Journal of Biomechanics 98 (2020) 109449

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Can shank acceleration provide a clinically feasible surrogate for individual limb propulsion during walking?

Noah L. Pieper [a], Michael D. Lewek [b], Jason R. Franz [a,*]

[a] Joint Department of Biomedical Engineering, University of North Carolina at Chapel Hill and North Carolina State University, Chapel Hill, NC, USA
[b] Division of Physical Therapy, University of North Carolina at Chapel Hill, Chapel Hill, NC, USA

ARTICLE INFO

Article history:
Accepted 21 October 2019

Keywords:
Biomechanics
Ankle Power
Wearables
Accelerometers
Push-off

ABSTRACT

Aging and many pathologies that affect gait are associated with reduced ankle power output and thus trailing limb propulsion during walking. However, quantifying trailing limb propulsion requires sophisticated measurement equipment at significant expense that fundamentally limits clinical translation for diagnostics or gait rehabilitation. As a component of joint power, our purpose was to determine if shank acceleration estimated via accelerometers during push-off can serve as a clinically feasible surrogate for ankle power output and peak anterior ground reaction forces (GRF) during walking. As hypothesized, we found that young adults modulated walking speed via changes in peak anterior GRF and peak ankle power output that correlated with proportional changes in shank acceleration during push-off, both at the individual subject ($R^2 \geq 0.80$, $p < 0.01$) and group average ($R^2 \geq 0.74$, $p < 0.01$) levels. In addition, we found that unilateral deficits in trailing limb propulsion induced via a leg bracing elicited unilateral and relatively proportional reductions in peak anterior GRF, peak ankle power, and peak shank acceleration. These unilateral leg bracing effects on peak shank acceleration correlated with those in peak ankle power (braced leg: $R^2 = 0.43$, $p = 0.028$) but those effects in both peak shank acceleration and peak ankle power were disassociated from those in peak anterior GRF. In conclusion, our findings in young adults provide an early benchmark for the development of affordable and clinically feasible alternatives for assessing and monitoring trailing limb propulsion during walking.

1. Introduction

The plantarflexor muscle-tendon units are functionally important for powering the human ankle during daily locomotor activities such as walking. Indeed, they generate up to 50% of the total mechanical power needed for forward propulsion and leg swing initiation, and thereby regulate step length and walking speed (Farris and Sawicki, 2012; Meinders et al., 1998; Neptune et al., 2009; Neptune et al., 2001). While the underlying mechanisms are diverse and multi-factorial, aging and many gait pathologies (e.g. stroke) elicit disproportionate and functionally relevant reductions in ankle power generation and thus declines in trailing limb propulsion (Farris et al., 2015; Franz, 2016; Judge et al., 1996; McGibbon, 2003; Studenski et al., 2011; Winter et al., 1990). For more than 50 years, our field has sought to quantify the complex changes in walking biomechanics due to aging and gait pathology (Marks and Hirschberg, 1958; Murray et al., 1969), in part because these may point to translational opportunities for prevention and rehabilitation (Franz, 2016). However, despite significant advancements in our understanding, quantifying trailing limb propulsion requires sophisticated measurement equipment (force plates, 3D motion capture) at significant expense that fundamentally limits clinical translation for diagnostics or gait rehabilitation.

Contemporary biomechanics laboratories have the ability to quantify the timing and magnitude of trailing limb propulsion during walking, and regularly do so at the individual joint (e.g., peak ankle power) and individual limb (e.g., peak anterior ground reaction force, GRF) levels. For example, the anterior component of the GRF during walking contributes to accelerating the body's center of mass during push-off, and bilateral deficits thereof due to age (Franz and Kram, 2013a) or unilateral deficits thereof, for example due to post-stroke hemiparesis (Bowden et al., 2006) and cerebral palsy (Martin Lorenzo et al., 2018), can impact mobility and independence. Inverse dynamics analyses that incorporate 3D motion capture can complement GRF measurements to provide joint-level insight into deficits in trailing limb propulsion during walking. For example, older adults generate 11–35% less peak ankle power and 13–22% less peak anterior GRF than young adults (Browne and Franz, 2018; DeVita and Hortobagyi, 2000; Franz and Kram, 2013a, b, 2014; Judge et al., 1996; Kerrigan et al.,

* Corresponding author at: 10206C Mary Ellen Jones Building, CB 7575, Chapel Hill, NC 27599, USA.
E-mail address: jrfranz@email.unc.edu (J.R. Franz).

https://doi.org/10.1016/j.jbiomech.2019.109449
0021-9290/

1998; Silder et al., 2008; Winter et al., 1990). Moreover, people following stroke exhibit up to 80% less ankle power and up to 66% smaller peak anterior GRF on their hemiparetic side compared to their unaffected limb (Bowden et al., 2006; Farris et al., 2015; Jonkers et al., 2009; Turns et al., 2007). The development and widespread adoption of instrumented treadmills and motion capture systems have made it increasingly easier to quantify trailing limb propulsion as a means to detect the onset and progression of clinical impairment or to quantify the efficacy of gait rehabilitation paradigms. However, while these sophisticated measurements should continue to serve as our gold standard for scientific discovery, their expense represents a distinct barrier to clinical translation or real-world assessment. Thus, there is a need to develop more cost-effective and translationally palatable solutions to the assessment of trailing limb propulsion during walking.

