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Short communication

Are lower back demands reduced by improving gait symmetry in unilateral transtibial amputees?

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ABSTRACT

Background: Gait asymmetry and a high incidence of lower back pain are typical for people with unilateral lower limb amputation. A common therapeutic objective is to improve gait symmetry; however, it is unknown whether better gait symmetry reduces lower back pain risk. To begin investigating this important clinical question, we examined a preexisting dataset to explore whether L5/S1 vertebral joint forces in people with unilateral lower limb amputation can be improved with better symmetry.

Methods: L5/S1 compression and resultant shear forces were estimated in each participant with unilateral lower limb amputation ($n = 5$) with an OpenSim musculoskeletal model during different levels of guided gait asymmetry. The amount of gait asymmetry was defined by bilateral stance times and guided via real-time feedback. A theoretical lowest L5/S1 force was determined from the minimum of a best-fit quadratic curves of L5/S1 forces at levels of guided asymmetry ranging from −10 to +15%. The forces found at the theoretical lowest force and during the 0% asymmetry level were compared to forces at preferred levels of asymmetry and to those from an able-bodied group ($n = 5$).

Findings: Results indicated that the forces for the people with unilateral lower limb amputation group at the preferred level of asymmetry were not different then at their 0% asymmetry condition, theoretical lowest L5/S1 forces, or the able-bodied group (all p-values $>$ .23).

Interpretation: These preliminary results challenge the premise that restoring symmetric gait in people with unilateral lower limb amputation will reduce risk of lower back pain.

1. Introduction

People with unilateral lower limb amputation (PULLA) often walk asymmetrically (Sagawa et al., 2011; Wedge et al., 2022). Unfortunately, gait asymmetry results in higher metabolic cost of transport (Ellis et al., 2013) and is potentially responsible for secondary disorders that impact quality of life (Sagawa et al., 2011). Lower back pain (LBP) is another frequently cited complication for PULLA, with approximately twice the reported prevalence as the able-bodied population (Hammarlund et al., 2011). Greater LBP prevalence coupled with an increasing number of amputations (Ziegler-Graham et al., 2008) has prompted several biomechanical investigations into the estimated internal demands placed on the lower back during gait in PULLA (Devan et al., 2014).

Lower back vertebral joint forces estimated from musculoskeletal computer models are generally greater in PULLA gait than in able-bodied controls (Shojaei et al., 2016). These increased spinal demands have been attributed to prosthetic limitations that during gait can lead to reduced stance time, greater lateral trunk flexion during prosthetic limb stance, greater forward trunk lean, and more sustained erector spinae activity than in able-bodied gait (Devan et al., 2014; Sagawa et al., 2011). Despite these reported differences and symmetry being a target of therapy, it is still unclear how PULLA's preferred asymmetric gait mechanics relate to lower back demands and ultimately LBP.

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Therefore, it is of clinical importance as to whether lower back demands could be reduced by adjusting the degree of gait asymmetry. To begin addressing this novel question, we used data from a complimentary study (Wedge et al., 2022) to compare lower back L5/S1 vertebral joint forces estimated from a musculoskeletal computer model in PULLA, for both their preferred level of asymmetry and for different levels of gait asymmetry. We hypothesize that the L5/S1 joint forces during the preferred level of asymmetry cannot be significantly lowered by enforcing gait symmetry nor will it be lower at a theoretically lowest force amount, as interpolated from multiple asymmetric gait conditions. We further hypothesize that PULLA will have greater L5/S1 joint forces than an able-bodied comparison group.

2. Methods

Five PULLA and five able-bodied participants (Table 1) took part in this study which was approved by an Institutional Review Board. All participants were healthy adults and provided informed written consent. PULLA participants were limited to people with a transtibial amputation from non-vascular causes at least 1-year post amputation and capable of walking with variable cadences (i.e., >K3 (Gailey et al., 2002)).

