP, further alterations in trunk-pelvic motion have been identified [3-5], yet the effects of these altered motions with LBP on spinal loads remain unclear. The purpose of this study was to evaluate the influences of LBP on trunk-pelvic motion and spinal loads among persons with LLA. We hypothesized that there are differences in trunk-pelvic motions that are associated with larger spinal loads between persons with LLA with and without LBP, supporting a pathway wherein repeated exposure to abnormal spine mechanics can adversely affect spine health.

METHODS

Eighteen persons with LLA – 8 with LBP (“LLA-P”) and 10 without LBP (“LLA-NP”) – and 10 uninjured controls (“CTR”; without LBP) participated (Table 1). The LLA-P group reported chronic LBP ($n=7$; every day or nearly every day in the most recent 6 months, $n=1$; at least half of days in the most recent 6 months) [6]. Mean (standard deviation) pain in the past seven days = 3.8 (1.3). Participants walked overground across a 15m walkway at 1.3 m/s, with speed enforced by auditory feedback. An 18-camera motion capture system (Qualisys, Göteborg, Sweden) tracked (120Hz) trunk and pelvis kinematics via 10

reflective markers. Marker trajectories were low-pass filtered (Butterworth, 6Hz).

Table 1. Mean (SD) participant demographics.

	LLA-P	LLA-NP	CTR
Age (yr)	35.1 (8.7)	36.4 (6.8)	29.7 (8.9)
Stature (cm)	177.5 (8.0)	179.3 (5.9)	176.0 (6.3)
Mass (kg)	86.8 (11.5)	91.6 (14.6)	73.2 (13.4)
Time (yr)	5.2 (2.6)	10.5 (3.1)	N/A

Tri-planar (global) trunk and pelvis angles, and pelvis center of mass position were calculated in Visual3D (C-motion, Germantown, MD, USA), time-normalized to stride, and subsequently input to a non-linear finite element model of the spine [8]. A heuristic optimization procedure, controlled via MATLAB (Mathworks, Natick, MA, USA), estimated trunk muscle forces and spinal loads by minimizing the sum of squared muscle stress (i.e., the cost function) across 56 muscles. Individual muscle forces were summed across local (i.e., connecting individual lumbar vertebrae to the pelvis) and global muscles (i.e., connecting the thorax/rib cage to the pelvis). Spinal loads were compiled from the intervertebral level at which maximum spinal loads occurred (i.e., L5/S1). Peak spinal loads and muscle forces were determined and normalized to body mass. Trunk ranges of motion (ROM) were also determined. Separate one-way repeated-measure ANOVAs assessed the effect of group (LLA-P, LLA-NP, CTR) on all outcomes ($P<0.05$). Bonferroni-corrected t-tests ($P<0.0167$) assessed pairwise differences when main effects were observed.

RESULTS AND DISCUSSION

Main effects were observed in frontal ($P=0.003$) and transverse ($P<0.001$) plane trunk ROM (Table

1). In the frontal plane, trunk ROM were larger in LLA-P ($P=0.001$) and LLA-NP ($P=0.011$) vs. CTR. In the transverse plane, trunk ROM were larger in LLA-P versus both LLA-NP ($P=0.015$) and CTR ($P<0.001$), but were similar between LLA-NP versus CTR ($P=0.022$). No main effects were observed in peak spinal loads ($P>0.078$) or global muscle forces ($P=0.076$; Table 1). However, peak local muscle forces differed between groups ($P=0.017$); local muscle forces were larger in CTR vs. both LLA-P ($P=0.012$) and LLA-NP ($P=0.016$).