One viable alternative to conventional measures of trailing limb propulsion is the use of wearable sensors. For example, Jiang et al. (2019) recently showed that two inertial measurement units (IMUs) on the foot and shank can reliably estimate patterns of ankle power output during walking. While promising, those estimates require the use of gold-standard measurements to train the requisite machine learning algorithms and thereby calibrate the predictions. Wireless accelerometers in particular provide a promising plug-and-play alternative due to their size, cost, and relatively widespread availability. Their use in gait biomechanics research also has a long and successful history (Mathie et al., 2004). For example, pelvis-mounted accelerometers can be used to estimate gait asymmetry (Barden et al., 2016; Kobayashi et al., 2014; Kobsar et al., 2014; Moe-Nilssen and Helbostad, 2004), while thigh-mounted accelerometers have been used to estimate surface incline while walking (Uyanik et al., 2015) and to monitor physical activity (Wullems et al., 2017). Given their close proximity to the ankle, shank-mounted accelerometers in particular may have high potential to serve as surrogates for trailing limb propulsion, especially in their application to deficits arising from reduced ankle power output. Several prior studies have used shank acceleration to estimate functionally relevant biomechanical outcomes, including average walking speed (Bishop and Li, 2010; Li et al., 2010) and gait cycle events (Jasiewicz et al., 2006; Maqbool et al., 2016). However, no study to our knowledge has tested the efficacy of shank-mounted accelerometers as a surrogate for trailing limb propulsion. Using a biofeedback paradigm, our lab recently showed that, independent of their age, people modulate trailing limb propulsion more by increasing ankle angular velocity than ankle moment (Browne and Franz, 2019). This outcome suggests that a local kinematic signal such as shank acceleration may vary in proportion to more conventional measurements – namely, peak ankle power output and anterior GRF.

Therefore, our purpose was to determine if shank acceleration estimated via accelerometers during push-off can serve as a clinically feasible surrogate for trailing limb propulsion during walking. We first hypothesized that inducing modifications in walking speed would elicit changes in shank acceleration that would correlate with changes in trailing limb propulsion measured via peak anterior GRF and peak ankle power. Second, we hypothesized that unilateral deficits in trailing limb propulsion induced via leg bracing would elicit reductions in shank acceleration that would correlate with proportional changes in peak anterior GRF and peak ankle power.

2. Methods

2.1. Participants

Twelve unimpaired young adult subjects participated after providing written, informed consent according to the University of North Carolina Biomedical Sciences Institutional Review Board. Subjects had a mean ± standard deviation age of 21.1 ± 1.2 years, height of 1.77 ± 0.12 m, and body mass of 69.7 ± 12.9 kg. All subjects were free of neurologic impairments and musculoskeletal injuries.

2.2. Experimental protocol

Using a photo cell timing system separated by 2 m in the middle of a 10 m walkway, we determined each subjects' preferred overground walking speed (i.e., PWS) as the average of 3 times taken to walk the full distance. Once the PWS was calculated, each participant completed a five minute warmup at PWS on a dual-belt, force-measuring treadmill (Bertec, Columbus, OH) to allow their movement patterns to stabilize. Subjects then walked on the treadmill for 2 min each at PWS (1.37 ± 0.21 m/s), at one fixed speed slower than preferred (0.75 m/s), and at one fixed speed faster than preferred (1.75 m/s). Subjects also walked at PWS while wearing unilateral knee (TROM Advance, Donjoy, Dallas, TX) and ankle (Slimline, Darco, Huntington, WV) braces on the right limb. The purpose of this leg bracing was to impose diminished right limb propulsion that would represent that commonly associated with unilateral gait pathology (e.g., following stroke). Previous studies have successfully incorporated leg bracing as a model for unilateral deficits in trailing limb propulsion characteristic of gait pathology such as that following stroke or in people with cerebral palsy (Lewek et al., 2012; Wutzke et al., 2012).

2.3. Measurement and analysis

A 14-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) operating at 100 Hz recorded pelvis and lower extremity kinematics via 17 anatomical markers and an additional 14 tracking markers affixed using rigid clusters. A standing trial, as well two unilateral hip circumduction trials used to estimate functional hip joint centers (Arnold et al., 2010), also included markers placed on the left and right medial knee and malleoli. We also affixed one tri-axial accelerometer (Delsys Trigno, Natick, MA) operating at 1000 Hz on the right and left anterior tibia placed at 50% of shank length.