Participants attended two laboratory sessions. The first was required to identify preferred walking speed (PWS) (Johnson et al., 2020) and preferred level of asymmetry for each participant, and to acclimate to walking on a treadmill (Treadmetrix, USA) at guided levels of asymmetry. During the second session, whole-body kinematic data (Qualisys AB, Sweden) were collected at 240 Hz. In both sessions, participants first walked on a motorized treadmill with their preferred gait mechanics (i.e., PWS and preferred level of asymmetry) and then performed up to six different randomly presented conditions of gait asymmetry (−10% to 15% in 5% increments [0% = symmetrical]). Preliminary tests revealed the −15% asymmetry condition was potentially unsafe for PULLA, so they were not presented with this condition. All conditions lasted for five minutes and were performed at their PWS. Gait asymmetry levels were defined as the difference between intact (dominant) and prosthetic (non-dominant) leg stance times relative to the combined stance time (Dingwell et al., 1996). Participants were provided a visual line as a target level of asymmetry and their real-time two-stride moving average of asymmetry based on insole foot switches (B&L Engineering Inc., USA) (Wedge et al., 2022).

L5/S1 vertebral joint forces were calculated for each asymmetry level using a full-body OpenSim model evaluated for gait (Banks et al., 2022; Delp et al., 2007). Participant-specific models were all scaled according to marker placement (Wedge, 2019) and body mass and did not consider the prosthetic properties for PULLA. Static optimization (Crownshield and Brand, 1981) was used to balance the kinetic demands across six lower back joints (i.e., L5/S1 thru T12/L1) with 238 individual musculotendon actuators. L5/S1 vertebral joint forces were expressed relative to the local coordinates of the S1 vertebrae. Participant average and peak L5/S1 vertebral joint compression and resultant shear forces were estimated from the ensemble average of three consecutive strides taken from the end of each five-minute trial.

One-sample *t*-tests were applied to compare level of preferred asymmetry of each group relative to zero ($\alpha < .10$ for all analyses) (Curran-Everett and Benos, 2004). For each participant, theoretical lowest L5/S1 vertebral joint forces were calculated as the minimum point of a best-fit quadratic curve across all measured levels of asymmetry (Fig. 1) (Keppel, 1991). L5/S1 forces at the preferred level of asymmetry were then compared separately to both the calculated theoretical minimum and forces during the 0% asymmetry condition within the PULLA group using paired *t*-tests. Pooled t-tests compared the between-group anthropometric, gait characteristics, and L5/S1 forces at their preferred asymmetric gait. All applicable experimental effects were further quantified with corresponding Cohen's *d* tests to accommodate our sample size (Cohen, 1988).

3. Results

The L5/S1 joint forces at the PULLA groups preferred gait were not significantly different than in symmetrical gait (i.e., 0% asymmetry) for any of the four metrics (Table 2). Similarly, there were no significant differences in any of the force metrics at preferred relative to the PULLA groups theoretical lowest values. All corresponding effect sizes for both of the aforementioned comparisons were small to medium.

PULLA and able-bodied group average and peak L5/S1 compression and resultant shear forces at their preferred gait asymmetry level were not significantly different and had corresponding small to medium effect sizes. Groups were not significantly different in mass and height, but able-bodied participants tended to be younger and preferred to walk faster and with less asymmetry (Table 1). Preferred levels of asymmetry were different from symmetric gait (i.e., 0%) in the PULLA group (p-value $< .01$), but not the able-bodied group ($p = .65$; Table 1). Most participants successfully completed all gait conditions (see Supplementarary data for performance metrics), however one PULLA could not complete the −10% and another the +15% asymmetry conditions.

