BIOFEEDBACK GAIT RETRAINING UNDER REAL-WORLD RUNNING CONDITIONS

CHAN YAU SHAN ZOE

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Biofeedback Gait Retraining under Real-World Running Conditions

CHAN YAU SHAN ZOE

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CERTIFICATE OF ORIGINALITY

I hereby declare that this thesis is my own work and that, to the best of my knowledge and belief, it reproduces no material previously published or written, nor material that has been accepted for the award of any other degree or diploma, except where due acknowledgement has been made in the text.

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Chan Yau Shan Zoe

ABSTRACT

Gait retraining has been used as an intervention to mitigate the risk of running-related injuries among distance runners. Lab-based gait retraining studies have used various feedback strategies to modify the running gait and have demonstrated promising results in changing biomechanical parameters associated with injuries. However, lab-based training is not accessible to the general running population, and there was limited evidence that supports the transfer of training effect to conditions that resemble real-world running. For these reasons, gait retraining in runners' natural training environments might be preferred. The main objective of this thesis was to optimize the training protocol for gait retraining under real-world conditions. To this end, five studies were conducted to address three specific aims: 1) identify the limitations of lab-based gait retraining protocols, 2) assess habitual gait adaptations in real-world running, and 3) establish the technical specifications for systems used to modify gait outside of the lab. Regarding the first specific aim, two studies were conducted to examine the transfer of training effect to untrained conditions, including overground and slopes. Results of both studies suggested that runners who regularly train overground and on slopes may not benefit fully from lab-based training. Based on such findings, gait retraining along overground running routes that include slopes was recommended. Our third study addressed the second specific aim, it examined the natural biomechanical adaptations while running on slopes. By analyzing real-world training records, changes in speed and cadence were observed along slopes. These habitual adaptations could interact with the training parameters in gait retraining, subsequently affecting the training effect. Therefore, an adaptive feedback model with the training target set based on different sloped conditions was recommended. With the existing wearable technology, tibial acceleration can be measured using wireless accelerometers outside of the lab. Tibial acceleration is a common parameter used as feedback during training and as

an outcome measure to assess the effect of gait retraining. The fourth and fifth studies addressed the third specific aim and presented the technical considerations required for accurate and reliable tibial acceleration measurements under conditions that resemble training in the real world. Based on the findings of these two studies, accelerometers with an operating range wider than ±16-*g* were recommended for accurate tibial acceleration measurements. Also, it was recommended to measure a minimum of 100 consecutive strides during each session when evaluating training performance to ensure reliable measurements. To conclude the findings of the five studies, a gait retraining protocol designed for training under real-world conditions was proposed and evaluated in a proof-of-concept study. This final study demonstrated the feasibility of using adaptive feedback in real-world training using a wearable sensor system. Tibial acceleration was measured under real-world training conditions and was used as feedback to guide the runner in modifying the gait pattern while training along slopes. Overall, the findings of this thesis provided insights for further optimization of gait retraining protocols and the future development of feedback systems suitable for use under real-world conditions.

PREFACE

Chapter 3 is based on a manuscript prepared for submission to *Clinical Biomechanics*. The study has also been presented at the XXVII Congress of the International Society of Biomechanics (ISB-ASB 2019), 31 Jul - 4 Aug 2019, Calgary, Canada.

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The proof-of-concept study presented in **Chapter 8** has been presented at the XXIX Congress of the International Society of Biomechanics (ISB-JSB 2023), 30 Jul – 3 Aug, Fukuoka, Japan. Chan Z.Y.S., Cheung R.T.H., Tsang S.M.H., Ferber R. Adaptive biofeedback gait retraining on slopes: a proof-of-concept study.

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

LIST OF ABBREVIATIONS

PTA	Peak tibial acceleration
IMU	Inertial measurement unit
RRI	Running-related injury
RCT	Randomized controlled trial
GRF	Ground reaction force
VALR	Vertical average loading rate
VILR	Vertical instantaneous loading rate
RFS	Rearfoot strike
FFS	Forefoot strike
3D	Three-dimensional
MFS	Midfoot strike
FSA	Foot-strike angle
g	Gravitational constant (9.81 m/s^2)
ANOVA	Analysis of variance
BW	Body weight
SD	Standard deviation
MDD	Minimal detectable difference
AccR	Resultant acceleration
$CI_{95\%}$	95% confidence interval

CHAPTER 1

INTRODUCTION

1.1 Rationale

Around the world, distance running is among the top five most popular physical exercises.[1] However, the risk of sustaining an injury is high among runners.[2,3] Research attention has been given to identifying risk factors of injuries through biomechanical analysis,[2,4,5] and mitigating the risk factors through various interventions, of which one is modifying the gait pattern.[6] Gait retraining has demonstrated promising results in modifying kinetics, kinematics, and spatiotemporal parameters among runners.[5,7] Conventional gait retraining involves motorized treadmill running augmented with visual, auditory or haptic feedback to facilitate gait modification.[7,8] From an injury prevention perspective, it is important for the runner to retain the gait modifications while running in their typical training environments. However, there is a lack of evidence that demonstrates the transfer of training effects from treadmills to conditions that better resemble real-life running environments. Furthermore, motor learning has been suggested to be most effective within conditions that closely match those required during performance.[9] Therefore, gait retraining conducted within a runner's typical training environment may be most beneficial in the long term.

1.2 Objective and specific aims

The main objective of this thesis was to develop a gait retraining protocol for runners to modify their gait in real-world environments. To address this objective, five studies were conducted through three specific aims which include:

- Identify the limitations of conventional gait retraining protocols;
- Assess habitual gait adaptations in real-world running; and

- Establish the technical specifications for assessing and modifying gait in real-world running conditions.

Regarding the first specific aim, Studies 1 and 2 (Chapters 3 and 4) were performed to examine the carry-over effects of conventional gait retraining to untrained conditions. Both studies examined the change in foot-strike pattern, cadence and loading rates after a program targeting foot-strike pattern modification. Study 1 aimed to examine the carry-over effects of treadmill-based gait retraining to overground running and Study 2 aimed to examine the carry-over effects of level surface training to sloped running.

Study 3 (Chapter 5) was conducted to address the second specific aim. Study 3 assessed the natural adaptations on running speed and cadence when runners were training on slopes in the real world.

Studies 4 and 5 (Chapters 6 and 7) were designed to establish protocols for valid and reliable measurements of peak tibial acceleration (PTA) using wireless inertial measurement units (IMUs) outside of the lab. Study 4 examined the between-day reliability of the continuous measurement of PTA along an overground track. Study 5 evaluated the performance of a low operating range accelerometer and a signal restoration algorithm on PTA measurements.

A gait retraining protocol designed to reduce PTA during overground running on slope conditions was proposed based on the findings of Studies 1 to 5. To evaluate the proposed protocol, a proof-of-concept study was presented as part of the general discussion in Chapter 8.

1.3 Organization of the thesis

The doctoral thesis is presented in a manuscript-based style. The content of the thesis is organized as follows:

Chapter 2 summarizes background information relevant to the research topic;

Chapters 3 to 7 are based on the five studies completed, these chapters have been modified from manuscripts that have been published or prepared for publication to better suit the flow of this thesis. These chapters may contain redundant information;

Chapter 8 is the concluding chapter. It summarizes and discusses the findings of Studies 1 to 5, and presents the results of a proof-of-concept study based on the proposed gait retraining protocol. This chapter also presents the limitations and directions for future research.

1.4 Ethical statements

The studies presented in Chapters 3 to 8 were approved by relevant ethical committees.

Chapter 3: This study has been approved by the departmental research committee, Department of Rehabilitation Sciences of The Hong Kong Polytechnic University (HSEARS20161017001).

Chapter 4: This study has been approved by the departmental research committee, Department of Rehabilitation Sciences of The Hong Kong Polytechnic University (HSEARS20190928001).

Chapter 5: Data used for this study was extracted from the Wearable Technology Citizen Science Level-4 secure research database housed at the University of Calgary. The collection of data through this database was approved under the study title "The Wearable Technology Citizen Science Program" by the Conjoint Health Research Ethics Board at the University of Calgary (REB20-0572_REN2).

Chapter 6: This study has been approved by the Conjoint Health Research Ethics Board at the University of Calgary (REB16-1183).

Chapter 7: This study has been approved by the Human Research Ethics Committee at Western Sydney University (H14514).

The proof-of-concept study presented in Chapter 8: This study has been approved by the Conjoint Health Research Ethics Board at the University of Calgary (REB22-0931).

CHAPTER 2

BACKGROUND

2.1 Gait retraining

2.1.1 Overview of running gait retraining

Running gait retraining aims to alter an individual's habitual motor pattern. Alternating a motor pattern that has been reinforced for years requires guidance and practice.[6,10] External cues, often described as biofeedback, provide information about performance and the intended changes. The runner responds to external cues during practice.[10] Biofeedback helps the runner develop a connection between the external cue and the internal sensory cues associated with the desired motor pattern.[11] This motor learning process can lead to permanent changes. In practice, real-time biofeedback gait retraining has demonstrated promising results in alternating running kinetics and kinematics in distance runners.[5,7,12]

2.1.2 Running-related injuries and gait retraining

From a clinical perspective, running gait retraining has been posited as a viable approach to mitigate the risk of running-related injuries (RRIs).[13–15] Theoretically, gait retraining can modify running biomechanics that contributes to RRIs, resulting in a "safer" running style. However, there is no consensus on an injury-free running style,[16] and evidence linking biomechanical risk factors and RRIs is limited and conflicting.[17]

The origins of RRIs are complex. Possible interactions between intrinsic factors (e.g., running biomechanics, injury history, anthropometric measures) and extrinsic factors (e.g., training intensities, running shoes) could affect the overall risk of injury.[2,17,18] Furthermore, different RRIs could have different injury mechanisms, and biomechanical risk factors are likely specific to the type of injury.[17] The complexity and specificity of RRIs might explain the inconsistent findings that associate running biomechanics with the overall injury risk within the literature.[17] Nevertheless, a recent study conducted a comprehensive review of risk factors of common RRIs and suggested reducing vertical loading rate as a strategy to reduce the risk of plantar fasciitis and tibial stress fracture,[17] the second and third most common RRIs in long-distance runners.[19]

The clinical benefit of reducing loading rate through gait retraining has been demonstrated in two randomized controlled trials (RCTs).[14,15] Runners were trained to land softer, and both studies reported reductions in vertical loading rates after the training. Moreover, injury incidences within one year after training were 62% lower in the gait retraining group compared to the control group,[14] and 64.6% lower compared to the year before the training.[15] In spite of the fact that gait retraining studies have presented positive results in reducing injury risk among runners, it should not be treated as a standalone approach. Experts in running research recommended gait retraining be used in conjunction with traditional interventions, including flexibility training and strength conditioning.[13,20] Other extrinsic risk factors, such as errors in training and the influence of footwear should also be addressed when modifying gait through training.[13,18]

2.1.3 Common kinetic outcome measures

Many gait retraining studies did not track injury incidence after the training, instead, biomechanics were compared before and after the training or against a control group.[12] This type of study determines the outcome measures based on established biomechanical risk factors of common RRIs. Common outcome measures are impact-related, including vertical loading rates and PTA.[12,20] Elevated loading rates and PTA have been associated with patellofemoral pain, plantar fasciitis and tibial stress fracture in specific populations.[4,17,21,22]

Force plates, either embedded within an instrumented treadmill or placed along an overground runway, are typically used to measure ground reaction force (GRF). Loading rates are calculated from the vertical GRF curve between the initial foot-ground contact and the impact peak. The specific region used for the calculation differs slightly between calculation methods.[23] A common method was to obtain the average and maximum slope between 20 to 80% of the vertical GRF magnitude at the impact peak as vertical average and instantaneous loading rates (VALR and VILR) respectively,[24] as shown in Figure 2.1a. Two retrospective studies and one prospective study have found higher VALR[4,22,25] and VILR[22,25] in injured rearfoot strike (RFS) runners with all types of RRIs. In studies which analyzed a specific type of RRI, higher VALR and VILR were found in female RFS runners with a history of tibial stress fracture,[21,26,27] and plantar fasciitis.[28,29] It is important to note that most studies were conducted on female runners, and all but one study[28] were conducted on RFS runners who make initial contacts with their heels. The association between high VALR/VILR with RRI could be limited to a specific group of runners. In fact, Johnson *et al*. found no significant difference in VALR and VILR between forefoot strike (FFS) runners who were injured and the FFS control group.[25] Even so, experimental reduction in VALR and/or VILR, through gait retraining or footwear,[1] have been found to reduce injury risk in various populations. Among gait retraining studies conducted on healthy runners within the past 12 years, at least 12 of the studies have measured VALR and/or VILR as kinetic outcome measures (Table 2.1).

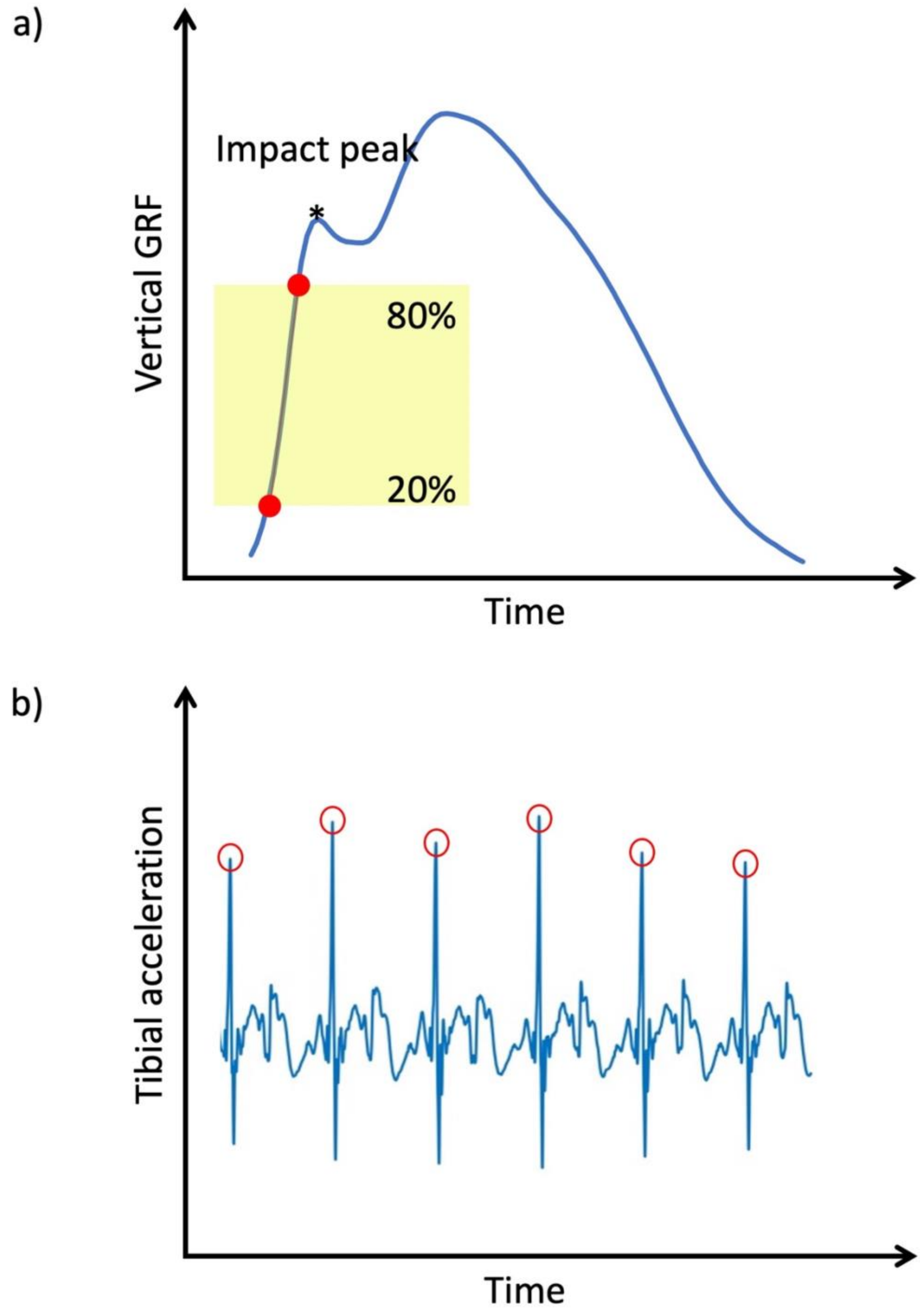


Figure 2.1. Schematic diagrams showing the calculations of a) loading rate from vertical ground reaction force (GRF) and b) peaks from axial tibial acceleration.
* marks the impact peak, shaded region indicates 20 to 80% of impact peak and circles mark the peaks in tibial acceleration.

Study	Trial population	Training target	Kinetic outcomes			
			VALR	VILR	Axial PTA	Resultant PTA
Bowser et al. 2018[30]	19 runners; RFS; axial PTA ≥ 8 *g*	50% of baseline axial PTA	↓	↓	↓	
Chan et al. 2018[14]	320 runners (C:154); VALR ≥ 70 BW/s	Reduce or diminish impact peak of vertical GRF	↓	↓		
Chan et al. 2020[31]	20 runners; RFS	MFS	=	=		
Cheung et al. 2018[32]	16 runners; PPA at heel ≥ 10 *g*	80% of baseline PPA at heel	↓	↓		
Ching et al. 2018[33]	16 runners; PPA at heel ≥ 8 *g*	80% of baseline PPA at heel	↓	↓	↓	
Clansey et al. 2014[34]	22 runners (C:10); RFS; axial PTA > 9 *g*	50% of baseline axial PTA	↓	↓	↓	
Crowell and Davis 2011[35]	10 runners; RFS; axial PTA ≥ 8 *g*	50% of baseline axial PTA	↓	↓	↓	
Derie et al. 2022 & Van den Berghe et al. 2022[36,37]	20 runners (C:10); RFS; axial PTA > 9 *g*	70% of baseline axial PTA		↓	↓	
Futrell et al. 2020[38]	15 runners; RFS; cadence ≤ 170 spm	FFS	↓	↓		

	18 runners; RFS; cadence ≤ 170 spm	7.5% above baseline cadence	=	=		
Hafer et al. 2015[39]	6 runners; RFS; cadence: 150 – 170 spm	10% above baseline cadence	↓			
Sheerin et al. 2020[40]	18 runners; Resultant PTA ≥ speed-specific threshold	90% of resultant PTA at beginning of each session				↓
Wang et al. 2020[41]	24 runners (C:12); RFS	7.5% above baseline cadence	↓	↓		
Willy et al. 2016[42]	30 runners (C:14); VILR ≥ 85 BW/s	7.5% above baseline cadence	↓	↓		
Yang et al. 2020[43]	17 runners (C:8); Non-FFS	FFS	↓			
Zhang et al. 2019[44]	15 runners; axial PTA > 8 *g*	80% of baseline axial PTA			↓	
Zhang et al. 2019[45]	13 runners	80% of baseline axial PTA			↓	

C, number of subjects in control group (no feedback); VALR, vertical average loading rate; VILR, vertical instantaneous loading rate; PTA, peak tibial acceleration; *g*, gravitational constant (g = 9.81 m/s^2); BW, body weight; RFS, rearfoot strike; MFS, midfoot strike; FFS, forefoot strike; spm, steps per minute.
↓ significant reduction after the training
= no significant difference between baseline and post-training assessments

Table 2.1. Summary of gait retraining studies on healthy runners with kinetic outcome measures.

Another common kinetic outcome measure of gait retraining is PTA.[12,20] Tibial acceleration is measured using segment-mounted accelerometers. Typically, a skin-mounted accelerometer is affixed to the distal tibial, with one of the axes aligned to the long axis of the tibia.[46] Axial tibial acceleration refers to the time-domain acceleration signal aligned to the long axis of the tibia; resultant tibial acceleration refers to the resultant vector of all three (x-, y- and z-) axes. The value of the peaks in the axial and resultant tibial acceleration at the time of initial contact are identified as axial PTA and resultant PTA respectively.[24,47,48] Figure 2.1b shows an example of peaks identified from the axial tibial acceleration signal. Tibial acceleration is commonly used as a surrogate measurement for GRF when force plates are not available.[46] Significant correlation has been reported between axial PTA and loading rates. Zhang *et al.* have reported a moderate to very strong intra-subject correlation between axial PTA and VALR (r = 0.56 – 0.95), and VILR (r = 0.61 – 0.95).[49] Between-subject correlation between axial PTA and VILR have also been reported, with strong to very strong correlations (r = 0.70 – 0.92).[27,50–52] Van den Berghe *et al.* also reported a strong correlation between resultant PTA and VILR.[50]

Compared to loading rates, fewer studies have investigated elevated PTA as a biomechanical risk factor of RRIs. Two retrospective studies found higher axial PTA among female RFS runners with a history of tibial stress fracture compared to healthy controls.[21,27] Interestingly, Zifchock *et al.* found differences in axial PTA between the limbs of runners who recovered from a tibial stress fracture, with the injured limb having a higher axial PTA.[26] However, it should be noted that there is a lack of evidence supporting a reduction in PTA through interventions that could lead to changes in injury risk. Nonetheless, axial PTA is a common kinetic outcome measure, Table 2.1 listed seven gait retraining studies which assessed the change in axial PTA before and after the training.