Marker trajectories and ground reaction forces (i.e., GRF) were filtered using 4th order low-pass Butterworth filters with cutoff frequencies of 6 Hz and 20 Hz, respectively. We then used the static standing calibration and functional hip joint centers to scale a seven segment, 18 degree-of-freedom model of the pelvis and right and left legs (Arnold et al., 2010). We used the filtered marker and force data to estimate ankle joint angle, moment, and power using an inverse dynamics routine described in detail previously (Browne and Franz, 2019). Due to errors in marker data, we report these inverse kinematics and dynamics results for 11 of the 12 subjects in our cohort. Consistent with the translational goals of this line of research, we opted to extract our outcome measures from the raw, unfiltered accelerometer data to test the efficacy of those signals to provide meaningful information about limb propulsion outside the lab and perhaps in real-time. Similar rationale has been utilized and explained previously (Arvidsson et al., 2019). Each sensor collected three-dimensional accelerometer data, which formed the components of a composite acceleration vector which we used to calculate the total acceleration magnitude as a function of trial duration. In order to synchronize the acceleration data with our anterior-posterior GRF and ankle power outcome measures, we used heel-strikes and toe-off events identified from the vertical GRF signals.

Using a 20 N threshold on the vertical GRF, we collated the composite acceleration magnitude, anterior GRF, and ankle power time series from 60 individual right and 60 individual left strides. We extracted our outcome measures of interest both on a stride-by-stride basis and from the bilateral, stride-averaged profiles and focused our analysis to the time of trailing limb push-off. Thus, we determined the efficacy of shank acceleration as a surrogate for stride-by-stride variations in conventional measures of trailing limb propulsion within an individual subject, and for variations thereof at the cohort level, respectively. During and immediately after the push-off phase of walking, we extracted peak acceleration magnitude and more conventional measures of trailing limb propulsion – namely, peak stance phase ankle power at the joint level and peak anterior GRF at the limb level. Specifically, focused our analysis between midstance and the end of initial swing (i.e., 30–73% gait cycle) based on gait cycle phases defined by Perry (1992). Finally, we also analyzed peak dorsiflexion and plantarflexion angles as secondary outcome measures.

2.4. Statistical analysis

Shapiro-Wilks tests confirmed normal distributions for each outcome measure. To analyze walking speed effects (i.e., Hypothesis 1), we first used a repeated measures analysis of variance (ANOVA) to test for significant main effects of walking speed on the bilateral, stride-averaged outcome measures. When a significant main effect was found, we focused pairwise post-hoc tests on comparisons of slower/faster trials versus walking at PWS. In addition, we calculated Pearson correlations between each outcome measure: (*i*) for each subject on an individual stride-by-stride basis and then averaged those coefficients and the resulting p-values across our cohort, and (*ii*) at the population level using the bilateral, stride-averaged data. To analyze leg bracing effects (i.e., Hypothesis 2), a series of paired-samples t-tests first evaluated the effects of unilateral leg bracing compared to normal walking at subjects preferred speed. Also, we correlated each outcome measure extracted from the braced leg and the difference thereof compared to unbraced measures during normal walking. We report effect sizes as partial eta square and Cohen's d for the ANOVA and t-tests, respectively. For all analyses, we defined significance using an alpha value of 0.05.

3. Results

Representative stride profiles for each outcome measure across changes in walking speed are shown in Fig. 1. Compared to values when walking at their preferred speed, subjects increased (decreased) peak anterior GRF by 34% (48%), peak ankle power by 41% (60%), and peak shank acceleration by 37% (52%) to increase (decrease) walking speed to 1.75 m/s (0.75 m/s), respectively (main effects, p-values <0.001, $\eta_p^2 \geq 0.911$) (Fig. 2). Pairwise comparisons revealed that these outcome measures at speeds slower (i.e., 0.75 m/s) and faster (i.e., 1.75 m/s) than preferred reached significance compared to PWS (p-values <0.001, $d \geq 2.06$). Individual subject stride-to-stride (group average) regressions revealed that 91% (74%) and 80% (80%) of the variance in peak anterior GRF and peak ankle power across walking speeds was explained by that in peak shank acceleration, respectively (Fig. 3, Table 1).

Leg bracing systematically and significantly reduced right limb propulsion (Fig. 2). Specifically, compared to values when walking at PWS, leg bracing decreased right leg peak anterior GRF, peak ankle power, and peak shank acceleration by an average of 51%, 84% and 44%, respectively (p-values $\leq$0.001, $d \geq 4.00$). Braced leg peak shank acceleration was positively correlated with peak shank acceleration ($R^2 = 0.43$, $p = 0.028$) but not with peak anterior GRF ($R^2 = 0.31$, $p = 0.061$) (Fig. 4). Conversely, contralateral peak shank acceleration and peak ankle power were unaffected by leg bracing (pairwise, p-values $\geq$0.22, $d \leq 0.37$), while contralateral peak anterior GRF exhibited a modest 3.9% increase compared to normal walking ($p = 0.03$). Leg bracing also caused notable effects in peak dorsiflexion, which decreased unilaterally by 71% compared to normal walking ($p < 0.001$, $d = 1.53$).