4. Discussion

The L5/S1 vertebral joint forces in PULLA during preferred level of asymmetry were not significantly different than with symmetrical gait

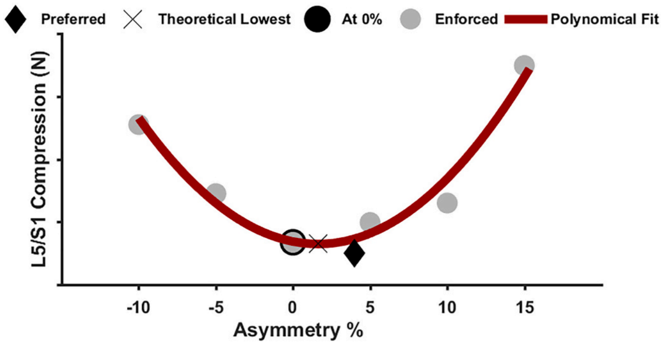

Fig. 1. Illustration representing an example of compression forces across the different asymmetry levels and the quadratic best-fit (*red line*) of all enforced (*grey* and *black circles*) trials and the calculated theoretical lowest force (*X*). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 1
Group average (±SD) anthropometry and preferred gait characteristics.

					Preferred	
	Male: Female (n)	Mass (kg)	Height (cm)	Age (years)	Walking Speed (m/s)	Level of Asymmetry (%)
PULLA	4:1	84.8 ± 22.0	178.6 ± 10.0	39 ± 12	1.1 ± 0.3	4.1 ± 0.9*
Able-Bodied	3:2	74.1 ± 13.4	177.2 ± 8.3	30 ± 6	1.4 ± 0.1	0.7 ± 3.2
p-value		.38	.81	.17	**.02**	**.05**
Cohen's *d*		0.58	0.15	0.96	1.83	1.45

Significant ($p < .10$) between group differences are bolded. Asterisk (*) denotes level of asymmetry significantly different from zero.

Table 2

PULLA L5/S1 vertebral joint force (N) metric averages (±SD) at their preferred level of asymmetry, 0% asymmetry, and theoretical lowest. Columns of *p*-values and Cohen's *d* effect size values reflect within group comparisons made between the corresponding (column to the left) condition and the preferred level. *p*-values and Cohen's *d* effect size values below able-bodied averages reflect between group comparisons at the preferred level of asymmetry.

Group	Preferred Level of Asymmetry	0% Asymmetry	*p*	*d*	Theoretical Lowest	*p*	*d*
Average Compression							
PULLA	895.7 ± 205.9	935.4 ± 272.0	.36	0.46	911.1 ± 227.3	.23	0.64
Able-Bodied	900.8 ± 164.5						
	$p = .97$; $d = 0.03$						
Peak Compression							
PULLA	1299.7 ± 467.0	1391.3 ± 665.5	.48	0.35	1313.8 ± 552.9	.79	0.13
Able-Bodied	1288.3 ± 314.4						
	$p = .97$; $d = 0.03$						
Average Resultant Shear							
PULLA	209.6 ± 58.5	222.1 ± 74.9	.25	0.60	210.3 ± 62.7	.86	0.08
Able-Bodied	191.1 ± 34.4						
	$p = .56$; $d = 0.39$						
Peak Resultant Shear							
PULLA	324.1 ± 133.4	344.2 ± 177.3	.51	0.32	318.5 ± 140.9	.52	0.32
Able-Bodied	278.7 ± 80.4						
	$p = .53$; $d = 0.41$						

and at a theoretical lowest joint force, as hypothesized. This lack of a significant effect potentially demonstrates that lower back demands are already reduced during preferred gait mechanics. The PULLA group joint forces at preferred level of asymmetry were not different from the able-bodied group, which differed from our initial hypothesis.

Asymmetric walking patterns are common, even in those with lower levels of amputation (i.e., transtibial) and who are relatively young and fit, and can result from several factors. Gait asymmetry in PULLA may be attributed to prostheses with passive components unable to produce power and provide stability as well as a biological lower limb (Hafner et al., 2002; Hak et al., 2014). Further, the residual limb / prosthesis interface is challenging, sometimes uncomfortable, and a source of power loss (LaPrè et al., 2018). To reduce cost of transport and discomfort, PULLA tend to spend more time on the intact limb and decrease gait speed (Sagawa et al., 2011). Similarly, able-bodied walkers minimize cost of transport at preferred stride frequencies, gait speeds, and step widths (Donelan et al., 2001; Ralston, 1958; Umberger and Martin, 2007). Our results may suggest that we are unable to improve upon gait lower back demands due to various biological and prosthetic constraints (Sparrow and Newell, 1998).