Despite larger trunk ROM in both the frontal and transverse planes, a lack of differences in spinal loads among LLA-P and LLA-NP vs. CTR are contrary to both our hypotheses (LLA-P \neq LLA-NP) and prior work [LLA-NP > CTR; 8]. Nevertheless, larger transverse motions with vs. without LBP are consistent with prior work in persons with (transfemoral) LLA [4]. Interestingly, all persons in the LLA-P group had transtibial LLA while those in the LLA-NP group comprised a combination of both transtibial (n=7) and transfemoral (n=3) LLA. In the absence of LBP, alterations in the characteristics of trunk-pelvic motion are typically larger in persons with transfemoral vs. transtibial LLA [2]. It is therefore possible that presence of chronic LBP has concurrently increased trunk-pelvic motions in a group that is otherwise more similar to uninjured CTR. Also, while we identified group differences in local muscle forces, further consideration may be warranted for the model/optimization assumptions regarding muscle recruitment strategies with vs. without LBP [9]. Of note, despite categorization of chronic LBP, participants in the LLA-P group at the time of testing reported mean (standard deviation) numerical pain scores of 2.2 (1.3).

In summary, although prior work has identified larger spinal loads in persons with vs. without LLA, the current results do not necessarily support the notion that larger spinal loads during walking influence the persistence of LBP. It is however possible that individuals in the LLA-P group experienced larger spinal loads at some point prior to developing LBP and, thus, future work is needed to longitudinally characterize the temporal relationships in these outcomes with time since LLA to better elucidate the causal relationships.

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Table 1: Mean (SD) peak anteroposterior (AP) and mediolateral (ML) shear forces, compression forces, and global and local muscle forces, and tri-planar trunk ranges of motion (ROM) for LLA-P, LLA-NP, and CTR. * indicate statistically different than CTR, # indicate statistically different from LLA-NP.

	Spinal Loads (N/kg)			Muscle Forces (N/kg)		Trunk ROM (°)		
	AP Shear	ML Shear	Compression	Global	Local	Sagittal	Frontal	Transverse
LLA-P	5.4 (1.6)	8.7 (2.1)	22.3 (3.5)	11.0 (2.0)	8.6 (0.9)*	2.9 (1.1)	7.4 (1.5)*	9.5 (1.9)*#
LLA-NP	5.3 (3.4)	10.4 (5.8)	23.1 (3.6)	10.9 (2.6)	8.8 (1.0)*	2.8 (0.8)	6.3 (3.0)*	7.2 (2.3)
CTR	4.4 (0.8)	6.3 (2.3)	22.5 (2.8)	13.5 (3.1)	10.6 (2.3)	2.2 (0.7)	3.6 (1.6)	5.2 (1.2)

PATTERNS OF ERECTOR SPINAE ACTIVATION AND TRUNK-PELVIS KINEMATICS IN PERSONS WITH LOWER LIMB AMPUTATION: INFLUENCES OF LOW BACK PAIN

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INTRODUCTION

A higher prevalence of low back pain (LBP) is reported in persons with lower limb amputation (LLA) vs. uninjured individuals; moreover, persons with LLA often report that LBP negatively impacts quality of life [1]. In uninjured persons, movement impairments at the trunk and pelvis during gait, as well as altered trunk muscle activities, have been associated with increased risk for LBP [2]. However, there has yet to be a comprehensive study which examines both trunk kinematic and electromyographic (EMG) data while walking in persons with LLA, and more specifically, compares the influences of spine health (i.e., presence/severity of LBP). Therefore, the purpose of this study is to examine the relationship between patterns of trunk/pelvis kinematics and trunk muscle activations while walking, specifically comparing individuals with LLA with and without LBP.