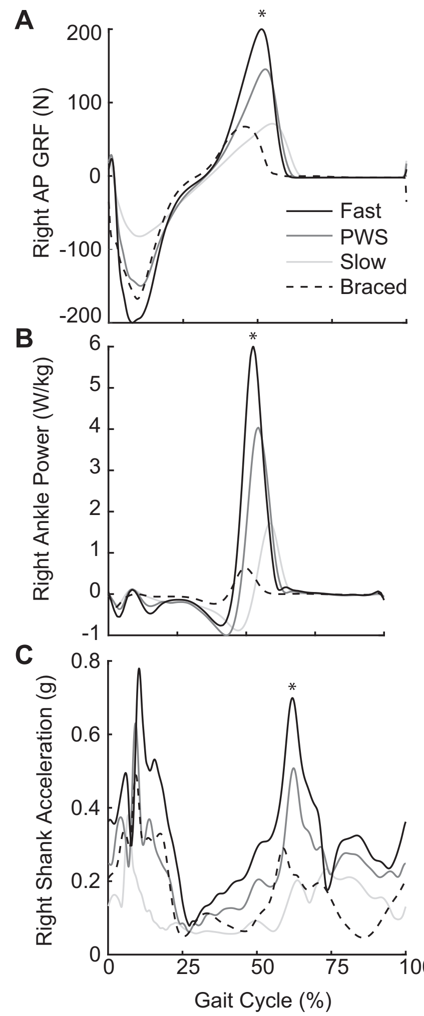

Fig. 1. Group average profiles for (A) anterior-posterior (AP) ground reaction force (GRF) (n = 12), (B) ankle power generation (n = 11), and (C) shank acceleration (n = 12) for each experimental condition plotted across an averaged gait cycle, from heel-strike to heel-strike. Conditions included PWS (preferred walking speed), Fast (1.75 m/s), Slow (0.75 m/s), and Braced (walking with a unilateral leg brace). Asterisks (*) indicate a significant main effect of condition on local maximum values during and immediately after the push-off phase of walking (p < 0.05).

4. Discussion

Characteristic reductions in trailing limb propulsion can limit walking performance and thereby independence. The purpose of this study was to determine if shank acceleration estimated via accelerometers during push-off can serve as a clinically feasible surrogate for trailing limb propulsion during walking. Consistent with our first hypothesis, our results reveal that young adults modulate walking speed via changes in peak anterior GRF and peak ankle power output that correlate with proportional changes in shank acceleration during and immediately after push-off. In addition, in partial support of our second hypothesis, we show that unilateral deficits in trailing limb propulsion prescribed via a leg brace elicited unilateral and relatively proportional reductions in peak anterior GRF, peak ankle power, and peak shank acceleration. How-

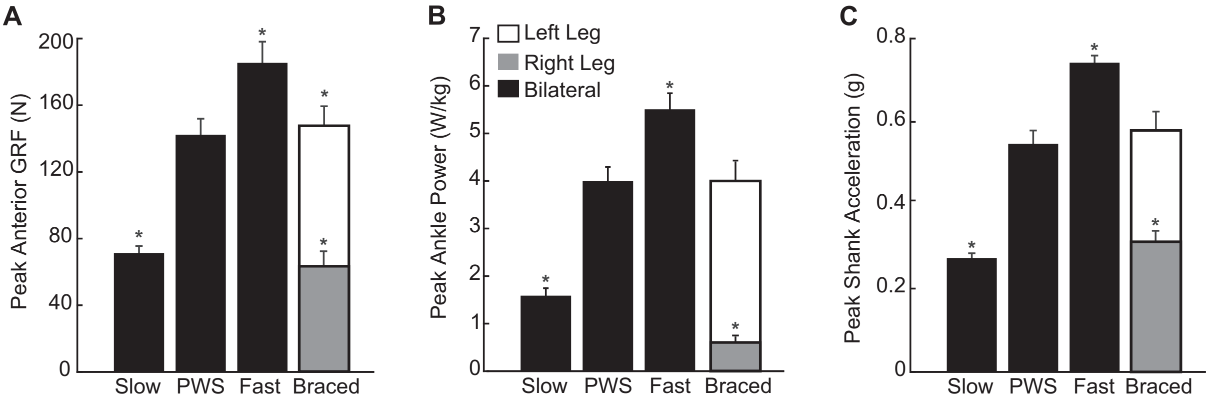

Fig. 2. Group average (standard error) (A) peak anterior ground reaction force (GRF) (n = 12), (B) peak ankle power (n = 11), and peak shank acceleration (n = 12) during and immediately following push-off. Conditions included PWS (preferred walking speed), Fast (1.75 m/s), Slow (0.75 m/s), and Braced (walking with a unilateral leg brace). Asterisks (*) indicate a significant pairwise comparison versus normal walking (p < 0.05).