Resultant PTA has not been investigated as a biomechanical risk factor for RRI and has only been measured in one gait retraining study.[40] However, impact forces are three-dimensional (3D) and tibial acceleration along each axis contributes differently under different conditions.[47,48] Resultant PTA is a metric that takes all measurement axes into account.[47,53]

2.1.4 Training targets and feedback for gait retraining

Among the 17 gait retraining studies (16 independent cohorts) on healthy runners with kinetic outcome measures listed in Table 2.1, the training targets can be categorized into three subgroups: impact, foot-strike pattern and cadence.

Ten studies used an impact-related training target, with axial PTA being the most common. These studies measured the axial PTA at baseline and set the training target as 50 – 80% of the baseline value.[30,34–37,44,45] Biofeedback on axial PTA varied between studies. In four studies, real-time axial PTA was displayed with a line indicating the training target, runners were asked to maintain their tibial acceleration below the line.[30,35,44,45] One group of researchers developed a system which superimpose pink noise onto music tracks, with the volume of the noise proportional to the difference between the measured PTA and the training target.[36,37] Clansey *et al.* provided both visual (i.e., red, yellow and green light) and audio (i.e., high pitch, low pitch and no sound) feedback which indicated axial PTA above 75%, 50 – 75% and below 50% of the baseline value.[34] Similar to Clansey *et al.*, Cheung *et al.* used visual feedback with red and green lights indicating the current acceleration peak above or below the training target;[32] Ching *et al.* used audio feedback with high and low pitch indicating the current acceleration peak above or below the training target.[33] Instead of PTA, Cheung *et al.* and Ching *et al.* used the peak positive acceleration measured at the heel of the shoe,[32,33] a common site for commercially available shoe-mounted devices.[54] Sheerin *et al.* measured resultant PTA before each training session and set the training target as 90% of the value.[40] Runners would receive haptic feedback in the form of a short vibration from the wristwatch whenever the training target was exceeded. One study provided participants with a visual display of the GRF curve, and runners were asked to run softer while trying to reduce or diminish the impact peaks.[14]

Another common training target is to increase cadence. Four studies measured cadence at baseline and set the training target as 7.5 – 10% above the baseline.[38,39,41,42] Among the four studies, three used a metronome to guide runners into meeting the training target,[38,39,41] with one giving runners the choice between a metronome and a music playlist with songs which tempo matches the training target.[39] Willy *et al.* used a shoe-mounted sensor paired with a smartwatch programmed to display cadence visually.[42] All studies found a significant increase in cadence after the training, however, the change in kinetic outcome varies. Willy *et al.* found 18.9 and 17.9% reductions in VALR and VILR when running with an 8.6% increase in cadence.[42] Similarly, Wang *et al.* found 15.6 and 14.7% reductions in VALR and VILR when running with a 5.7% increase in cadence.[41] However, Futrell *et al.* found no significant difference in VALR or VILR with a 7.2% increase in cadence,[38] even though the training target was the same as previously mentioned studies.[41,42]

Three studies trained runners to transition from a RFS to a midfoot strike (MFS) or FFS. Two studies provided runners with minimalist footwear as part of the intervention,[38,43] and trained runners to adopt a FFS running pattern. Futrell *et al.* instructed habitual RFS runners to run with FFS for three minutes before the training and measured tibial acceleration.[38] The peak value measured while running with FFS was recorded and used during the training. An audio cue was provided whenever this value was exceeded, indicating a non-FFS. Yang *et al.* used an insole with force sensors to identify foot-strike patterns and provided an audio cue to runners when a RFS was detected.[43] One study instructed runners to run with MFS.[31] A lab-based motion capture system was used with reflective markers to obtain the foot-strike angle (FSA) during the training. The foot-strike pattern was categorized based on the FSA ranges reported in a previous study.[55] A graphical display of the foot-strike pattern was provided to the runner as visual feedback. All three studies reported reduction in FSA, suggesting a transition to MFS or FFS. However, the reductions in VALR and VILR were only reported in the two studies that trained runners to adopt FFS,[38,43] no significance difference in VALR or VILR was reported in the study that trained runners to adopt MFS.[31]

2.1.5 Screening criteria for gait retraining

Most gait retraining studies have at least one inclusion criterion set according to the outcome measure or training target. The inclusion criteria for each study are listed under trial population in Table 2.1. The criteria were usually set to establish a need for gait retraining. In three gait retraining studies where runners were trained to adjust their foot-strike pattern to MFS or FFS, runners were screened for their habitual foot-strike pattern.[31,38,43] Only RFS or non-FFS runners who may benefit from the training were included. Several gait retraining studies included runners who were considered at risk of a certain type of RRI.[30,34,42] In these studies, a threshold on an impact-related variable (e.g., VILR and axial PTA) was used to screen runners. For example, Bowser *et al.*[30] set a threshold of 8 g (gravitational constant = 9.81 m/s^2) for axial PTA based on a retrospective study which compared the variable between runners with a history of tibial stress fracture and healthy controls.[21] Alternatively, one study set the inclusion criterion to avoid adverse effects caused by the targeted gait modification.[38] In Futrell *et al.*'s study where they trained runners to increase their cadence, runners with a natural cadence of above 170 steps per minute were excluded.[38] The rationale behind the selection criterion was that a further increase in cadence among runners with a naturally high cadence can potentially lead to injury.

2.1.6 Training protocols and limitations

The training protocols for gait retraining studies on healthy runners with kinetic outcome measures are listed in Table 2.2. Most studies trained runners on a treadmill inside the lab. Some training protocols may only be feasible on a treadmill or inside the lab, in particular, those that require a complicated set-up or visual display in the form of a monitor usually placed in front of the treadmill. This type of training protocol usually follows a strict pre-determined schedule, likely based on lab availability and access to necessary equipment. The overall training length was found to be within 2 to 3 weeks, with the number of sessions and training time per session fixed. Moreover, speed was kept constant throughout the training, most studies set the speed to match the participants' usual running speed.[14,30–33,35,38,44,45] One study used a fixed speed of 3.7 m/s,[34] and one study let runners choose between four pre-determined speeds.[40] From a research perspective, highly controlled lab-based gait retraining allows objective assessment of the training effect between runners. However, a major disadvantage is the low accessibility of labs and specialized equipment which may not be available to the general running population.[9] Secondly, an intensive training schedule within the lab could disrupt a runner's regular training routine. This may discourage competitive runners who are training for upcoming races.[20] Lastly, the carry-over of training effect to untrained conditions remains largely unknown. There have only been a few studies that examined the carry-over effect of treadmill-based gait retraining to overground running. Zhang *et al.* assessed the reduction of axial PTA on both overground and treadmill running following eight sessions of gait retraining on a treadmill. This study reported smaller reductions, as evidenced by smaller effect sizes, during overground running compared to the treadmill.[44] Similarly, Sheerin *et al.* also reported a smaller effect size on resultant PTA reduction when runners were assessed overground as compared to the treadmill.[40]

Recently, a research group conducted gait retraining overground.[36,37] They developed a wearable system that provides real-time feedback on axial PTA while runners were running laps along an indoor track. While the training still required the supervision of researchers due to the complexity of the set-up and followed a strict schedule, it has demonstrated the potential of training in more ecologically valid conditions.[56]

Four studies adopted an unsupervised training protocol where runners trained in their own preferred running conditions. Considering that increasing cadence is a modification that can be implemented easily,[39] and was considered less awkward and required less perceived effort to change than alternating foot-strike pattern,[57] it is not surprising that three of these in-field training studies targeted an increase in cadence. Moreover, metronomes or music with a tempo matching the target cadence require relatively simpler setups that can easily be implemented in real-world running conditions. In all four studies with an unsupervised training protocol, none reported control of running speed during the training, even though running speed was provided visually in one of the studies.[42] Results of these studies suggested that speed control during training may not be necessary; self-paced training is not likely to affect the effectiveness of training. Wang *et al.* and Yang *et al.* reported a drop-out/non-compliance rate of 20 and 26.7% within their prescribed 12-week training schedule. The lower drop-out rate in Wang *et al.*'s study might be attributed to the weekly group training sessions designed to ensure compliance.[41] Compliance may be further improved if the training can be integrated into the runner's normal training routine. For example, Hafer *et al.* instructed runners to use the feedback system for at least 50% of their regular training,[39] and Willy *et al.* instructed runners to use the smartwatch for visual display of cadence for their next eight regular training sessions.[42] All participants in these two studies complied with the training protocol, and no drop-out was reported.

Most of the studies listed in Table 2.2 adopted a faded feedback design. Biofeedback was provided constantly during the first half of the training (i.e., the initial two to four sessions), which has been considered the acquisition phase of the training. During the acquisition phase, the runner develops an association between the external cue (i.e., biofeedback) and the internal sensory cues required for the new gait pattern. In the second half of the training, which was considered the transfer phase, feedback was removed for certain periods of time within the training. This fading feedback strategy was designed to prevent reliance on external cues, enhance internalization and promote retention.[6,10,20]

Study	Training condition	Training schedule		Feedback strategy
		Length	Schedule	
Bowser et al. 2018[30]	Treadmill	3 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Chan et al. 2018[14]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Chan et al. 2020[31]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Cheung et al. 2018[32]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Ching et al. 2018[33]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Clansey et al. 2014[34]	Treadmill	3 weeks	6 sessions; 20 minutes	-
Crowell and Davis 2011[35]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Derie et al. 2022 & Van den Berghe et al. 2022[36,37]	Overground; indoor track	3 weeks	6 sessions; 20 minutes	Fading: T3 – T6
Futrell et al. 2020[38]	Treadmill	2 – 3 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8

Hafer et al. 2015[39]	Not controlled	6 weeks	> 50% of their regular training	-
Sheerin et al. 2020[40]	Treadmill	3 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Wang et al. 2020[41]	Not controlled	12 weeks	36 sessions; 5 to 48 minutes	-
Willy et al. 2016[42]	Not controlled	Not controlled	8 sessions; regular training	No feedback: T4, T6 and T8
Yang et al. 2020[43]	Not controlled	12 weeks	36 sessions; 5 to 48 minutes	-
Zhang et al. 2019[44]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8
Zhang et al. 2019[45]	Treadmill	2 weeks	8 sessions; 15 to 30 minutes	Fading: T5 – T8

Fading, feedback progressively removed at specified training session(s) (e.g., T5 refers to training session 5).

Table 2.2. Summary of training protocols for gait retraining studies on healthy runners with kinetic outcome measures.

2.2 Use of wearables in gait analysis

2.2.1 Overview of wearables used for gait analysis

A large-scale survey with 1,997 respondents published in 2017 reported that 86.2% of half-marathon runners have used at least one monitoring device to record their training over the past year.[58] Smartphone apps and GPS-enabled running watches were the most commonly used wearable devices.[58,59] These commercially available devices can record simple metrics such as distance, running time, speed and location of each run.[59] These detailed training records provide accurate data for researchers to objectively analyze the training habits of runners without the risk of recall bias. Our research group conducted a study to examine the impact of COVID-19 restrictions on runners' training habits by analyzing the training records before and after March 2020.[60] It was observed that runners increased their weekly mileage and training frequency during the pandemic when compared to a 9-month period prior to the implementation of COVID-19 restrictions. From an injury perspective, real-world training information obtained from wearable devices can help researchers identify training errors that may contribute to RRIs.[59]

More advanced metrics can be obtained using research-grade wearable systems. The advancement of wearable technology has presented researchers with the option to conduct gait analysis outside of the lab and under real-world conditions. A recent systematic review on wearable systems used in gait analysis summarized a list of biomechanics variables that can be measured using wearable systems.[61] The most common spatiotemporal parameters include ground contact time, cadence and stride length; kinematic parameters include ankle kinematics and foot strike pattern/angle; and kinetics parameters include tibial acceleration and plantar pressure.

The use of wearables to measure variables pertinent to this thesis, including foot-strike pattern, cadence and tibial acceleration, are presented in sections 2.2.2 to 2.2.4.

2.2.2 Foot-strike pattern

Foot-strike patterns can be classified visually into RFS, MFS or FFS using footages obtained from high-speed cameras.[62] It can also be quantified by strike index, the center of pressure in relation to the length of the foot at initial contact.[63] The measurement of the strike index typically requires the use of force plates or pressure mats to obtain the center of pressure. The sagittal FSA measured with two-dimensional or 3D motion capture systems was found to be strongly associated with the strike index.[55] The measurement of FSA have been widely used to identify foot-strike patterns in situations where force plates may not be available.[62,64] Recently, various methods using different wearable instruments have been developed to identify foot-strike pattern outside of the lab. Wearable instruments used to measure foot-strike patterns can be categorized into accelerometer/IMU-based and pressure/force sensing insole-based.[61]

Accelerometers have been used to detect foot-strike patterns. Giandolini *et al.* fixed two uniaxial accelerometers onto a running shoe, one at the heel above the midsole and one over the fifth metatarsal head.[65] The time difference between the acceleration peaks from the two signals was used to classify the foot-strike pattern. This method was validated against FSA obtained by two-dimensional video analysis. The running tests were repeated across different speeds and slope conditions. This study reported an overall accuracy of 83.1 – 86.3% in foot-strike pattern identification and a strong correlation with FSA (r = 0.916) except for extreme FFS.

Another method to detect foot-strike patterns is to use an IMU. An IMU has an accelerometer, a gyroscope and a magnetometer. Two methods used both the tri-axial accelerometer and the tri-axial gyroscope components to identify foot-strike patterns.[66–68] The IMU was fixed on to the dorsal surface of the shoe[68] or the foot.[66,67] The resultant accelerometer signal was used to identify initial contact. One method used the foot angle, calculated by integrating the gyroscope signal, to determine the change in direction of the foot within a short time window around initial contact.[68] This method was validated against the 3D motion analysis system and reported strong correlation across different speeds and foot-strike patterns (r = 0.98). Another method used the direction of angular rotation within 15 ms of initial contact to identify foot-strike patterns.[66,67] This method was validated against FSA measured by two-dimensional video analysis. This study reported a 92.2% accuracy in distinguishing between a RFS and a non-RFS and a strong correlation with FSA (r = 0.868).

Pressure sensors or force sensors embedded into insoles can also be used to identify foot-strike patterns. Pressure insoles (e.g., Pedar, Novel, Munich, Germany) can record the location of the center of pressure at initial contact and be used to compute the strike index and foot-strike pattern.[69] A low-cost alternative would be to use force sensors embedded in the insole at the heel and the second metatarsal.[70] The time difference between the onset of the two force sensors can be used to detect foot-strike patterns. This method was validated across different slope conditions and a strong correlation was found with the strike index (R^2 = 0.84).

2.2.3 Cadence

Cadence is among the most commonly reported variables in gait analysis using wearables.[61] Cadence refers to the number of foot-strikes within a minute. It can be calculated based on the number of gait events (e.g., initial contact, toe-off, resultant PTA peaks) over a known period of time. Multiple methods of gait event detection, using IMUs or force plates,[71,72] can be used to calculate cadence inside and outside of the lab. In general, cadence could be accurately measured by wearable devices mounted at the foot/shoe, the tibia and the lower-back.[61]

2.2.4 Tibial acceleration

Measures of tibial acceleration, including axial and resultant PTA, have been reported in at least 30 studies that used accelerometers or IMUs.[46,61] The devices used in these studies vary in a range of parameters including the operating range and sampling frequency.[46] Both parameters could affect the validity of PTA measurements.

The operating range is the upper and lower limits of the signal magnitude that can be captured, exceeding this limit will result in clipped signals.[46] Mitschke *et al.* measured axial PTA using an accelerometer where participants were asked to run in three different shoe conditions at 3.33 m/s. This study reported up to 28.17% decrease in accuracy for axial PTA measured using a low operating range accelerometer (± 8-*g* vs ± 70-*g*) and recommended research studies to use accelerometers with an operating range of ±16-*g* or larger for measurement of axial PTA.[73] However, clipped signals have still been reported in IMUs with an accelerometer operating range of ±16-*g*.[74,75]

A recent study assessed the reliability of axial and resultant PTA measurement using a range of acceleration sampling frequencies on concrete and grass running. This study recommended a minimum sampling frequency of 199 Hz for axial and resultant PTA measurements.[76] Another recommendation was made in Sheerin *et al.*'s review.[46] Power spectral analyses have revealed the predominance of tibial acceleration signal power was concentrated below 60 Hz, based on the Nyquist theory and signal noise contributed by human motion, Sheerin *et al.* recommended a minimum sampling frequency of 300 Hz for tibial acceleration measurements.

Apart from the device parameters, the placement and alignment of the accelerometer can also affect the validity and reliability of PTA measurements. A recent study found that a small proximal shift of 2 cm of the accelerometer could result in a lower PTA.[77] Axis alignment affects the reliability of axial PTA measurements, while resultant

PTA is less sensitive to alignment error and was found to be more reliable between sessions.[47,53] Overall, axial and resultant PTA have shown excellent within-session reliability and good between-session reliability within the lab.[50,78] However, reliability for real-world running have yet to be established.

2.3 Biomechanical differences between lab-based and real-world running

2.3.1 Running under real-world conditions

The advancement in wearable technology has offered new opportunities for researchers to analyze the running gait under real-world running conditions. However, "real-world running conditions" can vary between runners, including differences in location (i.e., indoor or outdoor) and running surface (e.g., treadmill, grass, concrete, track). While it is difficult to define a universal "real-world running condition" that applies to all runners, there are conditions that are generally considered a better representation of runners' usual running conditions. In section 2.3.2, the differences in biomechanical variables relevant to this thesis between treadmill and overground running are presented; and in section 2.3.3, the biomechanical adaptations during sloped running are presented.

2.3.2 Treadmill vs. overground running

The use of motorized treadmills within biomechanical research is popular.[79,80] When using treadmills for running gait assessments, researchers get better control of the environment, easy instrumentation to maintain a certain speed and slope, and higher test replicability.[81] However, treadmill running may not be fully representative of a natural running condition for the majority of runners. Large-scale surveys on recreational runners found less than 6% of runners predominantly train on a treadmill,[60,82,83] most runners prefer training on the road, trail or track.[83] The generalizability of treadmill running biomechanics to overground running remains equivocal.[79]

In a systematic review of biomechanical differences between treadmill and overground running, FSA was found to be significantly smaller during treadmill running than overground track running.[79] In the review, two studies were considered. Wank *et al.* compared FSA between indoor overground track running and treadmill running at 4 and 6 m/s. Significantly smaller FSA was observed in treadmill running for both running speeds, compared to overground running (Table 2.3). Chambon *et al.* also compared FSA between overground running along a lab runway and treadmill running, with different shoe conditions varying in rearfoot midsole thickness. Among all shoe conditions, FSA was significantly smaller in treadmill running than in overground running (Table 2.3). The difference in foot-strike pattern between overground and treadmill running was further demonstrated in a recent study. Lafferty *et al.* used a drone to capture sagittal views of runners while running overground along an outdoor track and on a treadmill, three raters proficient in clinical gait analysis identified the foot-strike pattern of each condition through the videos.[84] Exact foot-strike pattern for each condition was not reported in the study, however, the foot-strike pattern had only moderate agreement between the conditions.