METHODS

Seventeen persons with traumatic unilateral LLA – 9 with LBP (BP; 8 transtibial (TT), 1 transfemoral (TF), mean±standard deviation age (yrs): 34.7±7.8, stature (cm): 176.8±7.8, mass (kg): 85.4±11.6, and time since injury (yrs): 5.1±2.5) and 8 without LBP (NP; 5 TT, 3 TF, age: 36.8±7.6, stature: 179.9±5.4, mass: 91.8±15.8, and time since injury: 10.0±3.2) – and 6 uninjured controls (CTR, age: 28.2±9.0, stature: 174.5±4.5, and mass: 68.9±7.8) participated in this study. LBP was characterized via the NIH recommended minimal dataset [3]. Individuals walked across a 15m overground walkway at 1.3 m/s, with speed enforced by auditory feedback. An 18-camera motion capture system (Qualisys, Göteborg, Sweden) tracked (120Hz) trunk and pelvis kinematics via 10 reflective markers. Kinematic data were low-pass filtered (Butterworth, 6 Hz). Tri-planar trunk and pelvis angles and

angular velocities were calculated in Visual3D (C-motion, Germantown, MD). Global trunk and pelvic ranges of motion (ROM) were determined and trunk-pelvic continuous relative phases (CRP) were calculated in MATLAB (Mathworks, Natick, MA). EMG data were collected (1200 Hz, Motion Lab Systems, Baton Rouge, LA) bilaterally at two levels of the erector spinae (T9 (TES) and L2 (LES)), high-pass filtered (Butterworth, 20 Hz), and full-wave rectified. A 50ms RMS smoothing window was then applied. EMG data were normalized to the signal mean and down-sampled to match kinematic capture rate. EMG data were processed bilaterally for individuals with LLA (i.e., on both the intact and affected side) and for the right side of CTR as no differences were observed between left and right sides. Both kinematic and EMG data were time-normalized to stride. Cross-correlations related EMG and trunk and pelvic rotation time series during both intact and affected strides. Separate one-way repeated measures ANOVAs assessed the effect of group (CTR, BP, NP) on ROM, mean CRP, and magnitudes of cross-correlation coefficients (R), with significance concluded at $P<0.05$. Bonferroni-corrected t-tests assessed pairwise differences ($P<0.0167$).

RESULTS and DISCUSSION

Larger frontal plane trunk ROM were observed in both BP and NP vs. CTR ($P<0.008$). Transverse plane trunk ROM were larger in BP than CTR ($P<0.001$) and tended to be larger than NP ($P=0.032$). A main effect was observed in frontal plane mean CRP, with NP exhibiting smaller (i.e., more in-phase) CRP than CTR ($P=0.013$). Main effects were observed in the R-values of both intact and affected side LES and sagittal trunk angles (Figure 1). Affected-side LES activations more strongly correlated with sagittal trunk angles in BP

and NP vs. CTR ($P<0.003$). Intact-side LES more strongly correlated with sagittal trunk angles in NP vs. CTR ($P=0.005$), and with transverse trunk angles in NP vs. BP ($P=0.010$). R-values of affected-side LES and sagittal pelvis angles were greater in NP vs. CTR ($P=0.011$) and tended to be larger in BP vs. CTR ($P=0.023$). R-values of trunk and pelvis angles and TES did not differ between groups.

The observed changes in frontal plane CRP are consistent with prior work and posited to be a trunk stiffening strategy to prevent injury and/or mitigate pain [4]. Though sagittal plane kinematics were similar between groups, R-values of LES activation patterns and sagittal plane trunk and pelvis angles were larger in both BP and NP vs. CTR. This suggests persons with LLA utilize LES to control sagittal plane trunk and pelvis motion during walking. Such an active control strategy may be especially important in the BP group as they tended to walk with more anterior trunk lean than CTR ($P=0.023$); this anterior shift in center of mass has been associated with an increased risk of falls [5]. While increased LES contributions may compensate for decreases in passive stability, the increased demand on the muscles may contribute to LBP development. NP may minimize this risk by using LES bilaterally to regulate sagittal trunk motions, and distributing the associated demand across both sides. BP, meanwhile, seems to rely more heavily

on affected-side LES, which may increase the risk of injury on that side. In the transverse plane, smaller trunk ROM and stronger correlations between intact LES and trunk angles were observed in NP vs. BP. This suggests NP individuals are better able to control transverse plane trunk rotations, likely using LES to limit axial rotations. As increases in trunk and pelvic rotations are associated with increased spinal loads [6], the observed reductions in transverse plane trunk ROM may help mitigate the risk of LBP in the NP group. Thus, interventions training low-back musculature and enhancing trunk postural control strategies in persons with LLA are likely warranted.