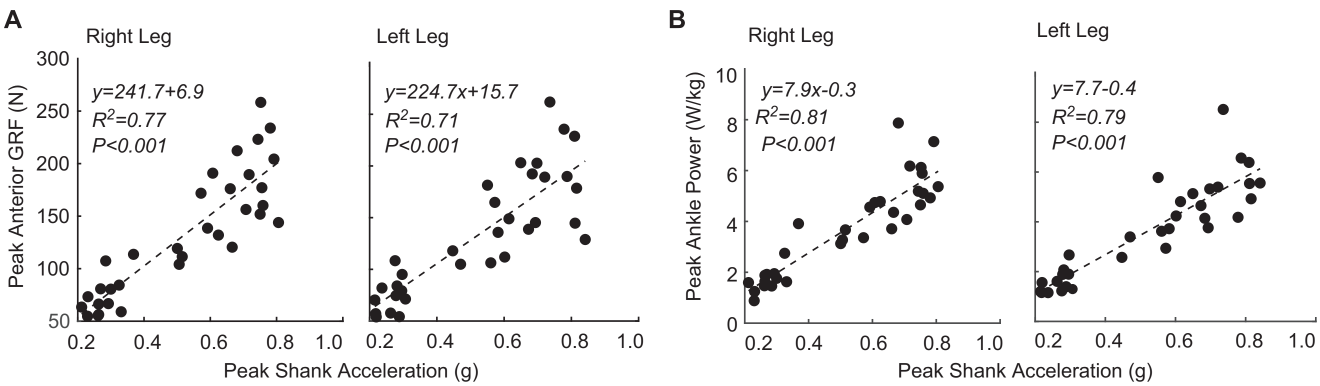

Fig. 3. Linear regression between peak shank acceleration during and immediately after push-off and (A) peak anterior ground reaction force (GRF) and (B) peak ankle power generation pooled across all subjects and walking speeds performed separately for the right and left legs.

Table 1
Correlations with shank acceleration across walking speeds.

Outcome Measure	Limb		r	p
Peak Anterior GRF	Right	Individual	0.955	<0.001
		Cohort	0.880	<0.001
	Left	Individual	0.953	<0.001
		Cohort	0.844	<0.001
Peak Ankle Power	Right	Individual	0.889	<0.001
		Cohort	0.902	<0.001
	Left	Individual	0.902	<0.001
		Cohort	0.889	<0.001

GRF: ground reaction forces. Individual: correlations performed across all strides taken by an individual subject. Cohort: correlations performed on subject-averaged data across all subjects.

ever, leg bracing effects on peak shank acceleration were more strongly associated with peak ankle power than peak anterior GRF.

Our measures of shank acceleration during walking agree well with available literature. For example, our stride-averaged profiles have similar patterns to those reported previously; there are two distinguishable peaks – one following heel-strike and another during and immediately after push-off (Bishop and Li, 2010; Havens et al., 2018; Hsu et al., 2018; Jasiewicz et al., 2006). Given the proximity of the accelerometers' placement to the ankle, the timing of this second peak appears to coincide reasonably well with local maximums in peak anterior GRF and peak ankle power output commonly associated with push-off. The most notable difference was that peak shank acceleration across conditions appeared to be consistently delayed compared to those in more conventional measures of trailing limb propulsion. This temporal delay, which yields a local maximum immediately following toe-off, likely arising from the cause-effect relation underlying how measures of limb propulsion ultimately yield a peak in shank acceleration.

Changing walking speed to that slower and faster than preferred elicited systematic changes in shank acceleration during and immediately following push-off that generally mirrored those in peak ankle power output and peak anterior GRF. These findings are important, as peak ankle power output and peak anterior GRF are key determinants of walking performance for which an inex-

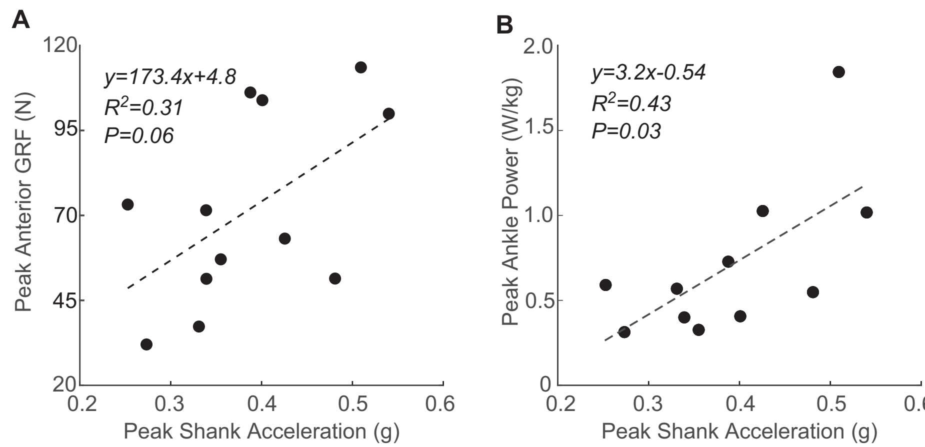

Fig. 4. Linear regressions to determine the effects of right leg bracing on associations between braced leg peak shank acceleration and (A) peak anterior ground reaction force (GRF) and (B) peak ankle power pooled across subjects.

pensive, "real-world" assessment would be highly desirable. Walking speed effects on anterior GRF and ankle power were consistent with those from previous studies (Farris and Sawicki, 2012; Nilsson and Thorstensson, 1989). As a major outcome of this study, the effect of walking speed on more conventional measures of trailing limb propulsion correlated with peak shank acceleration for the cohort stride averages, as well as within individual subjects on a step by step basis. We would thereby interpret our findings to suggest that shank acceleration may effectively distinguish intra-subject variations over time, for example due to functional decline or clinical rehabilitation. Moreover, our ability to do the same at the cohort level suggests that we can distinguish between group effects in the real-world that may be associated with clinically significant reductions in walking speed.