Study	Participants	Running speed (m/s)	Overground	Treadmill
			Foot-strike angle (º)	
Wank et al. 1998[85]	10 male runners	4 (OS: ± 0.05 m/s)	17.6 ± 12.5	6.7 ± 10.8
		6 (OS: ± 0.05 m/s)	6.9 ± 10.5	1.3 ± 8.9
Chambon et al. 2015[86]	12 male runners (RFS: 11; MFS: 1)	Preferred speed (OS: ± 5%)	a) 16.1 ± 8.3	5.2 ± 8.6
			b) 19.6 ± 8.9	8.0 ± 8.3
			c) 20.3 ± 8.3	11.4 ± 5.6
			Cadence (steps per minute)	
Elliott and Blanksby 1976[87]	12 male runners	5.41	182.4 ± 10.2	188.4 ± 12.6
	12 female runners	5.29	184.2 ± 11.4	204.0 ± 13.8
Wank et al. 1998[85]	10 male runners	4 (OS: ± 0.05 m/s)	159.6 ± 6.6	165.0 ± 7.2
		6 (OS: ± 0.05 m/s)	186.0 ± 9.6	196.2 ± 9.6
Riley et al. 2008[88]	20 runners (female: 10)	10-km race pace; 3.80 ± 0.61 (OS: 3.84 ± 0.64)	170.3 ± 15.8	175.1 ± 11.0
Tao et al. 2019[89]	21 runners (female: 17)	4.5 ± 0.4	175.6 ± 9.2	184.8 ± 12.8

OS, overground speed was controlled within the boundaries; RFS: rearfoot strike runners; MFS: midfoot strike runners.
a, b and c in Chambon *et al.*'s study referred to the midsole thickness of 0-, 4-, and 8-mm drop.

Table 2.3. Selected studies comparing foot-strike angle and/or cadence between overground and treadmill running.

In Van Hooren *et al.*'s review, eight studies were considered for the difference in cadence between treadmill and overground running.[79] Cadence was found to be either comparable or higher in the treadmill condition when the running speed was matched,[79] except for one condition in a study where cadence was found to be lower in treadmill running (treadmill: 189.0 ± 7.8 *vs.* overground: 199.2 ± 7.8 steps per minute) at high speed (6.4 m/s).[90] Table 2.3 presents the results of studies with a significantly higher cadence in treadmill running. Van Hooren *et al.* attributed the higher cadence to insufficient familiarization with treadmill running, which could result in a more cautioned running style with a higher cadence and shortened stride length.[91] An adaptation of no less than five minutes has been recommended.[91] And in both Wank *et al*. and Tao *et al.*'s studies, runners were given less than one minute to adapt to the treadmill before data collection.[85,89] While insufficient familiarization may affect the running style, it is unlikely the sole reason for the difference in cadence. Riley *et al.* also observed significantly higher cadence in treadmill running even though all their participants had previous experience in treadmill running, and were given 3 to 5 minutes to adapt and verbally confirmed to being comfortable prior to data collection.[88] Furthermore, in a recent study, Catalá-Vilaplana *et al.* compared the average cadence over 30-min runs on the treadmill and overground.[81] Higher cadence was still observed in treadmill running with 30 minutes of running. Another plausible cause of the difference in cadence is the difference in running speed perceived by the runner. In Tao *et al*.'s study, even though the treadmill speed was set to match the overground running speed, runners perceived the treadmill speed to be significantly faster.[89] In fact, when runners were asked to match their overground preferred speed on a treadmill, runners were found to run 27.1% slower on the treadmill.[92] Overground and treadmill speeds are perceived differently. In theory, speed is increased by increasing cadence and/or stride length, the increase in cadence observed on the treadmill could be an adaptation to the perceivably faster speed.[89]

A number of studies have compared axial and resultant PTA between treadmill and overground running.[75,79,93–95] Depending on the running surface of the overground condition, axial PTA have been found higher,[75,94–96] comparable[95,97–99] or lower[93] than treadmill running. In studies which compared PTA between relatively shorter indoor runways (30 to 75-m) and treadmills, similar axial PTA was reported,[95,97–99] with one study reporting similar resultant PTA.[95] Alternatively, axial PTA was found to be higher in various overground conditions, including rubberized track,[96] asphalt,[94] grass and sidewalk conditions compared to the treadmill.[95] Resultant PTA was also found to be higher in overground grass and sidewalk conditions.[95] Johnson *et al.* found axial and resultant PTA to be higher when running overground during a marathon compared to a lab-based treadmill test conducted before the race.[75] Sheerin *et al.* reviewed extrinsic factors that affect PTA measurements and suggested that the lower compliance of the treadmill surface compared to concrete runways contributes to the difference in PTA,[46] and yet, noted that the relationship may not be linear. Fu *et al.* found no significant difference in axial PTA between grass, synthetic track and concrete runways even though the three surface has different compliance. Similarly, axial and resultant PTA were comparable between dirt, gravel and paved runways.[100] Another factor that affects PTA measurements is the running speed, a faster running speed has been associated with higher PTA.[46,48,75,101] Instead of restraining runners to the same speed as the treadmill runs, Dillon *et al.* asked runners to run at their typical training pace and found significantly faster speed in the overground conditions.[94] The change in speed in this study may partially contribute to the higher PTA observed.

Although the difference in foot-strike pattern, cadence and PTA measurements between overground and treadmill running were presented separately, they are likely interdependent on each other and other biomechanical differences.[46,79] Furthermore, stride-to-stride variance was found higher in overground running than treadmill running, in terms of spatiotemporal parameters,[102,103] lower-body movement[104] and muscle activation pattern.[102] Considering that gait biomechanics during treadmill running may not fully represent overground running, it is important for studies investigating the effect of interventions, including gait retraining and footwear, to assess runners under conditions that match the typical running conditions.[79]

2.3.3 Sloped running

The vast majority of running studies, treadmill or overground, were conducted on a level surface.[105] And yet, slopes are often found within real-world outdoor running environments, including roads and trails. Recreational trail running has been popular in the US since the 1970s, and its popularity has expanded to Europe, Asia and South America since the 2000s.[106] Understanding the biomechanical changes on slopes is important for studies aiming to alter biomechanics in real-world conditions.

A review published in 2017 reported differences in foot-strike pattern when running uphill and downhill, as compared to a level surface.[105] Gottschall and Kram observed a transition from RFS to MFS in three out of ten runners when running at an incline (+10.5%).[107] Similarly, Lussiana *et al.* compared FSA across a series of grades (0, ±2, ±5 and ±8%) when runners were running in two shoe conditions (i.e. traditional *vs.* minimal) and reported reduced FSA in both shoe conditions, indicating a MFS/FFS transition at +8%.[108] This foot-strike pattern transition has also been observed in more extreme incline conditions during treadmill (+15.8%)[107] and overground running (+14 –

48%).[69] It is important to note that the runners in these studies were habitual RFS runners, the effect of incline on FSA or foot-strike pattern in MFS or FFS runners remains unknown. As for downhill running, most biomechanical studies have reported unaltered foot-strike pattern.[69,107,108] Studies have proposed that foot-strike pattern transition in downhill running could be affected by the runner's experience in trail running[109] or the terrain of the trail,[110,111] yet evidence was not conclusive and presented large between-subject variance.[111]

Changes in cadence between uphill/downhill conditions and level surfaces have been reported in previous studies conducted within the lab.[105] Generally, uphill running has been associated with a higher cadence than level running.[105,107,112,113] For example, Telhan *et al.* found 1.2% higher cadence at +7% grade when running at 3.33 m/s.[112] At higher speeds, Padulo *et al.* also found 4.1 – 12.1% increase in cadence among elite runners at +2% grade and an even greater increase (7.3 – 15.9%) at +7% grade.[113] On the other hand, the difference in cadence between downhill and level running was inconsistent. Two studies found no significant difference in cadence between level and across grades ranging between -15.9 and -2%.[107,112] However, Lussiana *et al.* observed a reduced cadence (0%: 163.2 ± 6.0, -5%: 159.6 ± 7.2, -8%: 159.0 ± 8.4) at -5 and -8% conditions when runners were tested at a slower speed (i.e., 2.72 m/s). It should be noted that in these studies with lab-simulated uphill and downhill conditions,[107,108,112,113] speed was kept constant between conditions; this experimental design has limited the runner's ability to regulate speed.[105,114] To investigate the change in speed in real-world conditions, Townshend *et al.* conducted a time trial among a group of experienced runners, where they were completely free to adjust their speed across an 11-km outdoor route with changes in gradient (±11.7%).[114] On average, runners were running 23% slower when running uphill and 13.8% faster when running downhill. Interestingly, the average cadence was similar across the conditions. In another real-world study where a world-class runner was monitored over a 45-km trail race (-18.5 to 34.7%), similar changes between slope and speed were observed.[110] This study found a significant negative correlation between slope and speed (i.e., slower speed at uphill), in addition, a significant negative correlation between slope and cadence (i.e., lower cadence at uphill).

In two treadmill-based experiments,[115,116] negative correlations were observed between slope and axial PTA (Figure 2.2). In Chu and Caldwell's study,[115] 10 male runners ran at five sloped conditions, and a 23% higher axial PTA was reported during downhill running at -12% grade. Similarly, Hamill *et al*. tested 10 male runners between -9 and +6% at increments of 3%. Compared to the level condition, axial PTA was 30% larger when running downhill at -9% grade, and 23% lower when running uphill at 6% grade.[116] However, this relationship between slope and PTA was more complex under real-world conditions. In Waite *et al*.'s study, a significant interaction was observed between slope condition and surface type.[117] Running on a +4% incline resulted in lower axial PTA than level running in the asphalt condition but not the grass condition. Moreover, natural adaptations to slopes, including changes in foot-strike pattern, running speed and cadence, could also influence axial and resultant PTA measurements. Giandolini *et al.* tested 23 experienced runners along a downhill running trail and monitored the foot-strike pattern and tri-axial tibial acceleration throughout the 6.5-km run.[111] They found that runners adopting a FFS during downhill running have higher axial PTA but lower transverse (antero-posterior axis) and resultant PTA compared to runners adopting RFS. The authors speculated that the tilted position of the lower leg at initial contact (i.e., more vertical for FFS) might contribute to the PTA differences found between foot-strike patterns. Apart from the magnitude of the peaks, the timings of the axial and transverse peaks within the gait cycle were also found different across slopes.[105] Given that the resultant tibial acceleration is a time-series signal, computed as the resultant vector of the tri-axial acceleration, difference in magnitude and timing of the axial and transverse peaks would affect the resultant PTA.[105,111] Cadence has also been found to affect axial and resultant PTA. In a lab-based study, increased cadence has been shown to result in lower axial PTA when running on a level surface at a set speed.[118] This negative correlation has also been observed in a real-world case study where an increased cadence resulted in lower vertical and resultant PTA.[110]

Considering the relationship between slope, foot-strike pattern, speed, cadence, and PTA, future investigations along slopes should be conducted under real-world conditions and runners should be free to self-regulate speed in order to enhance the ecological validity.

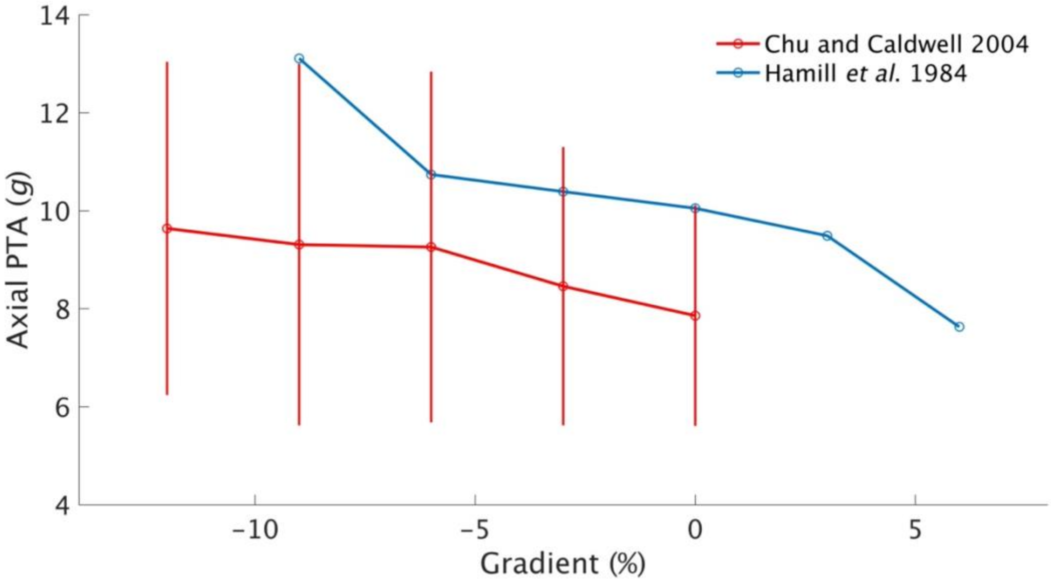


Figure 2.2. Negative correlation between gradient and axial peak tibial acceleration (PTA). Values adopted from Chu and Caldwell 2004[115] and Hamill *et al.* 1984.[116] Standard deviations indicated by vertical lines. PTA are presented in unit of g (gravitational constant, g = 9.81 m/s^2).

2.4 Summary

Lab-based gait retraining has demonstrated promising results in modifying runners' gait patterns and has been posited as an approach to mitigate the risk of RRIs. However, most training protocols were lab-based and runners were trained under a specific condition (i.e., treadmill). Low accessibility and limited transfer of the training effect to real-world running conditions are major drawbacks of lab-based gait retraining. To overcome these limitations, researchers have proposed to incorporate gait retraining within the runners' regular training. With the recent advancement of wearable technology, there is a potential for gait retraining to be conducted outside of the lab. Hence, the main objective of this thesis was to design a training protocol to effectively train runners under real-world conditions.

Few studies have examined the transfer of training effects to untrained conditions, this thesis presents two studies that aimed to identify the limit of transfer to better establish the desired conditions for training. Moreover, natural biomechanical adaptations under real-world conditions could affect the effectiveness of training, but these adaptations were not fully understood. The thesis presents a study that aimed to further examine these biomechanical adaptations. Furthermore, reliability studies are warranted to better inform on the protocol and equipment used for gait assessment under real-world conditions. The thesis presents two studies which examined the validity and reliability of using wearables to measure gait parameters relevant to the proposed gait retraining protocol.

Finally, a training protocol designed for training along outdoor routes with elevation changes was proposed based on the findings of the previous studies. To conclude the thesis, a proof-of-concept study aimed to explore the feasibility of adopting the proposed training protocol was presented. The findings of this thesis provide insights and inform future designs of gait retraining systems that are suitable for use under real-world conditions.

CHAPTER 3

CARRY-OVER EFFECTS OF TREADMILL-BASED GAIT RETRAINING ON FOOT-STRIKE PATTERN, CADENCE AND VERTICAL LOADING RATES DURING OVERGROUND RUNNING

3.1 Introduction

Running gait retraining has demonstrated promising results in modifying kinetics, kinematics and spatiotemporal parameters among healthy runners.[5,7] From a clinical perspective, it has been posited as a viable approach to mitigate the risk of RRIs.[13–15]

Many gait retraining studies did not track injury incidence after the training, instead, only assessed the immediate biomechanical changes. For instance, Cheung *et al.* examined a gait retraining protocol for reducing the risk of running injuries by comparing the changes in PTA and vertical loading rates before and after a 2-week gait retraining where runners were guided by a traffic-light indicator to soften their footfalls.[32] Elevated PTA and vertical loading rates, including VALR and VILR, have been associated with patellofemoral pain, plantar fasciitis and tibial stress fracture.[4,21] Similarly, Willy *et al.* measured and compared peak hip adduction in addition to vertical loading rates before and after an in-field gait retraining that was designed to increase the cadence of runners.[42] Excessive peak hip adduction has been associated with a history of tibial stress fractures and iliotibial band syndrome.[27,119] This type of study determines the training goals and outcome measures based on established biomechanical risk factors of common injuries. The findings of these studies have provided runners and clinicians evidence to inform on adopting such programs for mitigating injury risk. However, in order to obtain optimal clinical benefits, the modified running pattern has to be adopted in real-life running after the training. To date, the effect of gait retraining was mostly assessed only within the trained condition (i.e., treadmills). The carry-over effect to overground running remains largely unknown. There have only been a few studies that examined the carry-over effect of treadmill-based gait retraining to overground running. Sheerin *et al.* and Zhang *et al.* both assessed the reduction in PTA on both overground and treadmill running following a course of gait retraining on a treadmill.[40,44] Both groups reported larger reductions, as evidenced by larger effect sizes, during treadmill than overground running after completion of the gait retraining program.

Several studies have investigated the biomechanical differences between treadmill and overground running. A systematic review found that PTA, vertical loading rates and cadence were comparable between treadmill and overground conditions.[79] However, a larger discrepancy was reported among kinematic outcome measures, such as the FSA. On average, the FSA was found to be significantly lower and more towards a MFS during treadmill running than overground track running by 9.8°.[79] Modifying a runner's foot-strike pattern from RFS to MFS or FFS through gait retraining has been studied for the benefits of reducing vertical loading rates and potentially lowering the risk of injuries.[31,38,120,121] However, it is important to note that these previous studies have only conducted treadmill-based assessments. Taking into account the large discrepancy in FSA between treadmill and overground running,[79] it is unclear whether runners who were trained on a treadmill to change their foot-strike pattern would demonstrate the same changes in overground running.

This study aimed to examine biomechanical changes during both treadmill and overground running following a treadmill-based gait retraining. The gait retraining was designed for habitual RFS runners to transition to MFS. The primary variable of interest was FSA, a quantitative measure of foot-strike pattern. The secondary outcomes included cadence, VALR and VILR. It was hypothesized that participants would be able to run with a reduced FSA (e.g., towards MFS) with reduced vertical loading rates after the training, and the training effect would be larger during treadmill than overground running.

3.2 Methods

3.2.1 Participants

The sample size required was estimated using G*POWER 3.1 (Universitat Kiel, Germany). According to a previous study with an identical training protocol, an effect size of 1.06 for the primary variable of interest and a responder rate of 40% were used for the estimation.[31] With alpha set at 0.05, a sample of 12 runners would be sufficient to achieve a power of 0.8 in the present study. Recreational runners who were actively involved in running for more than 10 km per week and habitually adopted RFS were recruited from running clubs in Hong Kong. The exclusion criteria were set as follows: 1) any active lower-extremity injuries within the previous 6 months or other musculoskeletal or neurological conditions that might affect natural running gait and 2) actively participating in gait modification training. All potential participants underwent an initial screening, and based on a previously reported screening protocol,[31] runners with RFS for over 90% of the footfalls were included. All eligible participants were given a detailed explanation of the experiment before signing an informed consent approved by the departmental research committee, Department of Rehabilitation Sciences of The Hong Kong Polytechnic University (HSEARS20161017001).

3.2.2 Initial screening

The initial screening, gait assessments and gait training were conducted in the Gait and Motion Analysis Lab at the Hong Kong Polytechnic University. A self-reported training speed used for a typical 30-minute training session was recorded for each participant. This speed was used for all assessments and training sessions. Two reflective markers were firmly affixed onto each participant's right foot on the surface of the shoe, with one on the heel and one on the second metatarsal head, according to the model established in a previous study.[55] Three-dimensional marker positions were recorded using an 8-camera motion capture system (Vicon, Oxford, UK) with the participant standing still on a flat, level surface. The participants were instructed to run on an instrumented treadmill (force-sensing tandem treadmill, AMTI, Watertown, MA, USA) at the recorded speed while marker trajectories and GRF were recorded at 200 Hz and 1,000 Hz respectively for five minutes using the motion capture system. The foot-strike pattern was categorized based on the FSA ranges reported in Altman and Davis's study,[55] detailed computations of FSA and foot-strike pattern classification are described in section 3.2.5. Runners with RFS for over 90% of the footfalls were considered eligible for the study.

3.2.3 Gait assessment

Following the initial screening and a 15-minute rest, all eligible participants were evaluated in a baseline assessment session, wearing their usual running shoes. The same pair of shoes was used for each participant throughout the entire experiment. Two reflective markers were affixed at the same position as the initial screening. The assessment was separated into two parts, a treadmill assessment and an overground assessment, which were conducted in a randomized order.

For the treadmill assessment, the participant completed a 5-minute running bout at the determined speed. Marker trajectories and GRF of the last 10 right footfalls were sampled at 200 Hz and 1,000 Hz respectively using the motion capture system.

The overground assessment was conducted using two in-ground force plates (Total area: 0.4 x 1.2 m^2; AMTI, Watertown, MA, USA) embedded within a 10-m runway and a 10-camera motion capture system (Vicon, Oxford, UK). Five practice trials were allowed before data collection for participants to adapt to the running condition and the determined speed (±5%). The running speed of the trials was monitored by speed timing gates (Fusion

Sport, Brisbane, Australia) placed 4.2 m apart. Marker trajectories and GRF were recorded for 10 successive trials with the right foot striking the force plates. The sampling frequency was set to be identical to the treadmill assessment.