The L5/S1 vertebral joint force relationships for PULLA and able-bodied groups in this study are in line with some previous reports (Yoder et al., 2015). However, others have reported that PULLA at the transtibial level tend to have larger lower back demands as able-bodied when gait speed is considered (Hendershot et al., 2018). Worth noting, when we normalize to body mass and gait speed, our between group differences at their preferred level of asymmetry were mostly significant (see Supplementaary data).

Peak compression (~1300 N) and resultant shear force (~300 N) magnitudes for both groups were safely below recognized acute failure tolerances of the spinal tissues (Gallagher and Marras, 2012; Jäger et al., 1989). So, if lower back joint forces for PULLA are similar in magnitude to able-bodied, close to optimal, and apparently well below injury tolerances, then why are PULLA more susceptible to LBP? Lower back pain is a complex and multifaceted disorder (Farrokhi et al., 2017). PULLA attribute LBP to prosthetic limitations causing uneven postures and increased fatigue from greater demand on less efficient proximal muscles (Devan et al., 2015; Devan et al., 2017). Repeated and uneven postures can reduce lower back tissue tolerances by not allowing time to recover and by directing forces to tissues that are not intended to bear forces (Gallagher et al., 2005). As a result, even small increases in the loading magnitude, as seen here, when coupled with kinematic changes may initiate injury and lead to LBP.

Experimental limitations of this exploratory work include the low number of participants (Lakens, 2021), the stance-time definition of asymmetry, limited acclimation to the gait asymmetry conditions compared to preferred level of asymmetry, use of a treadmill, participant groups without a history of LBP, and how the joint forces were compared between and within groups (i.e., left unnormalized). In addition, estimating otherwise immeasurable *in vivo* lower back forces with a musculoskeletal computer model has its own inherent limitations (Dreischarf et al., 2016).

The current data are intended to begin assessing the lower back demand ramifications of a common clinical aim to "correct" the preferred level of asymmetry in PULLA by restoring gait symmetry (Darter et al., 2013). Our results suggest that training PULLA to walk more symmetrically may not reduce lower back demands, as they already tend to locomote at a preferred level of asymmetry that seemingly minimizes L5/S1 joint forces. Preferred level of asymmetry may be detrimental and warrant alteration for other reasons, such as gait stability and improving spine kinematics, but that has yet to be established. Therefore, without any recognized advantages, the motivation for PULLA to alter their preferred gait mechanics merits review.

Declaration of Competing Interest

None.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2022.105657.

References

Banks, J.J., Umberger, B.R., Caldwell, G.E., 2022. EMG optimization in OpenSim: a model for estimating lower back kinetics in gait. Med. Eng. Phys. 103 https://doi.org/10.1016/j.medengphy.2022.103790.

Cohen, J., 1988. Statistical Power Analysis for the Behavioral Sciences, 2nd ed. Routledge. https://doi.org/10.4324/9780203771587.

Crownshield, R.D., Brand, R.A., 1981. A physiologically based criterion of muscle force prediction in locomotion. J. Biomech. 14 (11), 793–801.

Curran-Everett, D., Benos, D.J., 2004. Guidelines for reporting statistics in journals published by the American Physiological Society. Am. J. Phys. 287 (97), 457–459. https://doi.org/10.1152/ajpregu.00346.2004.

Darter, B.J., Nielsen, D.H., Yack, H.J., Janz, K.F., 2013. Home-based treadmill training to improve gait performance in persons with a chronic transfemoral amputation. Arch. Phys. Med. Rehabil. 94 (12), 2440–2447. https://doi.org/10.1016/j.apmr.2013.08.001.Home-Based.

Delp, S.L., Anderson, F.C., Arnold, A.S., et al., 2007. OpenSim: open source to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. 54 (11), 1940–1950. https://doi.org/10.1109/TBME.2007.901024.