Unilateral leg bracing effectively imposed unilateral deficits in trailing limb propulsion similar to that one would anticipate in those with gait pathology such as stroke. These deficits were also effectively measured via peak shank acceleration. Indeed, contralateral peak acceleration and peak ankle power were largely unchanged compared to usual walking. Only peak anterior GRF exhibited a modest increase in comparison to normal walking. Interestingly, we found that leg brace effects on peak shank acceleration correlated with peak ankle power but not peak anterior GRF, perhaps governed in part due to the proximity of the accelerometer to the ankle joint. However, a post-hoc analysis revealed that peak ankle power itself was not significantly correlated to peak anterior GRF ($R^2 = 0.20$, $p = 0.169$), alluding more to a disassociation between joint- and limb-level effects of the leg brace. For example, inter-limb coordination during double support between trialing limb ankle power output and leading leg hip extensor power output was not included in this study but is likely relevant to the measured peak anterior GRF. Although the strength of the correlation was less compelling, these results do complement and extend those for walking speed effects, and suggest the promising potential for shank acceleration signals to also identify asymmetries in trailing limb propulsion. Furthermore, these asymmetries are functionally meaningful, having been implicated in, for example, the higher metabolic cost of walking in people following stroke compared to controls (Penke et al., 2019). Taken together, our results warrant field testing in people with deficits in trailing limb propulsion toward further establishing ecological validity and translational utility.

We also recognize other opportunities that would arise from a plug-and-play wearable surrogate for estimating trailing limb propulsion. Biofeedback has shown promise to enhance trailing limb propulsion in the laboratory, for example when cueing instantaneous changes in peak anterior GRF (Browne and Franz, 2018; Genthe et al., 2018; Schenck and Kesar, 2017) and in peak ankle power output (Browne and Franz, 2019) when walking on an instrumented treadmill. In addition, recall of those effects transfers to faster overground walking speeds (Browne and Franz, 2019). To our knowledge, shank acceleration has not yet been systematically investigated in a biofeedback or gait retraining paradigm, but could provide a necessary bridge outside of the laboratory setting to promote clinical translation. While it may ultimately be feasible that the novel machine learning algorithms recently introduced by Jiang et al. (2019) could be leveraged for similar translational purposes, the tuning required from such methodology introduces distinct challenges which may limit the ease of translation. In either event, it is evident that this field is *accelerating* on several fronts towards promising discoveries in gait rehabilitation.

We acknowledge several limitations of this study. First, we studied healthy young subjects with no impairments in trailing limb propulsion. Our findings warrant the need for future studies in older adults or individuals with gait pathology. In addition, future studies will also need to overcome challenges in identifying reliable timing data to identify heel-strike and toe-off events outside of the laboratory. Moreover, imposing a unilateral impairment using a leg brace likely fails to replicate that arising from neuromusculoskeletal limitations following gait pathology such as stroke. Here, the participating young adults continued to walk at their usual speed, whereas such deficits would in reality likely be met with slower self-selected speeds. Another limitation was that we opted to use a single accelerometer placement on the anterior shank at 50% of shank length, and we are thus unable to make specific recommendations for whether other placements would be more or less suitable for estimating trailing limb propulsion. In addition, the timing differences between the local maxima of shank acceleration and those in more conventional measures of limb propulsion may present challenges when leveraged to promote gait retraining. As one example, a shift to a more hip dominant strategy for limb propulsion and swing initiation may effectively increase shank acceleration with counterproductive effects on ankle power output. Finally, unlike the more sophisticated analytical procedures (i.e., machine learning) performed by Jiang et al. (2019), we opted to use simple correlations from the peak amplitudes while making no attempt to estimate stride average profiles of our laboratory-based measures in absolute units. Rather, we focused here on a single-sensor, more plug-and-play solution based on peak amplitudes toward immediate clinical application, for example via step-to-step biofeedback. However, such an approach comes at the expense of some precision. Ultimately, effective technological solutions should be well-matched to the clinical need at hand but no more sophisticated and complex than needed to promote widespread adoption and feasibility.

In conclusion, our findings in young adults provide an early benchmark for the development of affordable and clinically feasible alternatives for assessing and monitoring trailing limb propulsion during walking. We largely accept our hypotheses and reveal that shank acceleration during and immediately following the push-off phase of walking may serve as an effective and simple surrogate for more conventional, laboratory-based measures of trailing limb propulsion (i.e., peak anterior GRF and peak ankle power). While motion capture and instrumented treadmills will continue to serve as the gold standard in laboratory settings, those technologies cannot meet the widespread need for inexpensive, accessible, and plug-and-play solutions. Thus, our research is important for bridging the gap between scientific discovery and clinical translation.