A post-training assessment was conducted one hour after the last training session.[24] The procedure was identical to the baseline assessment, with the sequence of the treadmill and overground running assessment randomized.

3.2.4 Gait retraining

Participants were informed of the three foot-strike patterns (i.e., RFS, MFS and FFS) and were instructed to modify their foot-strike pattern and maintain a MFS pattern with the help of the visual feedback. The gait retraining was conducted on the instrumented treadmill.

At the beginning of each session, five reflective markers were affixed on the right shoe and leg of the participant (Figure 3.1), following the marker model used during the initial screening and reference markers at the lateral malleolus, lateral surface of the shank and lateral femoral epicondyle. The three reference markers were used to enhance the auto-tracking performance. Marker positions were recorded when the participant stood still on the treadmill. During training, marker trajectories and GRF of the five reflective markers were live-streamed from the motion capture system to MATLAB (R2019a, The MathWorks, Inc., MA, USA). At each right foot-strike, determined by GRF exceeding 10 N,[35] the FSA was calculated by subtracting the sagittal angle during standing from that measured during running. The foot-strike pattern was categorized based on the FSA ranges: RFS > 8°; 8° ≥ MFS ≥ -1.6°; FFS < -1.6°.[55] The visual feedback comprised of a graphical display of the foot-strike pattern and a 3-letter label (Figure 3.2), which was displayed on a monitor placed in front of the treadmill. The feedback was refreshed after each right foot-strike.

The training program has a total of eight sessions over two weeks, each last between 15 and 30 minutes. Visual feedback was provided continuously for the first four sessions. In the last four sessions, the feedback was progressively removed for two to 28 minutes to enhance motor learning and retention. This protocol was adopted from previous gait retraining studies.[14,31,35] None of the participants received any other types of feedback on foot-strike patterns outside of the training.

Figure 3.1. Marker model used for detection of foot-strike pattern.
Figure shows three reference markers and two markers (circled in red) used for computation of foot-strike angle.

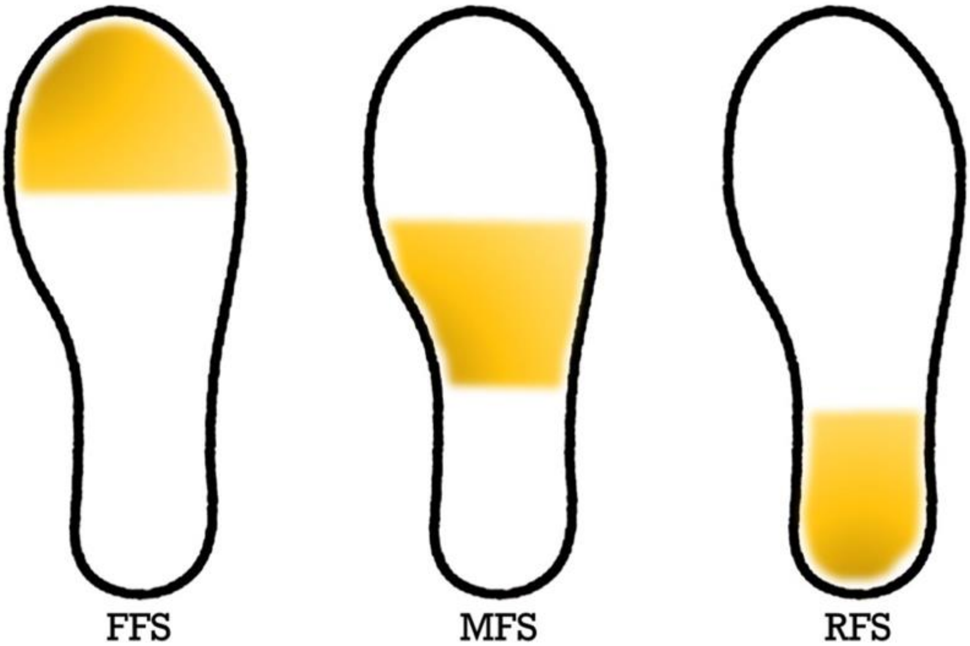


Figure 3.2. Visual feedback for forefoot strike (FFS; left), midfoot strike (MFS; middle) and rearfoot strike (RFS; right).

3.2.5 Data processing and analysis

Custom-written MATLAB code was used to analyze both kinematic and kinetic data. Marker trajectories of both treadmill and overground assessments were filtered by a fourth-order Butterworth recursive low-pass filter, with the cut-off frequency set at 8 Hz.[122] Time of initial contact was defined as the time the GRF exceeded 10 N.[35] The FSA was calculated as the angle of the foot with respect to the ground in the sagittal plane as described previously. Cadence was calculated as the number of foot-strike per minute using the number and time of initial contacts for treadmill assessment and the time between three initial contacts identified using the foot vertical position algorithm as described in Alvim *et al*.'s study within the middle of the runway.[123] Vertical GRF data from both treadmill and overground assessments was filtered by a fourth-order Butterworth recursive low-pass filter, with the cut-off frequency set at 50 Hz.[122] Loading rates, including VALR and VILR, were obtained by the method described in a previous study.[24] The average and maximum slope from 20 to 80% of the vertical GRF magnitude at the impact peak are presented as VALR and VILR respectively. In the absence of a detectable impact peak, the value at 13% of the stance period was considered as the vertical GRF magnitude of the impact peak.[124] Both VALR and VILR were normalized with body mass and all variables were averaged across all 10 footfalls for each condition.

A successful transition to MFS was defined as an average FSA within the boundaries of MFS in each running condition. The percentage of successful MFS transition was compared descriptively. Statistical analyses were performed using SPSS software (Version 26, SPSS Inc, Chicago, IL, USA). The global level of significance for all statistical calculations was set at 0.05. Repeated measures analysis of variance (ANOVA) with a 2 × 2 design were conducted to analyze the interaction between training (Baseline *vs.* Post-training) and the running condition (Treadmill vs. Overground). Post-hoc paired *t*-tests were conducted when applicable. For each running condition, Cohen's *d* was calculated for each variable to evaluate the effect size between time-points. The coefficients were interpreted as trivial, small, medium and large effects for $d < 0.2$, $0.2 \leq d < 0.5$, $0.5 \leq d < 0.8$ and $d \geq 0.8$ respectively.[73,125]

3.3 Results

Twelve male distance runners (mass = 68.1 ± 9.8 kg; height = 1.8 ± 0.1 m; weekly mileage = 31.7 ± 17.3 km; running experience = 5.4 ± 3.7 years) completed this study. All participants completed the gait retraining with no adverse effect reported. The speed used for assessment and training was 3.0 ± 0.3 m/s. At baseline, runners ran with RFS for 99.82 ± 0.63% and 99.24 ± 2.62% of the footfalls during treadmill and overground running respectively.

Six out of 12 participants were able to reduce their FSA to below 8°, indicating a successful adoption of a MFS running pattern during treadmill running. However, two participants in this sub-group were not able to maintain a MFS pattern during overground running. The percentages of successful MFS transition were 50% and 33% for treadmill and overground respectively. The effect of gait retraining on FSA did not interact with running conditions ($\eta p^2 = 0.008$, $p = 0.774$). Simple main effects analysis showed that training has a statistically significant effect on FSA ($p < 0.001$), but not the running condition ($p = 0.708$). While FSA was reduced in both running conditions after the training (*ps* = 0.001), a larger reduction of FSA was observed on the treadmill (Cohen's $d = 2.22$) than in overground running (Cohen's $d = 1.69$) (Table 3.1).

Significant interaction was found for cadence between training and running conditions ($\eta p^2 = 0.489$, $p = 0.008$). Post-hoc analyses revealed that runners ran with an average of 8.26 more steps per minute on the treadmill after the training (Table 3.1). However, the change in cadence was not observed in overground running ($p = 0.293$).

Regarding loading rates, no significant interaction was found in VALR ($\eta p^2 = 0.111$, p = 0.266). Simple main effects analysis showed that the running condition has a statistically significant effect on VALR (p = 0.032), but not training (p = 0.130). However, there was a significant interaction between training and running conditions on VILR ($\eta p^2 = 0.319$, p = 0.044). Post-hoc analyses revealed a large and significant reduction in VILR during treadmill running (Table 3.1), but not in overground running.

Condition		**Baseline**	**Post-training**	***P*-value**	**Cohen's *d***
Treadmill	FSA (°)	17.4 ± 4.0	7.9 ± 4.6	0.001*	2.22
	Cadence (steps/min)	177.37 ± 7.72	185.64 ±11.06	0.009*	0.87
	VALR (BW/s)	93.94 ± 31.97	78.25 ± 22.76	0.102	0.72
	VILR (BW/s)	112.06 ± 31.08	91.03 ± 23.82	0.046*	0.76
Overground	FSA (°)	20.2 ± 5.1	10.2 ± 6.6	0.001*	1.69
	Cadence (steps/min)	179.60 ± 8.92	178.13 ± 9.35	0.293	0.16
	VALR(BW/s)	80.69 ± 29.49	72.78 ± 24.58	0.280	0.26
	VILR (BW/s)	90.61 ± 29.07	82.75 ± 23.31	0.305	0.27

FSA, foot-strike angle; VALR, vertical average loading rate; VILR, vertical instantaneous loading rate; BW, body weight.
* significant difference ($P < 0.05$) between baseline and post-training assessments.

Table 3.1. Mean ± standard deviation and comparison of variables of interest at baseline and post-training assessments.

3.4 Discussion

The aim of this study was to examine the carry-over effect of a treadmill-based gait retraining to overground running. All participants completed a gait retraining program designed to modify their foot-strike pattern from RFS to MFS. Biomechanical parameters, including foot-strike pattern, cadence, VALR and VILR were assessed before and after the training on a treadmill and along an overground runway. When assessed on a treadmill, runners ran with a reduced FSA and VILR, and increased cadence after the training. However, only the reduction in FSA was observed in overground running. This finding was in partial agreement with our hypothesis, with a larger effect size in FSA for treadmill than overground runing. None of the changes in secondary outcomes was successfully carried-over to overground running, suggesting a limited transfer.

Runners have demonstrated their ability to adjust their foot-strike pattern. The transition from RFS to MFS or FFS can be achieved simply by running barefoot[126] or with verbal instructions such as "strike with the forefoot".[127,128] In this study, we aim to induce sustainable changes to the foot-strike pattern of habitual RFS runners, and potentially transferable to untrained conditions. Gait retraining with repeated training sessions has demonstrated positive results in sustainable biomechanical modifications, with changes in foot-strike pattern maintained for up to six months[38] and a reduction in impact loading for up to a year[30] after the training.

In this study, the average reduction of FSA was 9.5 º on the treadmill. Compared to Futrell *et al.*'s study where they reported a change of 17.6º,[38] our training protocol resulted in a smaller change in FSA. Furthermore, the average MFS accuracy was found to be 56.2% on the treadmill, comparably lower than 93 – 100% as previously reported by Cheung *et al.*[129] This discrepancy was likely due to the difference in the training target, Cheung *et al.*[129] and Futrell *et al.*[38] instructed runners to avoid RFS or to land with a FFS, while we specifically instructed runners to avoid over-correction into a FFS. It has been noted that the transition to FFS can lead to pain in the calves, Achilles tendon and the foot,[38] which could be associated with the increased muscle activation in the gastrocnemius medialis[130] and the increased plantar load within the forefoot region when running with a FFS.[131] When compared to our published study which modified runners to MFS, the reduction in FSA and the percentage of successful MFS transition within the trained condition (i.e., treadmill) were comparable.[31]

The reduction in FSA was also observed in overground running, indicating a successful carry-over. From a neuromuscular control perspective, the change in foot-strike pattern is achieved by modulating the time, duration and magnitude of muscle activity.[132] The ankle plantar-flexors, including the medial and lateral gastrocnemius, were consistently found to be activated earlier, longer and stronger in FFS than RFS running.[130,133,134] Similar muscle activity patterns within the pre-stance phase can be expected in MFS running. The muscle activity between treadmill and overground running was found to be similar, with the difference in magnitude observed mainly in thigh muscles (vastus medialis)[135] or within the stance phase.[99,135] The neuromuscular changes adopted during the MFS training are unlikely affected by the change from treadmill to overground running, which might explain the successful carry-over of foot-strike pattern modification observed in this study.

Among the secondary outcomes, we observed a 4.7% increase in cadence and an 18.8% reduction in VILR when participants were running on the treadmill. However, neither of these biomechanical changes was observed when the participants were running overground. The 2 × 2 repeated measures ANOVAs showed a significant interaction of the training effect and the running condition for both cadence and VILR, further indicating an absence of transfer from the treadmill to overground. Increasing cadence has previously been shown to reduce loading rates.[38,42] The absence of change in cadence when running overground might explain the lack of change in VILR in this study. Additionally, the change in mechanical properties between the treadmill surface and the overground runway could also potentially affect the transfer of changes in running kinetics. Even though the surface of instrumented treadmills is generally stiffer than conventional treadmills, it is likely that the treadmill surface has a lower stiffness when compared to in-ground force plates.[86] Surface stiffness was previously identified as a factor that influences the magnitude of impact force.[136] The difference in stiffness might explain the absence of reduction of VILR in the untrained condition.

To our knowledge, this is the first study to assess the carry-over effect of a treadmill-based foot-strike modification gait retraining to overground running. It has been reported that less than 6% of recreational runners run predominantly on a treadmill,[83] majority of runners prefer running on roads or trails, which are considered overground surfaces. Gait retraining has been proposed as a technique to reduce injury risk among runners,[6] and it would only be meaningful if the changes can be carried-over to conditions that are preferred by the runners. In this study, we found that runners were able to run with a MFS but their loading rate is still high when running overground. Since both foot-strike pattern and loading rate could potentially be related to the risk of RRIs,[4,137,138] a complete transfer might therefore be more desirable from an injury prevention perspective.

There are a few limitations to this study. First, we have only examined the immediate biomechanical effect of the training. Foot-strike pattern modification and reduction in loading rate have previously been found to sustain for at least six months on a treadmill.[38] While it is unclear whether the runners continue to run on a treadmill or overground, biomechanical changes were sustained. It is unknown whether our trained participants running in their natural training environment after the training would further modify their overground running gait. Secondly, we did not follow-up with participants on injury. The nature of RRIs is multifaceted[2] and changing modifiable biomechanical parameters associated with injury does not guarantee a reduction in risk. Thirdly, the overground assessment was conducted along an overground runway with the lab. Runners were required to accelerate and decelerate at the start and end of a relatively short runway, even though we measured speed and our variables of interest within the middle of the section, the runway might still be considered too short to establish a stable running pattern. Furthermore, landing impacts have been found to be higher in sidewalk or grass conditions compared to lab runways.[95] The overground condition used in this study may not fully reflect the running biomechanics in real-world running conditions.

3.5 Conclusion and implications for future studies

In conclusion, upon completion of a treadmill-based gait retraining which promotes MFS landing, a smaller FSA, increased cadence and reduced VILR were observed in habitual RFS runners. These modified gait mechanics were partially carried-over to overground running. The altered foot-strike pattern observed on the treadmill was transferred to overground running, as reflected by a reduction in FSA towards the bounds of MFS. However, the increase in cadence and the reduction in the VILR were only observed when assessed on the treadmill. In view of our findings, a treadmill-based running retraining protocol may not be adequate for the transfer of modified running kinetics to overground running.

Considering the limits of treadmill-based gait retraining, and learning is suggested to be most effective when practice is conducted in a natural environment,[9,139] gait retraining conducted on an overground surface may therefore be favored and merit further investigations.

CHAPTER 4

CARRY-OVER EFFECTS OF LEVEL TRACK-BASED GAIT RETRAINING ON FOOT-STRIKE PATTERN, CADENCE AND VERTICAL LOADING RATES UNDER SLOPED RUNNING CONDITIONS

4.1 Introduction

Some runners prefer running outdoors. Trail running provides a unique rewarding outdoor experience that gives runners an opportunity to escape the urban environment while exercising.[140] Compared to treadmill or track running, trail running routes are often longer in distance and have more variations in terrain and elevation.[69,141] There are specific mechanical challenges in their training, such as terrain and grade changes, and counteractive strategies are therefore needed to mitigate injury risk among trail runners.[142]

Runners experience greater impact force when running downhill, as compared to level running.[107,110] Repeated high impact loading experienced by trail runners may contribute to bone failure and stress fracture.[142] Prospective and retrospective studies have suggested that runners with RRIs exhibit higher vertical impact loading than their healthy counterparts.[4,22,143] Vertical impact loading, often represented by VALR and VILR, is affected by cadence and foot-strike pattern.[57,144,145] Huang *et al.* found that habitual RFS runners can reduce their VALR by 42% by adopting a FFS and 7% by increasing their cadence by 10%.[57] Runners demonstrate increased cadence and progressive adoption of RFS from MFS/FFS during downhill running,[105] which could contribute to changes in loading rates. Considering the association between the foot-strike pattern and impact loading, modifying the foot-strike pattern through gait retraining has been proposed to reduce impact loading during running and subsequently mitigate the risk of RRIs. There are studies that found foot-strike pattern modification from

RFS to non-RFS effective in lowering VALR and VILR.[129,146] And yet, a more recent study suggested that the effect of lowering impact loading through adopting MFS might not be universal.[31] The effect of gait retraining programs on foot-strike pattern modification and loading rates, therefore, merits further investigation.

Besides, our understanding of the carry-over effect of foot-strike pattern modification to sloped running conditions is limited. Most gait retraining studies that target foot-strike pattern modification only examined the training effects on a level surface.[31,129] It is possible that natural biomechanical adaptations to sloped running conditions could interfere with the training effect.

The aim of this study was to examine the carry-over effect of track-based gait retraining to sloped running conditions. Specifically, this study evaluates the changes in FSA, cadence, VALR and VILR on various slope conditions, including uphill (+10%), level (0%) and downhill (-10%) running. We hypothesized that after the gait retraining, runners would demonstrate a foot-strike pattern transition from RFS to non-RFS, and increased cadence and a reduction in VALR and VILR in all running conditions. In addition, we hypothesize a larger reduction in FSA, VALR and VILR during level running than in sloped conditions.

4.2 Methods

4.2.1 Participants

The sample size required was estimated using G*POWER 3.1 (Universitat Kiel, Germany). Based on alpha at 0.05, power at 0.8 and an effect size of 0.76 on VALR in a previous in-field gait retraining study using audio feedback,[147] a sample of 16 runners would be sufficient to power the present study. Recreational trail runners who were actively involved in running for more than 10 km per week and habitually adopted RFS were recruited from running clubs in Hong Kong. The exclusion criteria were set as follows: 1) any active lower-extremity injuries within the previous 6 months or other musculoskeletal or neurological conditions that might affect natural running gait and 2) actively participating in gait modification training. To avoid a floor effect, all potential participants underwent an initial screening, runners with RFS for over 90% of the footfalls and VILR above 70 body weight (BW) per second were included. A VILR higher than 70 BW/s was considered higher than the mean value plus one standard deviation (SD) in a healthy group of runners,[4] adjusted for the running speed in the current study.[148] All eligible participants were given a detailed explanation of the experiment prior to signing an informed consent approved by the departmental research committee, Department of Rehabilitation Sciences of The Hong Kong Polytechnic University (HSEARS20190928001).

4.2.2 Initial screening

The initial screening and gait assessments were conducted in the Gait and Motion Analysis Lab at the Hong Kong Polytechnic University. A self-reported training speed used for a typical 30-minute training session was recorded for each participant. This speed was used for the level condition during assessments and training sessions. Two reflective markers were affixed onto the heel and the second metatarsal head of the right foot, according to the model established in a previous study.[55] Three-dimensional marker positions were recorded using an 8-camera motion capture system (Vicon, Oxford, UK) with the participant standing still on a flat, level surface. The participants were instructed to run on an instrumented treadmill (force-sensing tandem treadmill, AMTI, Watertown, MA, USA) at 0% incline in the recorded speed while marker trajectories and GRF will be recorded at 200 Hz and 1,000 Hz respectively for one minute using the motion capture system. The foot-strike pattern was categorized based on the FSA ranges reported in

Altman and Davis's study.[55] VILR was obtained by the method described in Crowell *et al.*'s study.[24] Detailed computation of VILR, FSA and foot-strike pattern classification are described in section 4.2.6. Runners with RFS for over 90% of the footfalls and an average VILR of over 70 BW/s were considered eligible for the study.