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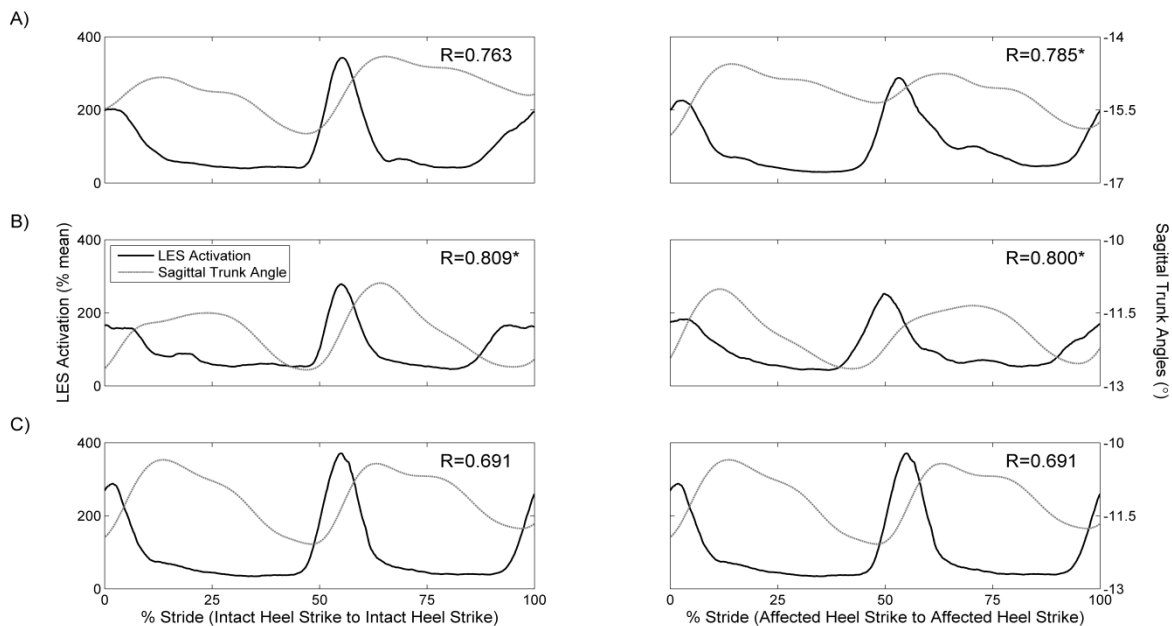


Figure 1: LES activation patterns (black), sagittal trunk angles (gray) and corresponding cross-correlation coefficients (R) for LBP (A), NP (B), and CTR (C) on the intact (left) and affected (right) sides. Asterisks (*) indicate statistically different from CTR.

INFLUENCES OF LOW BACK PAIN ON THE ENERGY CONTRIBUTIONS OF THE HIP AND SPINE DURING GAIT AMONG PERSONS WITH LOWER LIMB LOSS

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INTRODUCTION

Persons with vs. without unilateral lower limb loss (LL) walk with altered trunk-pelvic mechanics that, with repeated exposure, presumably represent a mechanistic pathway for low back pain (LBP) [1,2]. Specifically, persons with LL walk with increased lateral trunk motion over the stance limb, posited as an adaptive strategy to compensate for absent or weak musculature in the lower extremity. Moreover, hip ab/adduction moments may compensate for increased lateral trunk motion over the stance limb, suggesting an altered motor pattern that redistributes energy/power during gait [3]. Among uninjured individuals, impaired hip abductor strength is associated with LBP, suggesting impaired load/energy transfer between the lower extremity and lumbar spine. While persons with vs. without LL demonstrate increased positive phases of joint powers at L5/S1 in the frontal plane, the contributions of the unaffected hip powers to lumbar spine mechanics are unknown. Further, to date, no study has compared frontal plane low back and hip movement strategies among persons with limb loss and varying degrees of disability associated with LBP. Thus, the objective of the current study was to determine the contributions of hip and low back joint powers (L5/S1) to LBP-related disability among persons with limb loss, hypothesizing that persons with greater LBP disability will demonstrate larger power generation through the unaffected hip and low back.