Acknowledgements

We would like to thank Michael G. Browne for his assistance in this design of this experiment.

Declaration of Competing Interest

The authors have no conflicts of interest to disclose.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2019.109449.

References

Arnold, E.M., Ward, S.R., Lieber, R.L., Delp, S.L., 2010. A model of the lower limb for analysis of human movement. Ann. Biomed. Eng. 38, 269–279.
Arvidsson, D., Fridolfsson, J., Borjesson, M., Andersen, L.B., Ekblom, O., Dencker, M., Brond, J.C., 2019. Re-examination of accelerometer data processing and calibration for the assessment of physical activity intensity. Scand. J. Med. Sci. Sports 29 (10), 1442–1452.
Barden, J.M., Clermont, C.A., Kobsar, D., Beauchet, O., 2016. Accelerometer-based step regularity is lower in older adults with bilateral knee osteoarthritis. Front. Hum. Neurosci. 10, 625.
Bishop, E., Li, Q., 2010. Walking speed estimation using shank-mounted accelerometers. International Conference on Robotics and Automation.
Bowden, M.G., Balasubramanian, C.K., Neptune, R.R., Kautz, S.A., 2006. Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. Stroke 37, 872–876.
Browne, M., Franz, J.R., 2018. More push from your push-off: joint-level modifications to modulate propulsive forces in old age. PLoS One 13, e0201407.
Browne, M.G., Franz, J.R., 2019. Ankle power biofeedback attenuates the distal-to-proximal redistribution in older adults. Gait Posture 71, 44–49.
DeVita, P., Hortobagyi, T., 2000. Age causes a redistribution of joint torques and powers during gait. J. Appl. Physiol. 88, 1804–1811.
Farris, D.J., Hampton, A., Lewek, M.D., Sawicki, G.S., 2015. Revisiting the mechanics and energetics of walking in individuals with chronic hemiparesis following stroke: from individual limbs to lower limb joints. J. Neuroeng. Rehabil. 12, 24.
Farris, D.J., Sawicki, G.S., 2012. The mechanics and energetics of human walking and running: a joint level perspective. J. Royal Soc. Interface 9, 110–118.
Franz, J.R., 2016. The age-associated reduction in propulsive power generation in walking. Exerc. Sport Sci. Rev. 44, 129–136.
Franz, J.R., Kram, R., 2013a. Advanced age affects the individual leg mechanics of level, uphill, and downhill walking. J. Biomech. 46, 535–540.
Franz, J.R., Kram, R., 2013b. How does age affect leg muscle activity/coactivity during uphill and downhill walking? Gait Posture 37, 378–384.
Franz, J.R., Kram, R., 2014. Advanced age and the mechanics of uphill walking: a joint-level, inverse dynamic analysis. Gait Posture 39, 135–140.
Genthe, K., Schenck, C., Eicholtz, S., Zajac-Cox, L., Wolf, S., Kesar, T.M., 2018. Effects of real-time gait biofeedback on paretic propulsion and gait biomechanics in individuals post-stroke. Top Stroke Rehabil. 25, 186–193.
Havens, K.L., Cohen, S.C., Pratt, K.A., Sigward, S.M., 2018. Accelerations from wearable accelerometers reflect knee loading during running after anterior cruciate ligament reconstruction. Clin. Biomech. 58, 57–61.
Hsu, W.-C., Sugiarto, T., Lin, Y.-J., Yang, F.-C., Lin, Z.-Y., Sun, C.-T., Hsu, C.-L., Chou, K.-N., 2018. Multiple-wearable-sensor-based gait classification and analysis in patients with neurological disorders. Sensors 18, 3397.
Jasiewicz, J.M., Allum, J.H.J., Middleton, J.W., Barriskill, A., Condie, P., Purcell, B., Li, R.C.T., 2006. Gait event detection using linear accelerometers or angular velocity transducers in able-bodied and spinal-cord injured individuals. Gait Posture 24, 502–509.
Jiang, X., Gholami, M., Khoshnam, M., Eng, J.J., Menon, C., 2019. Estimation of Ankle Joint Power during Walking Using Two Inertial Sensors. Sensors 19 (12), 2796.
Jonkers, I., Delp, S., Patten, C., 2009. Capacity to increase walking speed is limited by impaired hip and ankle power generation in lower functioning persons post-stroke. Gait Posture 29, 129–137.
Judge, J.O., Davis 3rd, R.B., Ounpuu, S., 1996. Step length reductions in advanced age: the role of ankle and hip kinetics. The journals of gerontology. Series A. Biol. Sci. Med. Sci. 51, M303–M312.
Kerrigan, D.C., Todd, M.K., Della Croce, U., Lipsitz, L.A., Collins, J.J., 1998. Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. Arch. Phys. Med. Rehabil. 79, 317–322.
Kobayashi, H., Kakihana, W., Kimura, T., 2014. Combined effects of age and gender on gait symmetry and regularity assessed by autocorrelation of trunk acceleration. J. Neuroeng. Rehabil. 11, 109.