4.2.3 Gait assessment

All eligible participants were evaluated in a baseline assessment session, wearing their usual running shoes. The same pair of shoes was used for each participant throughout the entire experiment. To determine the speed for the uphill and downhill conditions, the treadmill speed was increased from 50% of their level running speed with increments of 0.2 km/hour at 10-second intervals until the participant achieved a self-perceived comfortable speed for that condition. Sixteen reflective markers were affixed following the marker placement for Plug-in Gait lower-body model,[149] the two markers affixed at the same positions as the initial screening were used for this study. Each participant completed three five-minute running trials on the instrumented treadmill in a randomized order: level (0%), uphill (+10%) and downhill (-10%). Marker positions for standing with feet flat on the treadmill surface were also recorded for each slope condition. For each trial, markers trajectories and GRF were recorded at 200 Hz and 1,000 Hz respectively during the last minute. A 5-minute rest period between trials was introduced to avoid fatigue.[14]

A post-training assessment was conducted one week after the last training session.[24] The procedure was identical to the baseline assessment, with the sequence of level, uphill and downhill conditions randomized.

4.2.4 Foot-strike pattern sensing insoles

A Brannock device (Brannock, Liverpool, NY, USA) was used to measure the foot size of each participant. The sizing was used for the customization of a pair of 3D-printed sensing insoles made from thermoplastic elastomer. The sensing insoles have been validated, demonstrating up to 84% accuracy on foot-strike pattern identification.[70] Each foot-strike detecting insole was installed with two force sensors (FSR 402 Short, Interlink Electronics, Camarillo CA, USA) located at the heel and the second metatarsal. The sensors were connected to a mini-circuit board which communicates with a smartphone app wirelessly through Bluetooth. Foot-strike patterns were categorized based on the onset time difference normalized by foot length as follows: FFS: onset time difference > 28.2 ms or no heel sensor onset is detected; MFS: onset time difference between 28.2 ms and -46.9 ms; RFS: onset time difference < -46.9 ms; where a positive onset time difference indicated an earlier triggering of the sensor at the second metatarsal than the sensor at the heel.[70]

4.2.5 Gait retraining

Gait retraining sessions were conducted at the Kowloon Bay Sports Ground. The innermost three lanes of the international standard 400-m running track were used for training. At the first session, participants were informed of the three foot-strike patterns (i.e., RFS, MFS and FFS) and were given a pair of sensing insoles and a smartphone with a pre-installed app that generates audio feedback based on the type of foot-strike (Figure 4.1). Researchers demonstrated the associated audio feedback – a ‘beep’ in a high, middle and low pitch when RFS, MFS and FFS were being detected respectively.

Participants wore the same pair of insoles during all eight sessions of their training, with only the right side turned on for real-time feedback. The participants were instructed to modify their foot-strike pattern and avoid a RFS pattern. Audio cues were provided through headphones at every right foot-strike. Laps were timed by researchers and the participants were told to speed up or slow down when their running speed exceeded ±5% of their determined speed.

The training program involved eight sessions over two weeks, each last between 15 and 30 minutes. Audio feedback was provided continuously for the first four sessions. In the last four sessions, the feedback will be progressively removed for two to 28 minutes to enhance motor learning and retention. This protocol was adopted from previous gait retraining studies.[14,31,35] None of the participants received any other types of feedback on foot-strike patterns outside of the training.

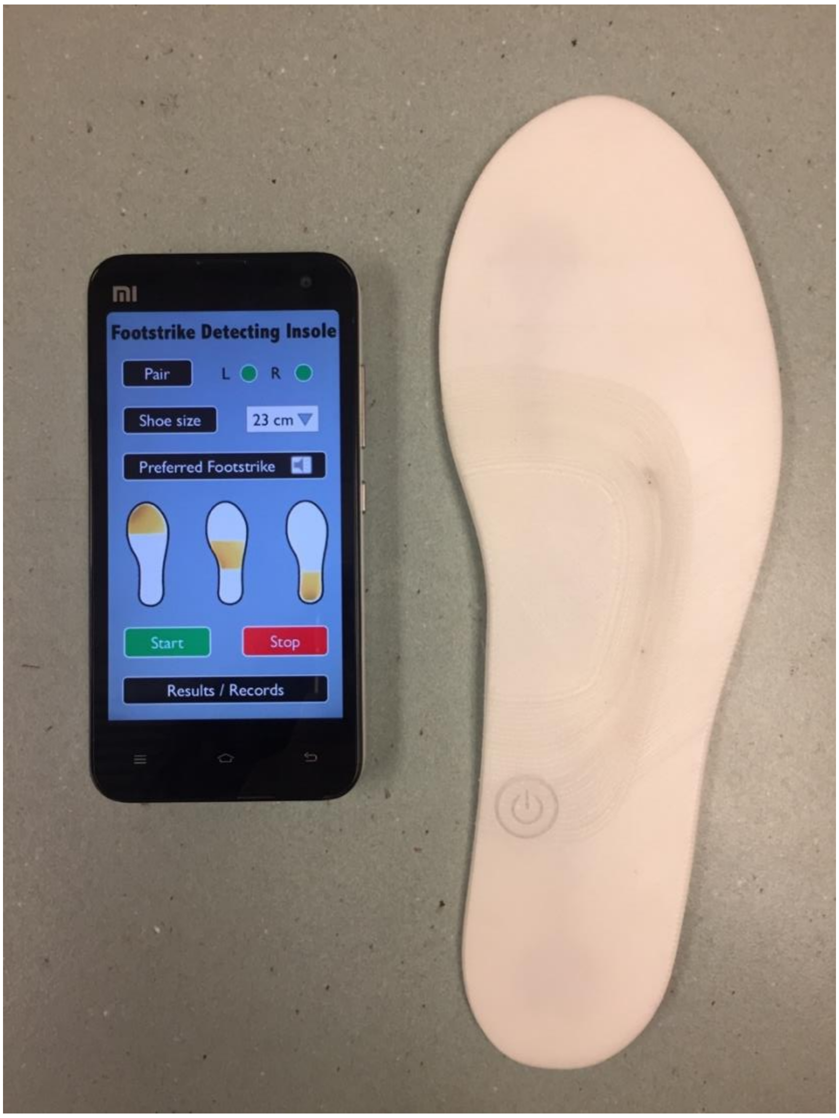


Figure 4.1. Feedback program shown on smartphone and a right foot-strike pattern detection insole.

4.2.6 Data processing and analysis

Custom-written MATLAB scripts were used to analyze both kinematic and kinetic data. Marker trajectories were filtered by a fourth-order Butterworth recursive low-pass filter, with the cut-off frequency set at 8 Hz.[122] Time of foot-strike was defined as the time the GRF exceeded 10 N.[35] For each right foot-strike, the sagittal angle formed by an imaginary line joining the heel and the second metatarsal head markers and the running surface was computed. FSA was calculated by subtracting the sagittal angle during standing at each condition (level, uphill and downhill). Cadence was calculated as the number of foot-strikes detected during the one-minute trial. Vertical GRF data were filtered by a fourth-order Butterworth recursive low-pass filter, with the cut-off frequency set at 50 Hz.[122] Loading rates, including VALR and VILR, were obtained by the method described in a previous study.[24] The average and maximum slope from 20 to 80% of the vertical GRF magnitude at the impact peak are presented as VALR and VILR respectively. In the absence of a detectable impact peak, the value at 13% of the stance period was considered as the vertical GRF magnitude of the impact peak.[124] Both VALR and VILR were normalized with body mass and all variables, except cadence, were averaged across all foot-strikes for each condition.

A successful transition to MFS was defined as an average FSA within the boundaries of MFS in each running condition. The percentage of successful MFS transition was compared descriptively. Statistical analyses were performed using SPSS software (Version 22, SPSS Inc, Chicago, IL, USA). The global level of significance for all statistical calculations was set at 0.05. To investigate the effect of the in-field gait retraining, 2 × 3 repeated measures ANOVAs were conducted to analyze the interaction between gait retraining (Baseline vs. Post-training) and slope conditions (level, uphill and downhill). Post-hoc analyses were conducted when applicable. For each running condition, Cohen's *d* was calculated for each variable to evaluate the effect size between time-points. The coefficients were interpreted as trivial, small, medium and large effects for $d < 0.2$, $0.2 \leq d < 0.5$, $0.5 \leq d < 0.8$ and $d \geq 0.8$ respectively.[73,125]

4.3 Results

Sixteen habitual RFS runners (13 males, 3 females; mass = 63.9 ± 9.2 kg; height = 1.7 ± 0.1 m; weekly mileage = 42.3 ± 14.2 km; running experience = 5.5 ± 3.9 years) completed the study. All participants completed the in-field gait retraining with no adverse effect reported. The assessment speeds are presented in Table 4.1.

Twelve out of 16 habitual RFS participants were able to reduce their FSA to below 8° on a level surface after training, indicating a transition from RFS to non-RFS for 75% of the participants. Nine (56.2%) and three (18.8%) of the participants were running with a MFS and FFS after training, respectively. The effect of training on FSA interacted with the slope conditions ($\eta p^2 = 0.192$, $p = 0.041$) (Table 4.2). Each slope condition was analyzed separately to examine the effect of training. Participants showed an average of 10.5° of reduction in FSA during level running. Pairwise comparison showed a large reduction in FSA ($p < 0.001$, Cohen's $d = 2.06$).

At baseline, ten participants were running with a MFS during uphill running (Figure 4.2). After the training, 12 and three runners were running with a MFS and a FFS respectively. The percentage of successful MFS transition was not calculated for uphill running since participants demonstrated foot-strike pattern transition at baseline. Still, participants demonstrated an average of 5.1° of reduction in FSA ($p < 0.001$, Cohen's $d = 1.51$).

	Level	Uphill	Downhill
Speed (m/s)	2.8 ± 0.4	2.2 ± 0.4	2.5 ± 0.5
FSA (º)	14.4 ± 3.2	7.5 ± 3.1	7.7 ± 10.3
Foot-strike pattern (%)			
RFS	97.13	37.28	52.04
MFS	2.87	62.63	25.31
FFS	0	0.08	22.65
Cadence (steps/min)	172.11 ± 11.07	172.11 ±10.25	165.84 ± 11.70
VALR (BW/s)	102.27 ± 28.24	58.65 ± 16.24	90.84 ± 34.64
VILR (BW/s)	115.33 ± 28.08	67.32 ± 16.80	104.30 ± 36.35

FSA, foot-strike angle; RFS, rearfoot strike; MFS, midfoot strike; FFS, forefoot strike; VALR, vertical average loading rate; VILR, vertical instantaneous loading rate; BW, body weight.

Table 4.1. Mean ± standard deviation of variables of interest at baseline assessment for each slope condition.

		F	*P*-value
Training * Slope	FSA (º)	3.56	0.041*
	Cadence (steps/min)	0.20	0.820
	VALR(BW/s)	1.87	0.172
	VILR (BW/s)	1.89	0.192
Training	FSA (º)	30.70	<0.001*
	Cadence (steps/min)	8.11	0.012*
	VALR(BW/s)	9.91	0.007*
	VILR (BW/s)	10.58	0.005*
Slope	FSA (º)	7.89	0.002*
	Cadence (steps/min)	17.48	<0.001*
	VALR(BW/s)	17.98	<0.001*
	VILR (BW/s)	18.98	<0.001*

FSA, foot-strike angle; VALR, vertical average loading rate; VILR, vertical instantaneous loading rate; BW, body weight.
* significant interaction or main effect ($P < 0.05$)

Table 4.2. Analysis of variance results for the effect of training and on variables of interest.

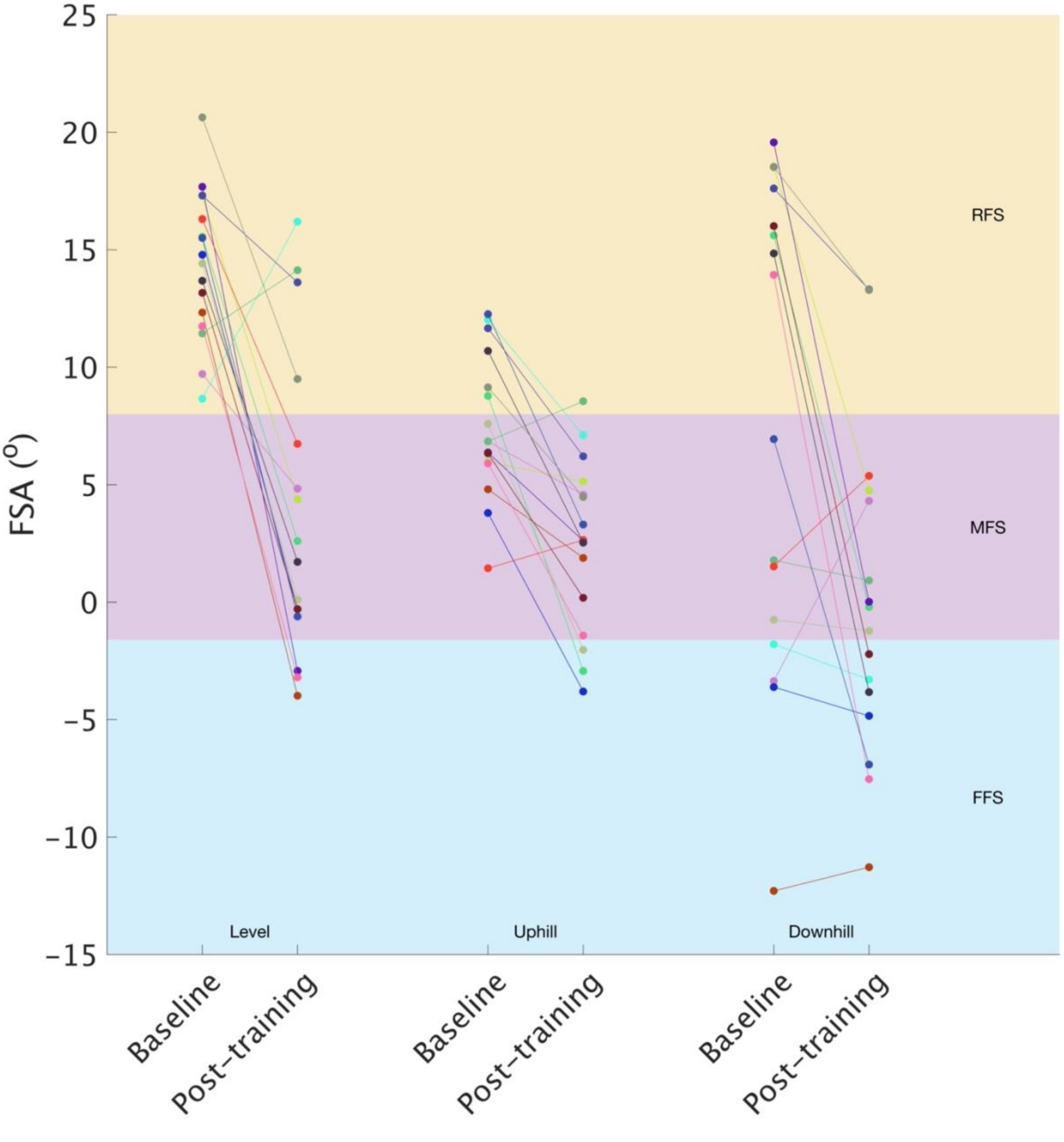


Figure 4.2. Foot-strike angle (FSA) during baseline and post-training assessments on level, uphill and downhill conditions.

Each colour represents one participant. The shaded areas indicate the region for rearfoot strike (RFS), midfoot strike (MFS) and forefoot strike (FFS) as labeled.

Similar to uphill running, participants demonstrated changes in the foot-strike pattern at baseline. Four and 4 participants were running with a MFS and FFS during downhill running (Figure 4.2). After the training, 14 runners were running with a non-RFS, with seven runners running with a MFS. On average, the FSA for downhill running was reduced by 7.7º, with the pairwise comparison showing a large and significant reduction (p = 0.006, Cohen's d = 0.88).

Results of the repeated measures ANOVAs indicated that there was no interaction between training and slope conditions on cadence, VALR and VILR (Table 4.2), but there were significant main effects for both training (ps < 0.012) and slope (ps < 0.002).

Cadence was increased during uphill (p = 0.010, Baseline: 172.11 ± 10.25, Post-training: 177.11 ± 9.25 steps per minute) and downhill running (p = 0.030, Baseline: 165.84 ± 11.70, Post-training: 169.49 ± 10.03 steps per minute), but remained unchanged at level running (p = 0.106, Baseline: 172.11 ± 11.07, Post-training: 176.41 ± 9.93 steps per minute). Uphill running showed a medium change (Cohen's d = 0.51), while downhill running showed a small change (Cohen's d = 0.34). VALR was reduced in both level (p = 0.003, Cohen's d = 1.02, Baseline: 102.27 ± 28.24, Post-training: 73.47 ± 28.43 BW/s) and uphill running (p = 0.008, Cohen's d = 0.80, Baseline: 58.65 ± 16.24, Post-training: 45.35 ± 16.95 BW/s). Similarly, VILR was reduced in both level (p = 0.002, Cohen's d = 1.09, Baseline: 115.33 ± 28.08, Post-training: 82.23 ± 32.46 BW/s) and uphill running (p = 0.004, Cohen's d = 0.93, Baseline: 67.32 ± 16.80, Post-training: 50.64 ± 19.13 BW/s). The extent of loading rate reduction was smaller in uphill than in level running. However, VALR (p = 0.071, Baseline: 90.84 ± 34.64, Post-training: 70.71 ± 32.31 BW/s) and VILR (p = 0.097, Baseline: 104.39 ± 36.35, Post-training: 84.11 ± 39.14) appeared to be unchanged at downhill running.

4.4 Discussion

The aim of this study was to examine the carry-over effect of track-based gait retraining to sloped running conditions. All participants completed eight sessions of training on a level surface, with real-time audio feedback provided through a foot-strike detection system. Biomechanical parameters, including foot-strike pattern, cadence, VALR and VILR were assessed before and after the training on level, uphill and downhill running conditions. When assessed on a level running surface, runners exhibited reduced FSA and reduced loading rates. Similar changes have also observed when runners were running uphill, however, a smaller reduction in FSA and no changes in loading rates were observed in downhill running. The findings of this study suggested carry-over to uphill running, but an incomplete transfer of training effect to downhill running.

When running on a level surface, this study achieved a higher success rate (75%) in transitioning runners from RFS to non-RFS, compared to 40 – 50% reported in our published study[31] and Study 1 (Chapter 3) where runners were trained on a treadmill. In this study, we instructed runners to avoid RFS. Runners could have attempted both MFS and FFS during the training and settled on a foot-strike pattern that felt natural to them. Among those that successfully transitioned, nine of them adopted MFS and three adopted FFS. The successful transition was further evidenced by a reduction in FSA, with the average post-training FSA of $3.9 \pm 6.5^{\circ}$.

Compared to level running, we observed a change in foot-strike pattern when runners were running uphill at baseline. Gottschall and Kram observed a transition from RFS to MFS in three out of ten runners when running at an incline (+10.5%) comparable to that used in this study.[107] Similarly, Lussiana *et al.* reported reduced FSA, indicating a MFS/FFS transition, at +8%.[108] This habitual foot-strike pattern transition has also been observed at more extreme incline conditions (+14 – 48%).[69,107] In the current study, ten runners adopted a MFS when running uphill at baseline. After the training, all but one runner was exhibiting a non-RFS during uphill running. The large and significant reduction in FSA further demonstrated successful carry-over of training effect from level training to uphill running. As for downhill running, the habitual foot-strike pattern transition was inconsistent. At baseline, half of the runners transitioned into MFS/FFS, while the other half maintained RFS (Figure 4.2). Most biomechanical studies have reported unaltered foot-strike pattern when running downhill.[69,107] The habitual transition observed in our study might be explained by the difference in the experience level of our participants. It has previously been reported that more experienced trail runners adopt MFS while mid-level runners adopt a RFS.[109] While a reduction in FSA was observed in downhill running after the training, the effect size was smaller than that in level and uphill running.

Regarding cadence, no change in cadence was observed in level running. Previous gait retraining that modified RFS to FFS on a treadmill reported a 6.1% increase in cadence.[38] The results of our study suggested that training overground may change the foot-strike pattern of runners without imposing changes in cadence. While significant changes were observed in both uphill (+3%) and downhill (+2%) running, such changes were unlikely to be interpreted as clinically important.[150] Gait retraining studies that target reduction in impact loading set the training target to be 7.5 to 10% above baseline,[38,42,150,151] higher than the changes reported in our study.