METHODS

Nineteen persons with traumatic unilateral lower LL ($n = 7$ transfemoral, $n = 12$ transtibial; mean \pm standard deviation age: 31.9 ± 12.5 yrs, stature: 1.8 ± 0.1 m, body mass: 89.1 ± 13.7 kg, and time since

injury: 8.6 ± 7.0 yrs) participated in this cross-sectional study after providing written informed consent to study procedures approved by the local IRB. Acute LBP was characterized using a Visual Analog Scale. The presence of chronic LBP was determined via self-report using the Oswestry Disability Index (ODI; "I have 'chronic pain' or pain that has bothered me for 3 months or more"), and was further quantified using ODI percent disability. Participants walked at 1.3 m/s ($\pm 10\%$) along a 15 m walkway with full-body biomechanical data captured using an 18-camera system (Qualisys, Göteborg, Sweden) and six force platforms (AMTI, Watertown, MA) embedded within the walkway. Marker positions and ground reaction forces (GRF) were smoothed using a fourth-order dual-pass Butterworth filter with cutoff frequencies of 6 Hz and 45 Hz, respectively. L5/S1 and hip joint powers were calculated as the product of joint moment and angular velocity using 6DOF inverse dynamics in Visual3D (C-motion, Germantown, MD) and normalized to body mass. Positive/negative work at the L5/S1 and hip joints, calculated as the total areas under the joint power curves, respectively indicate mechanical energy generation/absorption.

Multiple regression was used to determine the influences of frontal plane L5/S1 and hip joint powers on ODI percent disability among persons with LL. Independent t-tests were used to determine differences in work between persons with and without chronic LBP at L5/S1 and hip joints throughout the gait cycle ($P < 0.05$).

RESULTS AND DISCUSSION

After controlling for time since amputation and stride width, the total positive and negative powers for both L5/S1 and hip joints significantly predicted LBP-related disability ($F_{(6,18)} = 5.11$, $P = 0.008$), with all

but positive hip power significantly contributing to the prediction. Total positive and negative work through the unaffected hip and low back explain 58% of the variance in the model. While there were no significant ($P > 0.11$) differences in the total work at L5/S1 or hip joints between persons with LL whom identified themselves as having chronic LBP (acute pain = 1.4 ± 1.7 , ODI percent disability = 26.5 ± 24.7) and those who did not (acute pain = 0.1 ± 0.3 , ODI percent disability = 11.3 ± 12.4), there were distinct differences in joint power waveform characteristics throughout the gait cycle. At L5/S1, persons with LL and chronic LBP demonstrate greater energy absorption during loading response, whereas those without LBP demonstrate greater energy absorption just prior to toe-off (Figure 1). Persons with LL and chronic LBP walk with greater trunk motion during early stance yet demonstrate larger energy absorption at the hip. Although not reported here, this is likely the result of greater hip joint angular velocity, counteracting larger trunk lateral flexion and contralateral pelvic drop during early stance, as a means to maintain mediolateral balance. Such a hip dominant strategy could also have implications for the increased joint loading and prevalence of hip osteoarthritis among persons with LL [5]. The two distinct negative power phases at L5/S1 among persons with LL without chronic LBP are similar to previous reports [4]; in contrast, persons with LL and chronic LBP demonstrate a smaller L5/S1 negative peak power at toe-off that is coupled with a larger positive peak power at the hip. The larger power generation at the hip suggests an active hip strategy to control the mediolateral movement of the center of mass as it moves from peak lateral flexion over the stance limb towards the subsequent heel-strike of the affected limb. Thus, persons with limb loss and LBP may adopt a compensatory strategy to avoid pain and/or to account for impaired neuromuscular control of the trunk. Future research should focus on developing interventions geared towards improving neuromuscular control strategies of the trunk-pelvic-hip complex, thereby reducing possible mechanisms of LBP-related disability.