Devan, H., Hendrick, P.A., Ribeiro, D.C., Hale, L.A., Carman, A.B., 2014. Asymmetrical movements of the lumbopelvic region: is this a potential mechanism for low back pain in people with lower limb amputation? Med. Hypotheses 82 (1), 77–85. https://doi.org/10.1016/j.mehy.2013.11.012.

Devan, H., Carman, A.B., Hendrick, P.A., Hale, L.A., Ribeiro, D.C., 2015. Spinal, pelvic, and hip movement asymmetries in people with lower-limb amputation: systematic review. J. Rehabil. Res. Dev. 52 (1), 1–20. https://doi.org/10.1682/JRRD.2014.05.0135.

Devan, H., Hendrick, P.A., Hale, L.A., Carman, A.B., Dillon, M.P., Ribeiro, D.C., 2017. Exploring factors influencing low back pain in people with nondysvascular lower limb amputation: a national survey. Phys. Med. Rehabil. 9 (10), 949–959. https://doi.org/10.1016/j.pmrj.2017.02.004.

Dingwell, J.B.B., Davis, B.L., Frazier, D.M.M., 1996. Use of an instrumented treadmill for real-time gait symmetry evaluation and feedback in normal and trans-tibial amputee subjects. Prosthetics Orthot. Int. 20 (2), 101–110. https://doi.org/10.3109/03093649609164426.

Donelan, J.M., Kram, R., Kuo, A.D., 2001. Mechanical and metabolic determinants of the preferred step width in human walking. Proc. R. Soc. B Biol. Sci. 268 (1480), 1985–1992. https://doi.org/10.1098/rspb.2001.1761.

Dreischarf, M., Shirazi-Adl, A., Arjmand, N., Rohlmann, A., Schmidt, H., 2016. Estimation of loads on human lumbar spine: a review of in vivo and computational model studies. J. Biomech. 49 (6), 833–845. https://doi.org/10.1016/j.jbiomech.2015.12.038.

Ellis, R.G., Howard, K.C., Kram, R., 2013. The metabolic and mechanical costs of step time asymmetry in walking. Proc. R. Soc. B Biol. Sci. 280 (1756) https://doi.org/10.1098/rspb.2012.2784.

Farrokhi, S., Mazzone, B., Schneider, M., et al., 2017. Biopsychosocial risk factors associated with chronic low back pain after lower limb amputation. Med. Hypotheses 108, 1–9. https://doi.org/10.1016/j.mehy.2017.07.030.

Gailey, R.S., Roach, K.E., Applegate, E.B., et al., 2002. The amputee mobility predictor: an instrument to assess determinants of the lower-limb amputee's ability to ambulate. Arch. Phys. Med. Rehabil. 83 (5), 613–627. https://doi.org/10.1053/apmr.2002.32309.

Gallagher, S., Marras, W.S., 2012. Tolerance of the lumbar spine to shear: a review and recommended exposure limits. Clin. Biomech. 27 (10), 973–978. https://doi.org/10.1016/j.clinbiomech.2012.08.009.

Gallagher, S., Marras, W.S., Litsky, A.S., Burr, D., 2005. Torso flexion loads and the fatigue failure of human lumbosacral motion segments. Spine (Phila Pa 1976) 30 (20), 2265–2273. https://doi.org/10.1097/01.brs.0000182086.33984.b3.

Hafner, B.J., Sanders, J.E., Czerniecki, J.M., Fergason, J., 2002. Energy storage and return prostheses: does patient perception correlate with biomechanical analysis? Clin. Biomech. 17 (5), 325–344. https://doi.org/10.1016/S0268-0033(02)00020-7.

Hak, L., van Dieen, J.H., van der Wurff, P., Houdijk, H., 2014. Stepping Aymmetry Amoung individuals with unilateral transtibial limb loss might be functional in terms of gait stability. J. Phys. Ther. 94 (10), 1480–1488 file:///C:/Users/LaptopMax/OneDrive/MMT2014/Scriptie%5Cnideeen/Scriptieidee/ContentServer.pdf.