We hypothesized a reduction in VALR and VILR following a transition from RFS to MFS/FFS based on previous studies.[57,144] Similar to Study 1, this study aimed to induce changes within the trained condition (i.e., level running) and examine the changes within untrained conditions (i.e., uphill and downhill conditions). Study 1 successfully reduced VILR among runners with high VILR (112.1 ± 31.1 BW/s) while our published study reported no changes among runners with a lower baseline VILR (89.5 ± 24.5 BW/s). In the current study, we screened and included runners with high VILR (> 70 BW/s) to avoid a potential floor effect. When accessing the effect of training on loading rates, significant and large reductions were observed in both level and uphill running, but not during downhill running. To exclude the effect of habitual foot-strike pattern change on the training effect, we have isolated the eight runners who maintained a RFS at baseline downhill running and conducted paired *t*-tests to examine the training effect. While we found a significant reduction in FSA ($p < 0.001$) among this sub-group, no significant changes were found in VALR and VILR (*ps* = 0.052 to 0.067). Downhill running has been associated with other kinematic differences,[105] including more knee extension at initial contact and greater hip range of motion, which could also affect shock attenuation and loading rates. The transition of foot-strike pattern alone might not be sufficient to induce a reduction in VALR and VILR during downhill running.

Modifying foot-strike patterns has its own limitations. Based on our results, uphill running showed similar changes in foot-strike pattern and loading rates to level running after the track-based gait retraining, but changes in kinetics were not transferred to downhill running. Natural adaptation of foot-strike patterns along slopes could interact with the training effect, resulting in ineffective transfer. Based on the assumption that motor learning is most effective when conducted within a natural environment,[9,139] gait retraining along slopes might provide better results for runners who encounter slopes during their own training. However, training that targets foot-strike pattern transition may not be suitable. We found that half of the runners naturally changed their foot-strike to non-RFS during downhill running. These runners would not be cued to change their foot-strike pattern if the training protocol of the current study was repeated along downhill slopes. An alternative would be to reduce PTA, a surrogate of loading rates.[50] Using feedback on PTA as training targets has demonstrated successful transfer from treadmill-based training to overground running.[40] Zhang *et al.* also observed a similar transfer effect from level to sloped running after a treadmill-based training.[44] Further investigations are required to examine the effect of reducing PTA through real-time feedback gait retraining along sloped conditions.

There are a few limitations to the current study. The assessments before and after the training were conducted on a treadmill. Participants were trained overground, there is a possibility that the effects of training are larger within the trained condition. It is unknown if the training effect was transferred to overground sloped running. Future studies should consider an overground or in-field evaluation to fully comprehend the carry-over effect. Secondly, the screening criteria could have introduced bias. Habitual RFS runners with high loading rates might be more motivated to adjust their foot-strike pattern. This could have increased the effect of training, without a control group, a causation relationship between the training and the biomechanical changes cannot be established. Nonetheless, the findings of our study still support the training in modifying foot-strike pattern in RFS runners with high loading rates, whose risk of developing RRIs are relatively high.[22] Lastly, similar to Study 1, the long-term clinical effect of foot-strike pattern modification was not examined. Large-scale longitudinal research with a control group is required to further examine the effect of foot-strike modification on runners.

4.5 Conclusion and implications for future studies

In conclusion, upon completion of an overground running retraining along a level surface which promotes non-RFS landing, participants were running with a smaller FSA and reduced VALR and VILR. The changes in foot-strike patterns were also observed in uphill and downhill running. However, the reductions in VALR and VILR were observed only in uphill running, but not downhill running. The incomplete carry-over suggested that gait retraining conducted along slopes might be necessary to reduce loading rates while running downhill. Considering the natural changes to foot-strike pattern, and its interaction with the training effect, using PTA as feedback might be more suitable for training along slopes.

CHAPTER 5

SPEED AND CADENCE ADAPTATIONS DURING OVERGROUND SLOPED RUNNING UNDER REAL-WORLD CONDITIONS

5.1 Introduction

Runners demonstrate biomechanical adaptations when running on slopes. These adaptations include changes in foot-strike pattern, impact forces and cadence,[105] which are commonly measured in gait analyses.[14,31,42,44] Compared to level running, uphill running has been associated with a higher cadence and a less dorsiflexed ankle at foot-strike,[105,152] while downhill running has been associated with lower cadence and greater tibial acceleration.[105,108,153,154]

Runners' natural adaptation to slopes should be considered when evaluating the effectiveness of interventions, such as gait retraining, on sloped running conditions.[69,155] However, it could be a challenge to measure natural adaptations in the lab. Surface gradient, cadence and running speed are interdependent. To isolate the effect of slope on cadence, most published studies reported changes in running biomechanics in lab-simulated inclined surfaces,[107,113,156] in which the running speed was set and controlled by the researchers. Such experimental design has limited the runners' ability to regulate their running speed. In fact, a case study of a world-class athlete who ran a 45-km trail with 1,627 m of positive elevation, showed a much slower speed (6.8 – 8.2 km/h) during the uphill sections (gradient: +14.6 – 14.9 %) when compared to level sections (speed: 11.1 – 14.2 km/h; gradient: +1.4 – 1.6%).[110] Similarly, Townshend *et al.* found that runners increased their speed during downhill sections of the 9.5-km trail and reduced their speed during uphill sections.[114] Interestingly, the hypothesized changes in cadence, which were present in speed-controlled treadmill-based studies, were not observed. It is important to note that in both in-field studies,[110,114] only competitive runners were included, and the runners were asked to complete a time trial. The pacing strategy could be a result of experience in trail races and runners trying to match their oxygen consumption with their ventilatory threshold while running uphill and improving time on downhill sections.[114]

With the rapid growth in wearable technology, researchers are now presented with the opportunity to obtain quantifiable data from runners in a real-world setting.[60,157,158] Commercially available accelerometer-equipped GPS running watches can provide a series of metrics, including cadence, speed, distance and elevation. This study aimed to examine the relationship between surface gradient, cadence and running speed for recreational runners in real-world settings. More specially, comparisons were made between uphill, level and downhill sections of recorded runs. It was hypothesized that runners' would demonstrate grade-specific adaptations, and the running speed and cadence would be different between uphill, level and downhill running.

5.2 Methods

5.2.1 Pre-screening data from the database

Data were extracted from the We-TRAC Wearable Technology Citizen Science Program database. This database stored activity records within a Level-4 secure server housed at the University of Calgary. Activity records were collected from individuals through a web portal (https://wetrac.ucalgary.ca) where they can upload data from their GPS-enabled smartwatches. The collection of data through this database was approved under the study title "The Wearable Technology Citizen Science Program" by the Conjoint Health Research Ethics Board at the University of Calgary (REB20-0572_REN2). Retrospective activity records, for up to 24 months prior to sign up, and prospective weekly data, that had been collected since sign up, were stored within the database.

Activity records within the whole database were pre-screened. A small subset containing 573 records was extracted from the database and used to set the screening criteria for extracting the relevant activity files for this study. Activity records meeting all the criteria below were extracted for further processing:

The activity was recorded by a Garmin device.

The activity was classified as “running”, “track running”, “street running” or “trail running”.

- The total distance of the run was longer than or equal to 5 km.
- The total elevation gain or loss was over 100 m.
- The average speed was greater than 1.2 m/s.

5.2.2 Data processing and analysis

The activity records were further screened to ensure a significant uphill or downhill segment and a level segment for comparison. Recorded time, elevation, horizontal distance, and cadence in 1 Hz were extracted from each activity record. Elevation and distance data were smoothed with a 10 s moving average.[159] Each run was segmented into 100-m non-overlapping sections. The average gradient (%) of each 100-m section was computed as overall elevation gain (or loss) over the horizontal distance traveled. Speed was calculated using the elevation change and horizontal distance traveled over time for each section. Each section was then classified into 3 conditions based on the average gradient: (i) uphill (+3 to 15%), (ii) level (-2 to +2%) and (iii) downhill (-15 to -3%).[159] Unclassified portions (i.e., inclination does not match uphill, level or downhill conditions) and portions with running speed less than 1.8 m/s were removed.[160] Data from the first 500 m of each training record were considered the warm-up period and were removed from the analysis.[161] To minimize the potential effects of fatigue, only data within the first 10 km of each run were considered.[159] Consecutive 100-m sections with the same condition were combined into segments and only segments of at least 600 m in length were kept for further analysis. Activity records that contain level and uphill/downhill segments suitable for analysis were kept for further analysis. An example of segmentation of a recorded run is presented in Appendix III.

5.2.3 Statistical analysis

All dependent variables were tested against a normal distribution using separate Shapiro-Wilk tests. The average speed and cadence were calculated from the second to sixth 100-m sections of each condition to ensure a steady-state gait.

Paired *t*-tests were used to compare speed and cadence between the uphill/downhill segments and the level segment within the same training record. To ensure an equal contribution of each runner, the run closest to 10 km was selected if multiple runs from the same runner meet the criteria. Statistical analyses were performed using SPSS software (Version 26, SPSS Inc, Chicago, IL, USA). The global level of significance for all statistical calculations was set at 0.05. Cohen's d was calculated for each variable to evaluate the effect size between time-points. The coefficients were interpreted as trivial, small, medium and large effects for $d < 0.2$, $0.2 \leq d < 0.5$, $0.5 \leq d < 0.8$ and $d \geq 0.8$ respectively.[73,125]

A previous study suggested that up to five runs were needed to define a stable running pattern across inclination conditions. A data subset was created with runners who have at least five runs for each condition. The speed and cadence for each condition were averaged across the five runs closest to 10 km. Repeated measures ANOVAs were used to determine significant differences among slope conditions (i.e., level, uphill and downhill) for speed and cadence. Post-hoc analyses with a Bonferroni adjustment were conducted when applicable.

To determine the intra-subject relationship between gradient and the change in speed or cadence, Pearson correlations coefficients (*r*) were calculated for runners with more than 20 available records. The coefficients were interpreted as indicated in Table 5.1.

Strength	Positive	Negative
Negligible	0 to 0.09	-0.09 to 0
Weak	0.1 to 0.39	-0.39 to -0.1
Moderate	0.4 to 0.69	-0.69 to -0.4
Strong	0.7 to 0.89	-0.89 to -0.7
Very strong	0.9 to 1	-1 to -0.9

Table 5.1. Interpretation of correlation coefficient.[162]

5.3 Results

Data were accessed on July 19, 2022. A total of 12,317 running records between July 16, 2018 and July 13, 2022 met the screening criteria. After further screening, data from 148 runners (3,001 activity records) were retained for further analysis. Inquiry emails were sent to the runners to obtain their year of birth and gender. We received email replies from 85 (57.4%) of the runners at the time of writing. The average age for runners that replied to our inquiry email was 43.4 ± 13.3 years, with 60% (n = 51) being males. All variables were tested and found to be normally distributed.

Paired *t*-tests were conducted to compare speed and cadence between the uphill/downhill segments and the level segment of the same training record. The number of records, length of run and average gradient were presented in Table 5.2. Compared to the level segment of the same activity record, significantly lower cadence ($p < 0.001$) and reduced speed ($p < 0.001$) were found in the uphill segment. Runners were also found to be running significantly faster in the downhill running segment compared to the level segment ($p = 0.013$), but no significant difference was found in cadence ($p = 0.694$).

A total of 65 runners have at least five runs in each sloped condition. Based on the result of the repeated measures ANOVA, running speed differed among slope conditions ($F = 197.06$, $p < 0.001$). Running speed was significantly higher in level and downhill running (*ps* < 0.001) compared with uphill running (Figure 5.1). Running speed was also higher in downhill than in level running ($p = 0.029$). Similarly, cadence differed among slope conditions ($F = 8.33$, $p < 0.001$). Cadence was similar between level and downhill running ($p = 0.341$) and between uphill and downhill running ($p = 0.082$), but significantly higher in level ($p < 0.001$) compared with uphill running (Figure 5.1).

		Characteristic of runs						Pairwise comparison	
		Length of run (km)	Average gradient (%)		Speed (m/s)			*P*-value	Cohen's *d*
Level vs.	n		Level	Uphill/ Downhill	Level	Uphill	Downhill		
Uphill	134	1.10 ± 0.35	-0.11 ± 0.57	+6.24 ± 1.74	2.96 ± 0.45	2.38 ± 0.61		<0.001*	1.13
Downhill	141	1.15 ± 0.47	-0.06 ± 0.56	-6.35 ± 1.66	3.02 ± 0.50		3.13 ± 0.74	0.013*	0.213
					Cadence (steps/minute)				
Level vs.	n		Level	Uphill/ Downhill	Level	Uphill	Downhill		
Uphill	127	1.08 ± 0.24	-0.12 ± 0.58	+6.24 ± 1.70	169.11 ± 8.76	166.98 ± 9.77		<0.001*	0.30
Downhill	138	1.15 ± 0.54	-0.05 ± 0.56	-6.38 ± 1.69	169.43 ± 9.00		169.25 ± 10.16	0.694	0.04

n, number of training records used for the pairwise comparison.
* significant difference ($P < 0.05$) in level vs. uphill or downhill

Table 5.2. Mean ± standard deviation of variables of interest, length of run and average gradient used for pairwise comparisons.

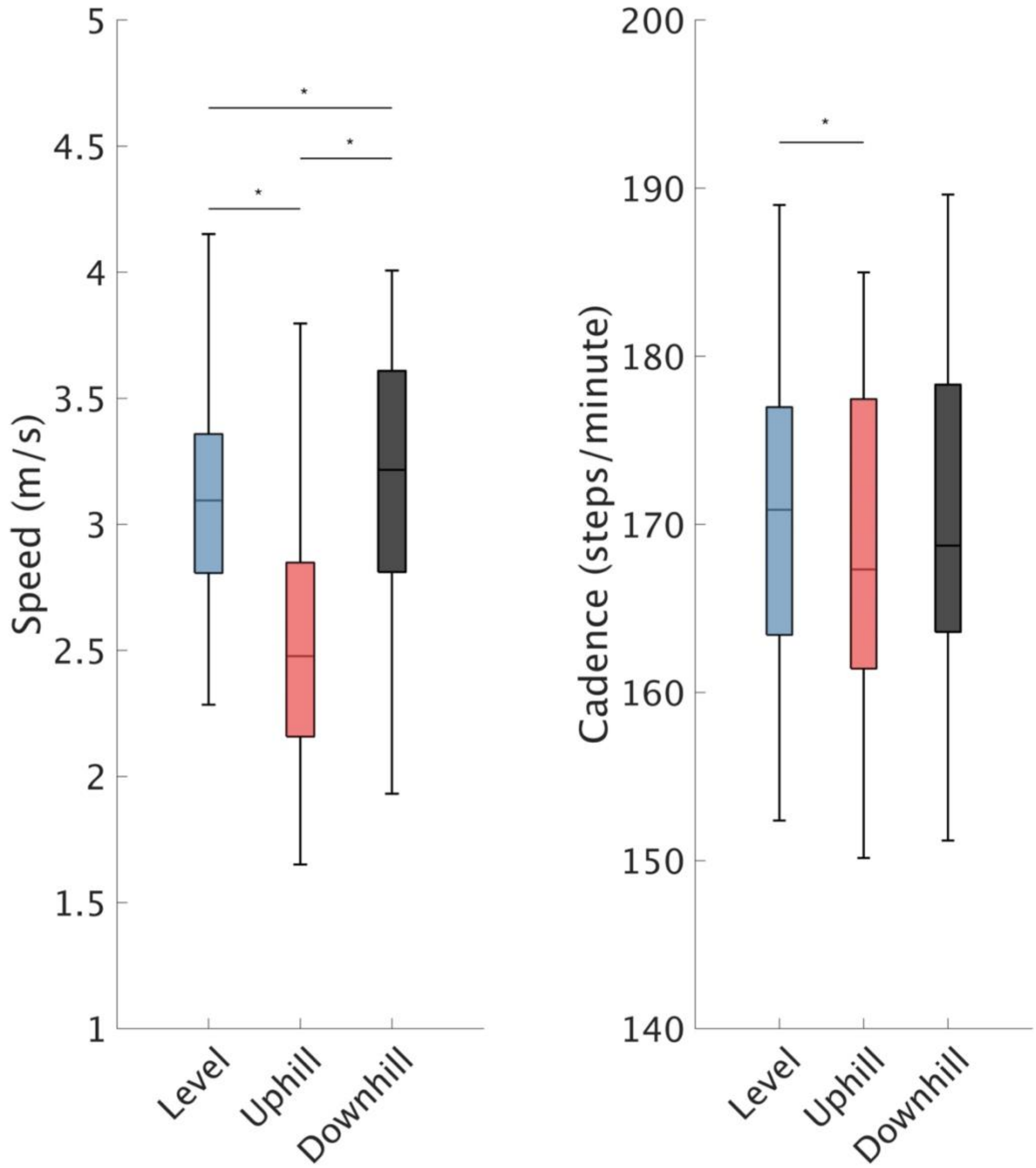


Figure 5.1. Box plots of running speed (left) and cadence (right) across level, uphill and downhill running.

* denotes $p < 0.05$ at post-hoc Bonferroni pairwise comparison

To establish the relationship between gradient and change in speed and cadence (Δ = Uphill/Downhill – Level), Pearson's *r* was calculated for each runner. Due to data availability, different numbers of runners were tested for the intra-subject association for each variable (Δ speed and Δ cadence) and condition (uphill and downhill). Table 5.3 shows the number of runners with at least 20 records available for analysis.

Significant associations were found in a small group of runners, as presented in Table 5.3. We found most associations in uphill running, with 13 out of 28 runners showing a reduced speed as the gradient increased and 7 out of 26 runners showing a negative relationship between gradient and cadence. Pearson's *r* presented in Table 5.3 are the minimum and maximum values between all runners that showed a significant association for that condition. A line of best fit was also presented in Figure 5.2 for the runners with significant associations.

While significant associations were also found in downhill running, the number of runners with significant association was even smaller. Moreover, large between-subject variances were observed. Regarding the change in speed, there were two runners showing significant relationships. While both relationships are considered weak, they were in different directions, indicating that speed increased as the slope got steeper for one runner (r = -0.37) but speed decreased (r = 0.39) for the other. This between-subject variance was also observed in the change in cadence where *r* ranged from -0.59 to 0.64.

	Number of runners with sufficient records	Number of runners with significant correlation ($p < 0.05$)	Pearson's r [Min, Max]
Uphill, Δ speed	28	13	-0.63, -0.19
Downhill, Δ speed	34	2	-0.37, 0.39
Uphill, Δ cadence	26	7	-0.59, -0.26
Downhill, Δ cadence	32	5	-0.59, 0.64

Δ, difference from level.

Table 5.3. Number of runners with sufficient data for intra-subject correlation analysis and with significant association, and their correlation coefficient.