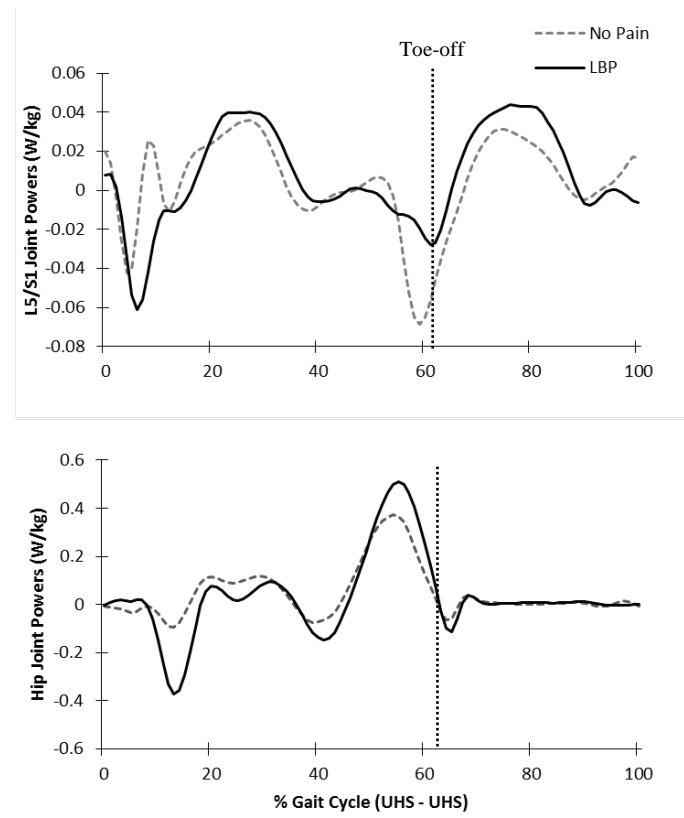


Figure 1: L5/S1 (top) and unaffected hip joint (bottom) powers during the gait cycle (unaffected heel strike (UHS) to unaffected heel strike (UHS)) among persons with limb loss with vs. without self-identified chronic LBP.

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Toward Optimizing Long-term Health after Limb Loss: Comprehensive Evaluations of Secondary Health Conditions

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Background: Extremity trauma, including limb loss, is commonly associated with an increased prevalence of and risk for developing secondary health conditions (e.g., low back pain and osteoarthritis) [1,2]. While the underlying etiologies of these disorders after limb loss remain unclear, there is growing support for the biopsychosocial model toward identifying the multifactorial contributors to their onset and progression [3,4]. Importantly, such an approach requires comprehensive and concurrent evaluation of several domains to effectively characterize risk. The objective of this meta-analysis is therefore to explore and describe relationships among biological and psychosocial outcomes associated with low back pain and (contralateral) knee joint health among individuals with unilateral lower limb loss.