Kobsar, D., Olson, C., Paranjape, R., Hadjistavropoulos, T., Barden, J.M., 2014. Evaluation of age-related differences in the stride-to-stride fluctuations, regularity and symmetry of gait using a waist-mounted tri-axial accelerometer. Gait Posture 39, 553–557.
Lewek, M.D., Osborn, A.J., Wutzke, C.J., 2012. The influence of mechanically and physiologically imposed stiff-knee gait patterns on the energy cost of walking. Arch. Phys. Med. Rehabil. 93, 123–128.
Li, Q., Young, M., Naing, V., Donelan, J.M., 2010. Walking speed estimation using a shank-mounted inertial measurement unit. J. Biomech. 43, 1640–1643.
Maqbool, H.F., Husman, M.A.B., Awad, M.I., Abouhossein, A., Mehryar, P., Iqbal, N., Dehghani-Sanij, A.A., 2016. Real-time gait event detection for lower limb amputees using a single wearable sensor. 38th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC).
Marks, M., Hirschberg, G.G., 1958. Analysis of the Hemipalegic Gait. Ann. NY Acad. Sci. 74, 59–77.
Martin Lorenzo, T., Rocon, E., Martinez Caballero, I., Lerma Lara, S., 2018. Medial gastrocnemius structure and gait kinetics in spastic cerebral palsy and typically developing children: a cross-sectional study. Medicine 97, e10776.
Mathie, M.J., Coster, A.C.F., Lovell, N.H., Celler, B.G., 2004. Accelerometry: providing an integrated, practical method for long-term, ambulatory monitoring of human movement. Physiol. Measure. 25, R1–R20.
McGibbon, C.A., 2003. Toward a better understanding of gait changes with age and disablement: neuromuscular adaptation. Exerc. Sport Sci. Rev. 31, 102–108.
Meinders, M., Gitter, A., Czerniecki, J.M., 1998. The role of ankle plantar flexor muscle work during walking. Scand. J. Rehabil. Med. 30, 39–46.
Moe-Nilssen, R., Helbostad, J.L., 2004. Estimation of gait cycle characteristics by trunk accelerometry. J. Biomech. 37, 121–126.
Murray, M.P., Kory, R.C., Clarkson, B.H., 1969. Walking patterns in healthy old men. J. Gerontol. 24, 169–178.
Neptune, R.R., Clark, D.J., Kautz, S.A., 2009. Modular control of human walking: a simulation study. J. Biomech. 42, 1282–1287.
Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J. Biomech. 34, 1387–1398.
Nilsson, J., Thorstensson, A., 1989. Ground reaction forces at different speeds of human walking and running. Acta Physiol. Scand. 136, 217–227.
Penke, K., Scott, K., Sinskey, Y., Lewek, M.D., 2019. Propulsive forces applied to the body's center of mass affect metabolic energetics poststroke. Arch. Phys. Med. Rehabil. 100, 1068–1075.
Perry, J., 1992. Gait Analysis: Normal and Pathological Function. SLACK inc., Thorofare, N.J..
Schenck, C., Kesar, T.M., 2017. Effects of unilateral real-time biofeedback on propulsive forces during gait. J. Neuroeng. Rehabil. 14, 52.
Silder, A., Heiderscheit, B., Thelen, D.G., 2008. Active and passive contributions to joint kinetics during walking in older adults. J. Biomech. 41, 1520–1527.
Studenski, S., Perera, S., Patel, K., Rosano, C., Faulkner, K., Inzitari, M., Brach, J., Chandler, J., Cawthon, P., Connor, E.B., Nevitt, M., Visser, M., Kritchevsky, S., Badinelli, S., Harris, T., Newman, A.B., Cauley, J., Ferrucci, L., Guralnik, J., 2011. Gait speed and survival in older adults. JAMA 305, 50–58.
Turns, L.J., Neptune, R.R., Kautz, S.A., 2007. Relationships between muscle activity and anteroposterior ground reaction forces in hemiparetic walking. Arch. Phys. Med. Rehabil. 88, 1127–1135.
Uyanik, I., Khatri, A., Majeti, D., Ugur, M., Shastri, D., Pavlidis, I., 2015. Using accelerometer data to estimate surface incline and its walking app potential. 33rd Annual ACM Conference Extended Abstracts on Human Factors in Computing Systems.
Winter, D.A., Patla, A.E., Frank, J.S., Walt, S.E., 1990. Biomechanical walking pattern changes in the fit and healthy elderly. Phys. Ther. 70, 340–347.
Wullems, J.A., Verschueren, S.M.P., Degens, H., Morse, C.I., Onambélé, G.L., 2017. Performance of thigh-mounted triaxial accelerometer algorithms in objective quantification of sedentary behaviour and physical activity in older adults. PLOS ONE 12, e0188215.
Wutzke, C.J., Sawicki, G.S., Lewek, M.D., 2012. The influence of a unilateral fixed ankle on metabolic and mechanical demands during walking in unimpaired young adults. J. Biomech. 45, 2405–2410.