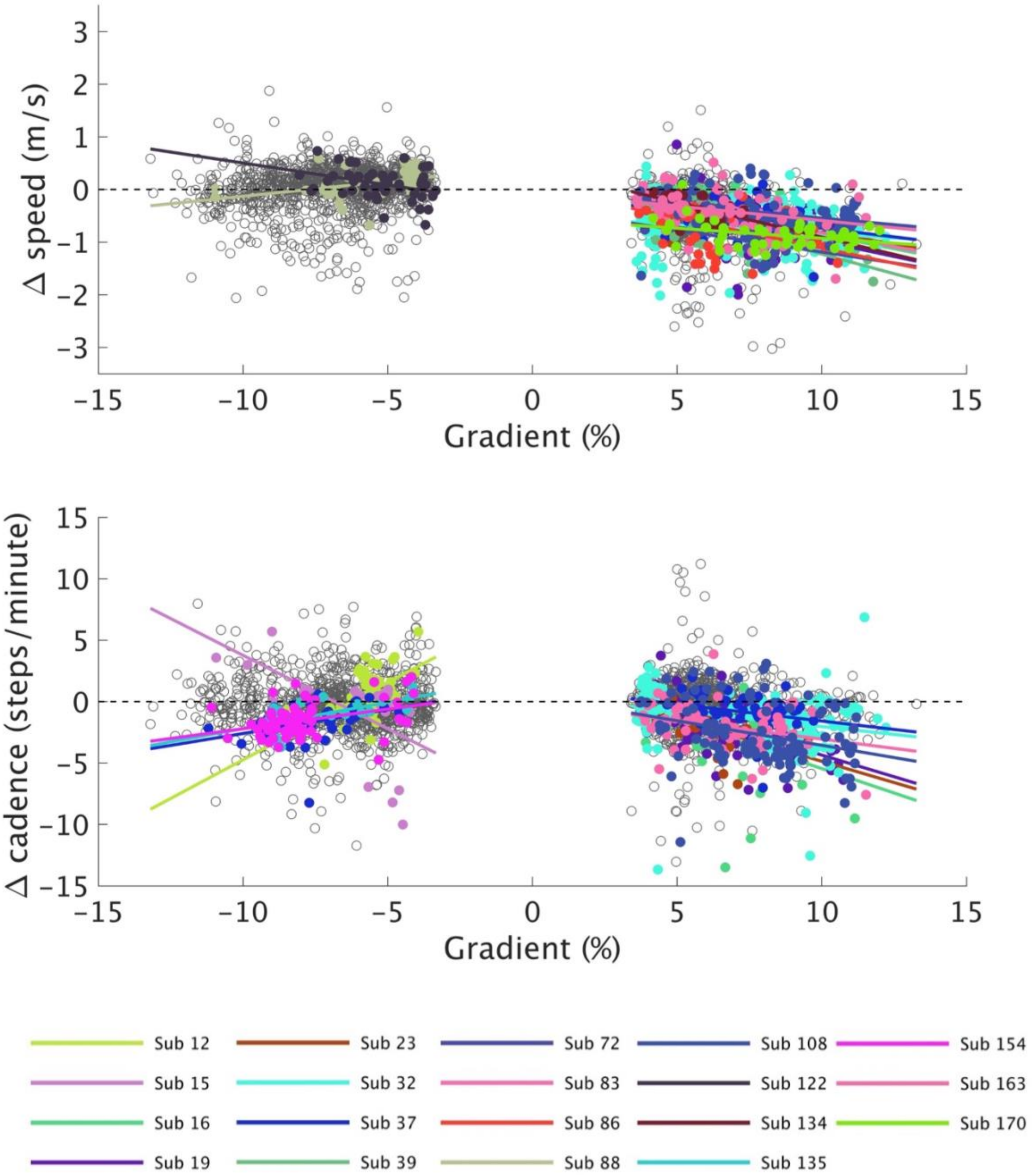


Figure 5.2. Correlation between gradient and change in speed (top) and cadence (bottom). The lines of best fit and data points are color coded for each runner with a significant association. Grey circles represent data points of runners with no significant association between the pair of variables tested. Negative difference (Δ) indicates value is larger during level running.

5.4 Discussion

The aim of this study was to examine the relationship between surface gradient, running speed and cadence for runners in real-world settings. The results of the current study supported grade-specific adaptations among runners. When compared to running on a level surface, reductions in speed and cadence were observed along uphill conditions, and an increase in speed along downhill conditions. Associations between the gradient and change in speed and cadence were observed in a small number of runners, with the relationship more consistent in uphill running than in downhill running.

Grade-specific adaptations have been reported in previous studies conducted within the lab.[105] Generally, uphill running has been associated with a higher cadence than level running. Telhan *et al.* found 1.2% higher cadence at +7% grade,[112] similar to Padulo *et al.* who also found 4.1 – 15.6% increase in cadence among elite runners at +2% and +7% grade.[113] Contradictory to the aforementioned studies, we found that runners were running with a lower cadence during uphill running compared to a level surface, with the average gradient (+6.2%) of the runs comparable to that of Telhan *et al.*'s study. This discrepancy is likely due to the change in speed found in uphill running. In the studies where an increase in cadence was observed, speed was predetermined and strictly controlled through the treadmill. In the current study, we obtained records of runners running overground outside of the lab, speed was determined by the runner. We found runners running 18.9% slower when averaging five runs and 19.6% when comparing the difference within the same run. In fact, this reduction in speed coupled with a reduction in cadence has previously been reported in an in-field study where participants completed a 40-km trail race.[163] Within the first 10 km, runners ran with a slower speed and lower cadence along the uphill section, compared to the level section.

Regarding downhill running, a few studies found comparable cadence at grades between -15.8 and -5.2% when compared to level running when speed was controlled.[107,112,153]

Lussiana *et al.* found lower cadence when runners were running downhill (-5 and -8%) in minimal and traditional footwear.[108] In an overground study where speed was not restricted, Townshend *et al.* found no changes in cadence but an increase in running speed.[114] This current study found a similar trend, no significant difference was observed in cadence and speed was increased when running downhill. We found an average of 2.7% increase in running speed when averaged across five runs and 3.6% when comparing against the level segment of the same run, smaller than that reported. A likely explanation would be the difference in pacing strategy and the goal of the run. Participants completing a time trial in Townshend *et al.*'s study might see downhill running as a potential to improve time, increasing their speed to a greater extent than during a regular training session.[114]

We also aim to determine the relationship between surface gradient and the change in speed and cadence. A few published studies have tested the change in cadence across a number of slope (incline and decline) conditions. For example, Lussiana *et al.* found that cadence was reduced at -5% grade, but a further reduction was observed when the slope was steeper (-8%). For uphill, Padulo *et al.* reported a greater reduction in cadence at +7% than +2% when compared to level running at adjusted speed (i.e., 4.4, 4.0 and 3.2 m/s at 0, +2 and +7%).[156] Giandolini *et al.*'s case study has similar observations, cadence decrease from 180.6 to 174.6. and 163.8 steps per minute at the sections of the trail with +1.2%, + 6.8 and +14.6%.[110] Based on the findings, we hypothesized that the change in speed and cadence is related to the magnitude of the incline or decline. A series of intra-subject correlations suggested that the relationship exists within a small group of runners. Along inclined surfaces, weak to moderate negative (i.e., decrease in cadence/speed as grade increases) relationships were found among 13 out of 28 runners for cadence and seven out of 26 runners for speed. Interestingly, the relationships between grade and change in cadence and speed were inconsistent along a declined surface. Fewer runners showed a significant relationship, only two out of 34 showed a relationship for speed and five out of 32 showed for cadence. The relationship between grade and change in speed was different among the two runners, one showed an increase in speed as the downhill slope gets steeper while the other showed an opposite trend. The relationship between grade and change in cadence during downhill also showed between-subject variance, with moderate relationships in both negative and positive directions. Running uphill imposed a greater metabolic demand, evidenced by higher muscle activation[164] and an increase in anaerobic energy production.[105,165] Runners could be adapting their speed and cadence to keep up with the increase in metabolic demand. Alternatively, running downhill does not impose such a metabolic demand and runners can adapt more freely based on their training goals. Future studies should consider analyzing parameters indicative of physiological demand, such as heart rate, to further understand the intra- and between-subject variance.

To our knowledge, this is the first study that used real-world training data from runners to examine the changes in speed and cadence along slopes. There were dissimilarities between our findings and lab-based studies. Grade-specific adaptation demonstrated in speed-controlled conditions should not be generalized to real-world running. In some in-field or overground studies along slopes,[69,117] speed was a control variable. A constant speed allows researchers to measure the dependent variable, however, it may not fully reflect the running biomechanics when runners are allowed to adjust their speed. This study has demonstrated the potential of utilizing real-world running data to understand runners within their natural environment. However, it is not without limitations. First, data were extracted from a large database; we have limited information about the runner and each running session. Runners' experience which could affect grade-specific adaptations[113] and stride length were not available to us. Also, external factors including the nature of the run (e.g., competition or training), surface type (e.g., grass, concrete or dirt) and weather could also affect speed and cadence. To minimize potential bias, intra-subject comparisons, an average of five runs and within-session comparisons were adopted in the current study. Second, GPS signal was missing in some running records, affecting the calculation of speed and elevation. We have thoroughly inspected the data records and removed files/segments with unreliable GPS signals, however, this process was time-consuming and subject to bias, a more systematic approach should be developed. Lastly, only segments of at least 600 m in length were analyzed This criteria was set to ensure a stable gait pattern and comparable results with a previous in-field running study.[114] The average slope within a 600-m segment was used to categorize slope conditions, and the range is relatively wide (3 – 15%). Changes within the 600-m segment and the differences between subtle and steep slopes might have been neglected in the current study.

5.5 Conclusion and implication for future studies

In conclusion, runners demonstrate grade-specific adaptations within their natural training environment. These adaptations include a reduction in speed and cadence when running uphill, and an increase in speed when running downhill. Associations between the gradient and change in speed and cadence were observed in a small number of runners, with the relationship more consistent within uphill running than in downhill running.

Our findings imply that runners naturally change their running biomechanics during sloped running. Future in-field gait retraining studies that aim to increase cadence should consider these adaptations, especially for training along slopes. Even though the change in cadence found in the current study was relatively small (1.2 to 1.3%), compared to a target of +7.5% set in published gait retraining studies,[38,42] such changes should not be neglected. There are other biomechanical adaptations, such as changes to foot-strike pattern and speed, which could also interact with the target training parameter. Since we proposed the use of PTA as a training target, which has been shown to be affected by both running speed and cadence,[46] a specific training threshold set for different slope conditions might therefore be required to ensure that the training target is relevant and attainable across various slope conditions.

CHAPTER 6

BETWEEN-SESSION RELIABILITY AND MINIMAL DETECTABLE DIFFERENCE FOR PEAK TIBIAL ACCELERATION DURING OVERGROUND RUNNING

6.1 Introduction

Impact mechanics during running have been studied for over three decades.[63] It has received significant attention due to its potential association with overuse running injuries.[4,21,166] Traditionally, impact mechanics are measured with in-ground force plates or instrumented treadmills.[63] Gait analyses were therefore confined to short runways or treadmills within labs. Physical constraints, such as the limited physical dimensions of the runway and the tightly regulated speed on a treadmill could affect the natural running gait.[103,167] Nowadays, impact mechanics can be measured outside of the lab, using lightweight accelerometers.

The measurement of axial and resultant PTA has become increasingly popular in out-of-lab research. For example, axial PTA has been measured to monitor fatigue in runners over extended runs (e.g. 110-km mountain ultramarathon).[168] Also, axial and resultant PTA have been measured to compare shock attenuation across various running surfaces.[100,117] Technology has paved the way for researchers to capture more natural running gait using wireless accelerometers.

Despite the widespread use, the reliability of measuring axial and resultant PTA using skin-mounted accelerometers has only been reported for limited conditions.[47,50,78] A previous study reported excellent within-session reliability but poor-to-moderate between-day reliability for axial PTA.[50] However, the study only analyzed four strides taken from separate short running bouts within the lab. Another study reported good reliability and moderate-to-good reliability for resultant PTA at 1 week and 6 months respectively.[47] In that study, runners ran on a treadmill at four pre-determined speeds. Biomechanical differences have been reported between treadmill and overground running,[79] and runners demonstrated more random patterns overground than on treadmills.[103] The reliability reported in previous studies may not be fully applicable to out-of-lab running assessments, where PTA is measured overground for consecutive strides over longer runs.

Oval running tracks are a common testing environment for overground, out-of-lab studies.[79,157] The choice of such a facility is possibly due to its resemblance to real-world running environments, accessibility, and the ability for researchers to control the running speed and other external factors. The length of the experiment (duration or distance) can also be easily manipulated by changing the number of laps. Evaluating the between-day reliability and statistically meaningful differences for continuous measurement of PTA along a running track might therefore be appreciated, especially for studies which evaluate changes across long-term interventions, such as gait retraining. Moreover, the analysis of additional running strides has been suggested to improve reliability.[50,78] The number of strides needed to obtain adequate reliability is yet to be determined.

This study aimed to determine the one-week and three-week reliability and the minimal detectable difference (MDD) of axial and resultant PTA during overground track running. Moreover, this study also aimed to determine the minimum number of strides required to achieve good and excellent between-session reliability for both axial and resultant PTA.

6.2 Methods

6.2.1 Participants

As part of a larger study, runners were recruited by convenience sampling around the campus of the University of Calgary. Recreational runners who were actively involved in running for more than 20 km per week were recruited. Runners with any active lower-extremity injuries within the previous 6 months or other musculoskeletal or neurological conditions that might affect natural running gait were excluded. All eligible participants were given a detailed explanation of the experiment prior to signing an informed consent approved by the Conjoint Health Research Ethics Board of The University of Calgary (REB16-1183).

6.2.2 Data collection

All participants completed three identical testing sessions (Baseline, Session 2 and Session 3) at the upper level of the Fitness Centre at the University of Calgary. The 200-m indoor running track was used in this study. Test sessions were scheduled one week and two weeks apart. Participants wore their personal running shoes, with the same pair being used for all three sessions.

During each session, an IMU containing a tri-axial accelerometer (Shimmer3®; Shimmer Inc., Ireland) was securely attached to the anteromedial aspect of the right tibia, with the y-axis aligned with the long axis of the tibia. The wide range accelerometer with a ±16-*g* operating range was set to sample tri-axial acceleration at 512 Hz. Participants were given five minutes to warm up along the track. After the warm-up, participants were asked to run 12 counter-clockwise laps at the inner-most lane (200 m) at a pace suitable for a 45-to-60-minute run. Each lap was timed and recorded by a researcher.

6.2.3 Data processing and analysis

Custom-written MATLAB scripts were used to process data in this study. Residual analysis was conducted to determine the cut-off frequency for filtering.[169,170] The acceleration data were filtered using a 2nd order Butterworth low-pass filter.[47,50] Resultant acceleration (AccR) was computed as the square root of the sum of the squared acceleration of the x-, y- and z-axes (i.e., $AccR = \sqrt{AccX^2 + AccY^2 + AccZ^2}$).

Data collected during the first and last minute were discarded to account for the acceleration and deceleration phases. Time of initial contact was defined as the local minimum which occurred within 0.075 s prior to a local maximum identified in the AccR signal. Peak axial (i.e., positive y-axis) and peak resultant acceleration were extracted within the first 40% of each stride as axial and resultant PTA.[100] The peaks were averaged across a selected number of strides for each test session for further analysis. The selected stride intervals include five to 300 with a one-stride interval, and 310 to 400 with a 10-stride interval.

One-way repeated measures ANOVAs were performed to compare the running speed and the average peak and resultant PTA at five, 10, 40, 100, 200 and 400 stride intervals across each test session. Test-retest reliability was evaluated by intraclass correlation coefficients ($ICC_{2,k}$) calculated using a two-way mixed-effects model, mean of k measurements with absolute agreement. The global significance for all statistical calculations was set at 0.05. ICC values were interpreted as poor, moderate, good and excellent for ICC < 0.5, 0.5 ≤ ICC < 0.75, 0.75 ≤ ICC < 0.9 and ICC ≥ 0.9 respectively.[47] The MDD was calculated using the equation $MDD = 1.96 \times \sqrt{2} \times (SD \times \sqrt{1 - ICC})$ where SD is the mean within-subject SD measured at Baseline.[50]

6.3 Results

Based on published reliability studies with a similar experimental design, it was determined that a sample of ten runners was needed in the present study.[47,50] Eleven recreational distance runners (5 males, 6 females; mass = 68.3 ± 11.4 kg; height = 1.7 ± 0.1 m; running experience = 5 – 15 years) completed this study. Session 2 was completed 11 ± 4.8 days after Baseline, and Session 3 was completed 23 ± 4.5 days after Baseline. The average speed were 3.09 ± 0.41 m/s, 3.06 ± 0.29 m/s and 3.09 ± 0.26 m/s for Baseline, Session 2 and Session 3 respectively, no significant difference was found in running speed (F = 0.094, p = 0.910) across sessions. Based on the results of the residual analysis, all data were low-pass filtered with a cut-off frequency of 75 Hz. The acceleration data were thoroughly inspected and were cleared for signal distortion caused by a ±16-g operating range accelerometer.

The mean and SD of axial and resultant PTA averaged across the selected number of strides are presented in Table 6.1. There was no significant difference found in axial (p = 0.522 – 0.693) or resultant PTA (p = 0.167 – 0.469) across the three sessions.

Overall, between-session reliability was moderate-to-good for axial PTA and moderate-to-excellent for resultant PTA when taking the average from five to 400 strides (Table 6.2). Figure 6.1 shows the $ICC_{2,k}$ for five to 300 at each stride interval and 310 to 400 at 10 stride intervals. To achieve good reliability (ICC ≥ 0.7), a minimum of 17 and 28 steps were needed for axial PTA, and to achieve excellent reliability (ICC ≥ 0.9) for resultant PTA, a minimum of 56 and 75 steps were needed. Compared to averaging 10 strides, increasing the number of strides to 100 reduced the MDD by 0.48 g for axial PTA and 1.54 g for resultant PTA.

Number of strides	Baseline (g)	Session 2 (g)	Session 3 (g)	One-way repeated measures ANOVA	
				F	P-value
Axial PTA					
5	7.94 ± 1.85	7.48 ± 1.12	7.48 ± 1.74	0.37	0.693
10	7.82 ± 1.71	7.35 ± 1.27	7.40 ± 1.56	0.50	0.612
40	7.95 ± 2.11	7.30 ± 1.16	7.53 ± 1.63	0.67	0.522
100	7.98 ± 2.06	7.26 ± 1.22	7.49 ± 1.66	0.57	0.577
200	8.06 ± 2.12	7.27 ± 1.18	7.56 ± 1.69	0.63	0.544
400	8.17 ± 2.18	7.33 ± 1.05	7.61 ± 1.70	0.54	0.589
Resultant PTA					
5	11.34 ± 3.13	11.12 ± 2.17	11.39 ± 3.34	0.95	0.403
10	11.25 ± 2.85	11.06 ± 2.46	11.26 ± 3.05	1.06	0.365
40	11.40 ± 2.93	10.90 ± 2.14	11.38 ± 2.95	1.96	0.167
100	11.46 ± 3.01	10.91 ± 2.33	11.19 ± 2.93	1.11	0.349
200	11.62 ± 3.25	11.00 ± 2.32	11.37 ± 3.02	1.38	0.275
400	11.86 ± 3.71	11.10 ± 2.41	11.44 ± 3.04	0.79	0.469

ANOVA, analysis of variance; g, gravitational constant (g = 9.81 m/s^2); PTA, peak tibial acceleration.

Table 6.1. Mean ± standard deviation and comparison of axial and resultant PTA averaged across different number of strides between sessions.

	$ICC_{2,k}$		MDD (g)	
Number of strides	Baseline *vs.* Session 2	Baseline *vs.* Session 3	Baseline *vs.* Session 2	Baseline *vs.* Session 3
Axial PTA				
5	0.73	0.67	3.01	3.70
10	0.68	0.70	3.32	3.30
40	0.75	0.82	3.09	2.77
100	0.77	0.83	2.96	2.70
200	0.77	0.84	2.98	2.73
400	0.77	0.82	2.93	2.91
Resultant PTA				
5	0.84	0.77	3.66	5.07
10	0.81	0.78	3.96	4.54
40	0.90	0.87	2.77	3.45
100	0.93	0.91	2.45	2.98
200	0.94	0.91	2.25	3.09
400	0.95	0.91	2.38	3.26

g, gravitational constant (g = 9.81 m/s^2); PTA, peak tibial acceleration; ICC, intraclass correlation coefficient; MDD, minimal detectable difference.

Table 6.2. Between-session reliability and MDD of axial and resultant PTA averaged across different number of strides.