Hammarlund, C.S., Carlström, M., Melchior, R., Persson, B.M., 2011. Prevalence of back pain, its effect on functional ability and health-related quality of life in lower limb amputees secondary to trauma or tumour: a comparison across three levels of amputation. Prosthetics Orthot. Int. 35 (1), 97–105. https://doi.org/10.1177/0309364610389357.

Hendershot, B.D., Shojaei, I., Acasio, J.C., Dearth, C.L., Bazrgari, B., 2018. Walking speed differentially alters spinal loads in persons with traumatic lower limb amputation. J. Biomech. 70, 249–254. https://doi.org/10.1016/j.jbiomech.2017.11.026.

Jäger, M., Luttmann, A., Jager, M., Luttmann, A., 1989. Biomechanical analysis and assessment of lumbar stress during load lifting using a dynamic 19-segment human model. Ergonomics. 32 (1), 93–112.

Johnson, R.T., Hafer, J.F., Wedge, R.D., Boyer, K.A., 2020. Comparison of measurement protocols to estimate preferred walking speed between sites. Gait Posture 77 (September 2019), 171–174. https://doi.org/10.1016/j.gaitpost.2020.01.007.

Keppel, G., 1991. Design and Analysis: A Researcher's Handbook, 3rd ed. Prentice-Hall Inc.

Lakens, D., 2021. Sample Size Justification. https://doi.org/10.31234/osf.io/9d3yf.

LaPrè, A.K., Price, M.A., Wedge, R.D., Umberger, B.R., Sup, F.C., 2018. Approach for gait analysis in persons with limb loss including residuum and prosthesis socket dynamics. Int. J. Num. Method Biomed. Eng. 34 (4) https://doi.org/10.1002/cnm.2936.

Ralston, H.J., 1958. Energy-speed relation and optimal speed during level walking. Int. Zeitschrift für Angew Physiol. 17 (4), 277–283. https://doi.org/10.1007/BF00698754.

Sagawa, Y., Turcot, K., Armand, S., Thevenon, A., Vuillerme, N., Watelain, E., 2011. Biomechanics and physiological parameters during gait in lower-limb amputees: a systematic review. Gait Posture 33 (4), 511–526. https://doi.org/10.1016/j.gaitpost.2011.02.003.

Shojaei, I., Hendershot, B.D., Wolf, E.J., Bazrgari, B., 2016. Persons with unilateral transfemoral amputation experience larger spinal loads during level-ground walking compared to able-bodied individuals. Clin. Biomech. 32, 157–163. https://doi.org/10.1016/j.clinbiomech.2015.11.018.

Sparrow, W.A., Newell, K.M., 1998. Metabolic energy expenditure and the regulation of movement economy. Psychon. Bull. Rev. 5 (2), 173–196.

Umberger, B.R., Martin, P.E., 2007. Mechanical power and efficiency of level walking with different stride rates. J. Exp. Biol. 210 (18), 3255–3265. https://doi.org/10.1242/jeb.000950.

Wedge, R.D., 2019. Metabolic Cost and Stability of Locomotoin in People with Lower Limb Amputations. https://doi.org/10.7275/14162705.

Wedge, R.D., Sup, F.C., Umberger, B.R., 2022. Metabolic cost of transport and stance time asymmetry in individuals with unilateral transtibial amputation using a passive prostheses while walking. Clin. Biomech. 94 https://doi.org/10.1016/j.clinbiomech.2022.105632.

Yoder, A.J., Petrella, A.J., Silverman, A.K., 2015. Trunk-pelvis motion, joint loads, and muscle forces during walking with a transtibial amputation. Gait Posture 41 (3), 757–762. https://doi.org/10.1016/j.gaitpost.2015.01.016.

Ziegler-Graham, K., MacKenzie, E.J., Ephraim, P.L., Travison, T.G., Brookmeyer, R., 2008. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Arch. Phys. Med. Rehabil. 89 (3), 422–429. https://doi.org/10.1016/j.apmr.2007.11.005.