Design/Methods: Eighteen males with traumatic, unilateral lower limb loss (10 transtibial, 8 transfemoral) completed a comprehensive evaluation consisting of biomechanical, biochemical, and psycho-social assessments. Estimates of mechanical loads at the contralateral knee (peak adduction moment) and low back (peak L5/S1 lateral moment) were calculated from full-body biomechanical data collected as participants walked along a 15m walkway. Biochemical data were obtained via blood draws to quantify serum levels of hyaluronan, stromelysin-1, and cartilage oligomeric matrix protein by ELISA as markers of cartilage degradation. Psychosocial outcomes were also obtained using validated, patient-administered instruments for fear of movement, pain catastrophizing, anxiety, and depression. These outcomes were collectively associated with low back pain disability [5] and knee-related quality of life [6] scores via regression analyses, controlling for level of amputation and time since injury (mean [SD] = 128 [86] months). All participants provided informed consent to procedures approved by the Walter Reed National Military Medical Center Institutional Review Board.

Results: With regard to low back pain disability (mean [SD]= 11.5 [18.0]), the collective suite of biopsychosocial outcomes accounted for 61.1% of variance within the model (vs. 13.5% when only controlling for level of amputation or time since injury exclusively); greater fear of movement was a significant predictor ($p=0.016$) of low back pain disability. With regard to knee-related quality of life (mean [SD]= 82.9 [17.5]), the comprehensive model accounted for 62.3% of variance (vs. 31.5%); greater anxiety/depression was a significant predictor ($p=0.042$) of lower knee-related quality of life.

Conclusions: Although risk for pain and joint degeneration within the low back and contralateral knee after unilateral limb loss is largely theorized as a biological/mechanical process [e.g., 7,8], psychosocial factors most influenced these outcomes in the current (cross-sectional) dataset. Additional participants and follow-up evaluations will be important to characterize biopsychosocial correlates (and their relative timing with respect to onset) of these secondary health conditions. Given the relatively young age of Service Members with extremity trauma and thus potential for cumulative, lifelong disability, such an approach is critical for optimizing long-term outcomes and quality of life.

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TRUNK MUSCLE FORCES AND SPINAL LOADS DURING SIT-TO-STAND AND STAND-TO-SIT ACTIVITIES: DIFFERENCES BETWEEN PERSONS WITH AND WITHOUT UNILATERAL TRANSFEMORAL AMPUTATION

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Low back pain (LBP) is a significant secondary health problem in persons with unilateral lower limb amputation. In particular, persons with versus without transfemoral amputation (TFA) often adopt different trunk postures/motions when performing activities of daily living to overcome the physical limitation(s) imposed by amputation. Such differences in trunk postures/motions, if associated with even moderate increases in spinal loads across all activities of daily living, can lead to LBP via cumulative damages in spinal tissues. The objective of this study was to compare spinal loads between persons with (n=10) and without (n=10) TFA when performing sit-to-stand and stand-to-sit activities. A non-linear finite element model of the lumbar spine and trunk muscles, adjusted for participant height and weight, was used to calculate trunk muscle forces and the resultant spinal loads. Model inputs were kinematics of thorax and pelvis measured when participants performed sit-to-stand and stand-to-sit activities. Forces within superficial muscles (attached between pelvis and thorax spine) were 145 N larger* in person with versus without TFA, while forces within deeper muscles (attached between pelvis and lumbar spine) were 57 N larger during stand-to-sit versus sit-to-stand. The resultant mean and peak values of compression force at L5-S1 were respectively 171 N (~12%) and 348 N (~16%) larger in persons with TFA. The maximum value of anterior-posterior shear force at L5-S1 was also 217 N (~24%) larger in persons with TFA. Finally, in persons with TFA the mean and maximum values of lateral shear force at L5-S1 were respectively 68 N (~92%) and 215 N (~81%) larger during stand-to-sit versus sit-to-stand. The peak value of shear force experienced at L5-S1 (~1.1 kN) among persons with TFA during sit-to-stand was within the reported range of threshold of injury (i.e., 1-2 kN) for lumbar spine motion segments. Considering we have recently reported persons with versus without TFA experience larger spinal loads during walking, characterization of these loads during (other) activities of daily living further highlights their potential role in LBP after TFA, and may assist with the development of trunk-specific movement retraining or other preventative therapies.

* $p < 0.05$ in all reported results