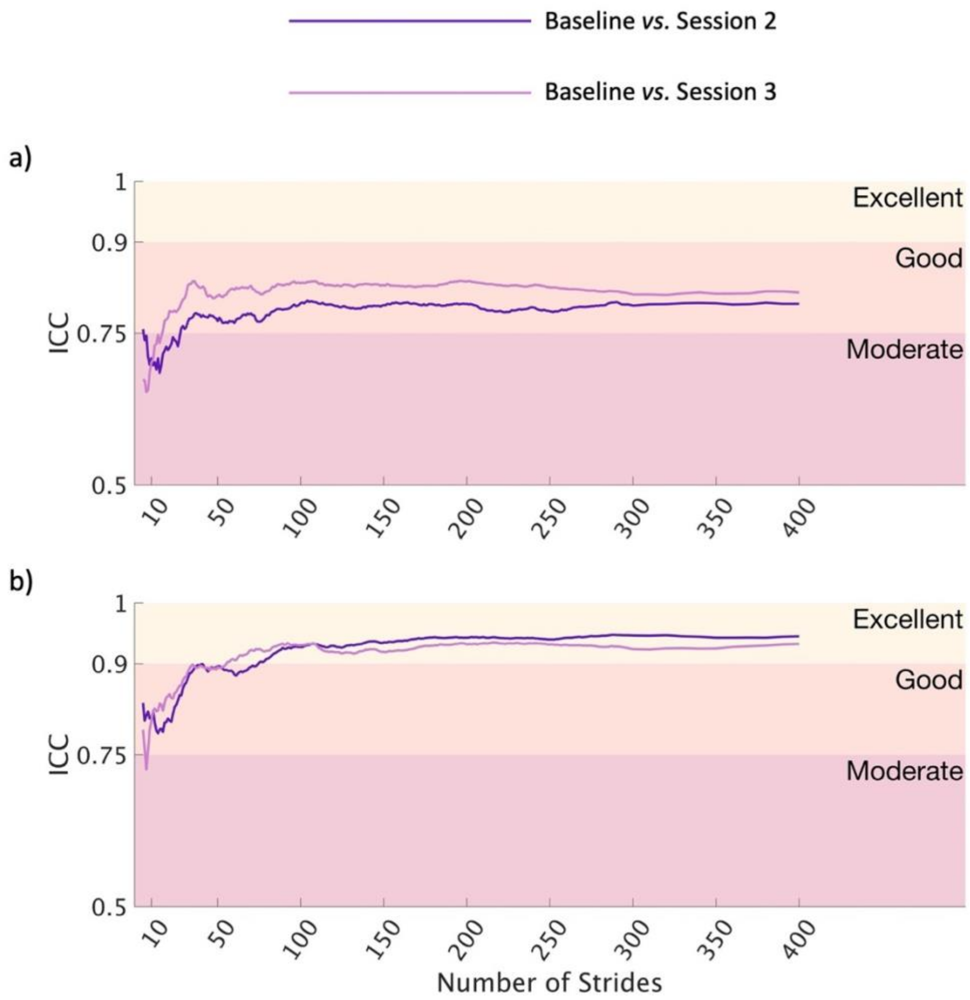


Figure 6.1. Between-session reliability of a) axial peak tibial acceleration (PTA) and b) resultant PTA averaged across five to 400 strides. ICC, intraclass correlation coefficient.

6.4 Discussion

The aim of this study was to establish the between-session reliability and MDD of axial and resultant PTA measured during overground running. Overall, axial PTA demonstrated moderate-to-good reliability, and resultant PTA demonstrated moderate-to-excellent reliability. This study also aimed to determine the number of strides required to achieve good or excellent reliability. Based on our findings, for axial PTA, we recommend averaging a minimum of 40 strides per session for good reliability with MDD of 2.93 *g*. For resultant PTA, it is recommended to average a minimum of 100 strides per session to achieve excellent reliability with MDD of 2.72 *g*.

Previous studies have reported within-session reliability on axial and resultant PTA measured overground.[50,78] Aubol *et al.* reported excellent within-session reliability for both axial (ICC = 0.99) and resultant PTA (ICC = 0.95) when the average was taken across the first five trials as compared to that taken across the last five trials within the same session.[78] Van den Berghe *et al*. also reported excellent within-session reliability (ICC ≥ 0.92) for both axial and resultant PTA across different speeds when comparing four strides taken within the same testing session.[50] As expected, the between-session reliability reported in our study was relatively lower than the within-session reliability. When runners were tested on separate days, there is a possibility of slight changes to the position of the accelerometer. While there are anatomical landmarks (e.g., medial malleolus) to guide researchers, identical placement is not guaranteed. A recent study found that a small proximal shift of 2 cm of the accelerometer resulted in a lower PTA.[77] Sensor placement difference between sessions could explain the lower between-session reliability compared to within-session reliability. In fact, Van den Berghe *et al*. repeated their test on a separate day and reported moderate reliability for axial (ICC = 0.58) and resultant PTA (ICC = 0.81),[50] comparable to our results at a similar speed when averaged across five strides.

With regards to comparing our results to between-session reliability measured for treadmill running, our results showed lower reliability in general. For axial PTA, Burke *et al.* found good between-session reliability (ICC = 0.80 – 0.82),[171] while we found ICC of 0.68 and 0.70 when averaging the same number of strides (i.e., 10 strides). The MDD (2.8 *g*) found in the treadmill study was also smaller compared to our study. Similarly, for resultant PTA, Sheerin *et al.* found excellent reliability (ICC = 0.91 - 0.96) for resultant PTA at a speed similar to that used in the current study.[47] Considering the difference, researchers should avoid relying solely on the reliability and MDD reported in treadmill studies for overground PTA measurements. The higher reliability and smaller MDD in treadmill running might be explained by the difference in variability between treadmill and overground running. There are physical constraints in treadmill running, including limited physical dimensions, single and uniform direction, and a regulated running speed, resulting in increased stride-to-stride regularity.[103] In the current study, runners were self-paced and ran along an oval track, higher variability in gait can be expected. While treadmill running might be the preferred protocol of choice given the higher reliability, it may not fully represent running in the real world. Fortunately, our results suggested that ICC for overground running can be improved by increasing the number of strides analyzed.

Increasing the number of strides averaged across each session has resulted in improved reliability and lower MDD previously, both within-session on a treadmill for axial and resultant PTA[78] and between-session overground for axial PTA.[50] Averaging 30 strides of treadmill running within the same session produced a 0.2 *g* and 0.4 *g* decrease in MDD for axial and resultant PTA compared to five strides.[78] In the overground between-session reliability study, the reliability of axial PTA shifted from moderate to good with a 0.4 *g* reduction in MDD when averaging six instead of four strides. However, there was insufficient data in this published study to examine if further increasing the number of strides would result in further improvement of between-session reliability.[50] This overground study was conducted within the lab, where each running trial along the 32-m runway produce one stride for analysis.[50] In this type of one-stride-per-trial study design, which accounts for 12% of IMU-based gait analysis,[80] fewer than 10 strides are usually available.[46] Increasing the number of strides would require additional running trials and would be time-consuming. In our study, runners were running along an oval track for 2.4 km. This allows continuous measurement of tibial acceleration, making it possible to collect a larger number of strides in a shorter amount of time. In the current study, each run lasted less than 16 minutes and more than 500 strides were available from each session for analysis.

Given enough strides per session, this study has determined the number of strides required to achieve good or excellent reliability. We have taken the average from the first five to 300 strides and compared it across sessions. Through visual inspection of data, the ICC value fluctuated and increased within five to 40 strides for axial PTA, and within five to 100 strides for resultant PTA, and leveled off into a plateau, suggesting no significant improvement in ICC by adding more strides. A simple "knee point detection" algorithm was used to confirm this observation, as presented in Appendix IV. For axial PTA, the minimum number of strides required to achieve good reliability was 32 across both days, and 75 strides were needed for resultant PTA to achieve excellent reliability.

The current study is the first to report between-session reliability and MDD for peak and resultant PTA through continuous tibial acceleration measurement. One limitation of this study is that we measured PTA along an oval running track. There are bends along the track which could add variability to the running gait. Reduced step velocity and stride length were found in sprinters during bend running.[172] In the current study, some runners demonstrated a periodic change in PTA along the 2.4 km run. An example of the raw acceleration signal (Subject 10, Baseline) is shown in Figure 6.2. A repeated pattern can be observed in the acceleration signal, and interestingly, each repetition was spaced 25 to 30 seconds apart, approximately the time for half a lap. The higher PTA could be recorded during the linear portion of the track. However, due to the lack of position data, we were unable to confirm this hypothesis. Stride-to-stride variability within our study could have lowered the reliability and increased MDDs compared to linear overground running, especially when averaged across few numbers of strides (< 40 strides). Another limitation was the limited generalizability of the results to conventional outdoor running where stride-to-stride variability is expected to be higher than along an indoor running track. Reliability and MDDs reported in our study should be used with caution for the interpretation of future study findings where the condition differs significantly from our testing condition.

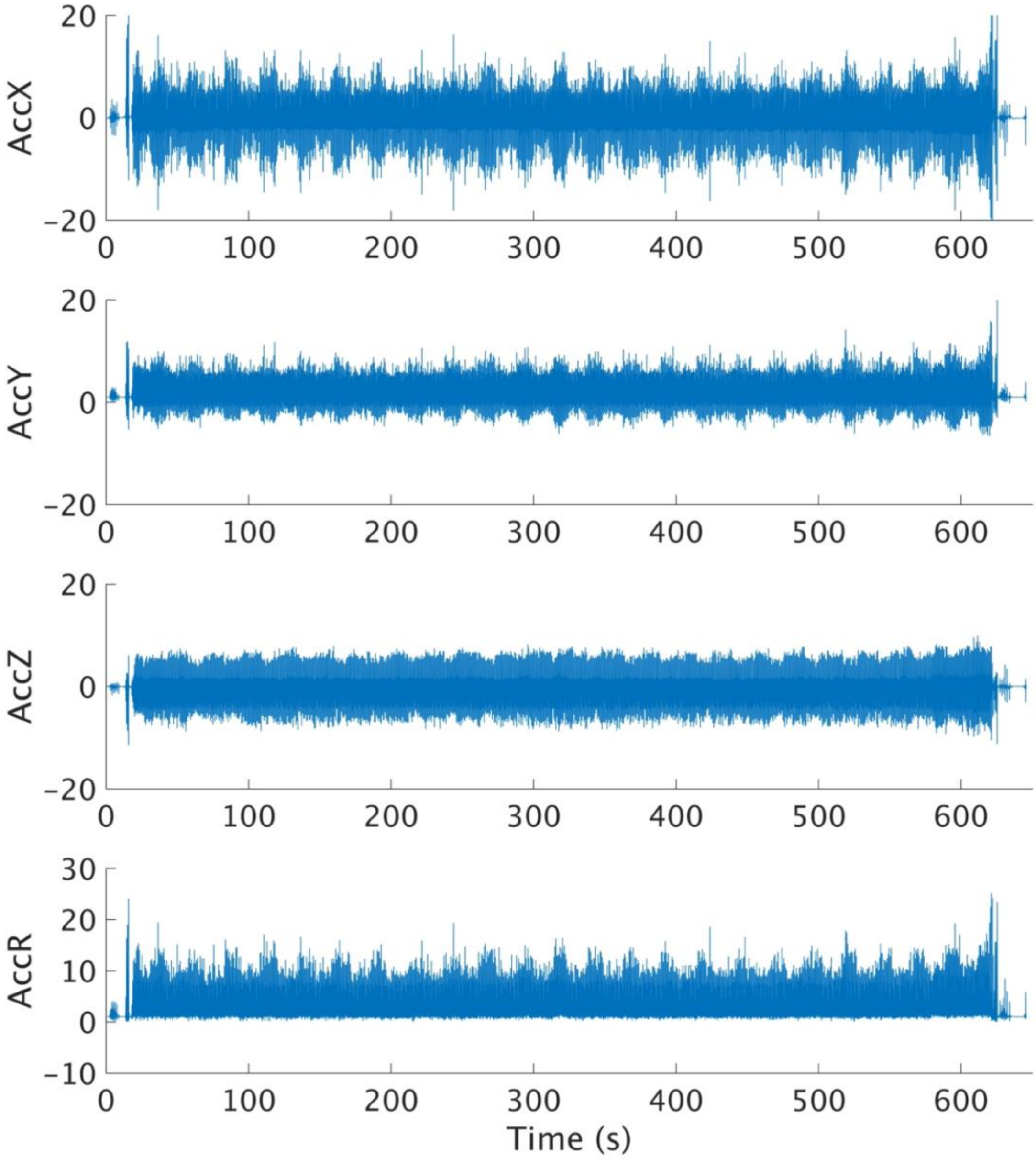


Figure 6.2. An example of raw tibial acceleration signal measured over a test session. Acceleration signals are displayed in g (gravitational constant, g = 9.81 m/s^2). AccX, AccY and AccZ: acceleration along x-, y- and z-axes, AccR: resultant acceleration.

6.5 Conclusion and implications for future studies

In conclusion, axial PTA demonstrated moderate-to-good between-session reliability, and resultant PTA demonstrated good-to-excellent reliability. In general, between-day reliability can be improved, together with a smaller MDD, by averaging more strides from each session. A minimum of 40 strides should be averaged when accessing axial PTA overground for good reliability, and 100 strides for resultant PTA for excellent reliability.

Based on our comparisons of baseline to one week and three weeks, axial and resultant PTA measured during overground running can be used to assess longer-term interventions, such as gait retraining. Gait retraining protocol typically lasts for two or three weeks, in which runners were assessed before and after.[36,44] A reliable measurement is crucial to assess the effect of training. The MDD reported in this study should be considered when interpreting results, as it represents the degree of change representative of a true change. The recommended number of strides and MDD values obtained in this study were considered when examining the proposed protocol.

CHAPTER 7

EFFECTS OF ACCELEROMETER OPERATING RANGE AND A CORRECTION ALGORITHM ON TIBIAL ACCELERATION MEASUREMENTS

7.1 Introduction

Distance running has become a popular activity for people of all ages around the world.[1] In the past 40 years, the running gait has been analyzed by sports scientists to evaluate performance and gain a better understanding of RRIs.[17,79] Traditionally, gait analyses were conducted in the lab. The insight gained from lab-based studies has informed injury treatment protocols and footwear designs.[12,173] However, recent studies suggested that gait parameters measured in controlled lab settings may not be representative of those produced in real-world conditions.[75,84,94,95]

High PTA has been established as a biomechanical risk factor of common RRIs.[17,21,27] Two retrospective lab-based studies found higher axial PTA among runners with a history of tibial stress fracture compared to healthy controls.[21,27] The values of PTA found in these lab-based studies have been used to screen runners for high risk of tibial stress fracture, and provide gait retraining to those in need.[30,34,35] However, PTA measured in the lab was found to be lower than that measured under real-world conditions.[95] Furthermore, cross-sectional lab-based studies may not completely capture the full range of PTA experienced by the runners.[2,17] Longitudinal data under real-world conditions, preferably on a large scale, can help to determine appropriate values for assessing injury risk.

Tibial acceleration can be measured outside of the lab using wireless IMUs or accelerometers. For example, two studies have continuously monitored tibial acceleration during marathon races using research-grade IMUs.[74,75] Unfortunately, the use of specialized equipment has limited the length of the monitoring period and the sample size.[74] Valuable data might also be lost if the IMU was not returned to the researchers after the data collection.[75] These logistic limitations might be overcome by using runner-owned devices. Wearable technology is widely adopted within the distance running community.[59] The most popular device used by runners is the GPS-enabled smartwatch. Most smartwatches are equipped with an IMU or accelerometer.[59] Real-world data obtained from runner-owned wearables may be a valuable source of information to scientists. It is possible to obtain a large database of tibial acceleration from runners if the devices can be reconfigured to measure tibial acceleration.

Commercially-available IMU or accelerometer-based systems may not have the same technical specification as those used by researchers. The operating range of the accelerometer might be narrower in runners-owned devices. In the case where the acceleration signal exceeds the operating range of the sensor, the measured signal could be clipped and the amplitude of the peak may not be registered. An example of a clipped axial tibial acceleration signal is shown in Figure 7.1. A previous study which investigated the influence of operating range on PTA measurements found that using a ±8-*g* accelerometer could induce an error of up to 2.65 *g*.[73] Hence, the first aim of the current study was to examine the accuracy of using a ±16-*g* accelerometer to measure axial and resultant PTA. This range was selected for its popularity among wearable devices.[61]

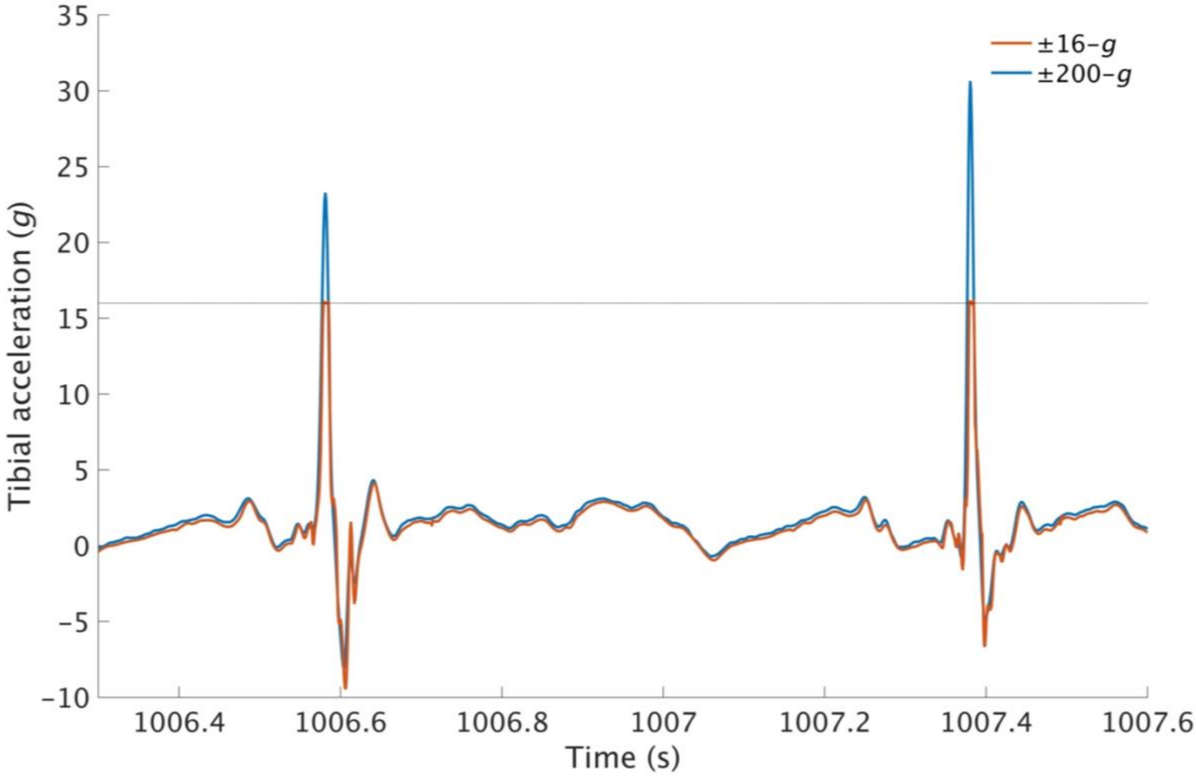


Figure 7.1. Axial tibial acceleration measured using low (±16-*g*) and high operating range accelerometers (±200-*g*).

Dotted line indicates the upper limit of the ±16-*g* accelerometer.

Acceleration is presented in unit of *g* (gravitational constant, *g* = 9.81 m/s^2).

A correction algorithm has been put forward by Ruder *et al.* to restore clipped tibial acceleration signals.[74] In the study, the authors identified all PTA between 15.0 to 15.9 *g* and artificially clipped the data at 15 *g*. An algorithm using spline interpolation was used to recalculate the missing peak values. On average, the peaks were found to be underestimated by 0.02 *g* (95% CI: -0.48 – 0.44-g). This simple algorithm has the potential for use to reconstruct peaks from clipped signals, compensating for the limitation of a narrow operating range found in runner-owned devices. However, the performance of the algorithm to restore peaks beyond 15.9 *g* has not been validated. Furthermore, resultant PTA, another commonly measured parameter,[47] have not been validated and therefore merit validation. The second aim of the current study was to assess the measurement agreement of axial and resultant PTA obtained from the restored signal.

In this study, signals from three conditions were used to obtain axial and resultant PTA. The three conditions included: 1) signal obtained using a low operating range accelerometer (LowOR-raw), 2) LowOR-raw signal but restored using the correction algorithm (LowOR-restored) and 3) signal obtained using a high operating range accelerometer (HighOR), which was considered the ground truth for comparison.

7.2 Methods

7.2.1 Participants

Participants were recruited by convenience sampling around the Campbelltown Campus of Western Sydney University. Individuals with any active lower-extremity injuries or other musculoskeletal or neurological conditions that might affect natural running gait at the time of data collection were excluded. All eligible participants were given a detailed explanation of the experiment prior to signing an informed consent approved by the Human Ethics Committee of Western Sydney University (H14514).

7.2.2 Data collection

All participants completed a data collection session at the Campbelltown Campus of Western Sydney University. An outdoor concrete course 270 m in length with changes in gradient (±15.8%) was used in this study. Participants wore their personal running shoes.

Prior to data collection, two wireless IMU sensors (Blue Trident IMU, Vicon, New Zealand) each housing two accelerometers with different operating ranges, Low-G (±16-*g*) and High-G (±200-*g*), were securely attached to the anteromedial aspect of both tibiae, with the y-axis aligned with the long axis of each tibia. Tri-axial acceleration data were sampled at 1,600 Hz and 1,125 Hz for the LowOR and HighOR accelerometers respectively. Participants were given five minutes to warm up along the course. After the warm-up, participants were instructed to complete four laps at their fastest pace.

The choice of an outdoor concrete running route with steep incline and decline, and instructing the participants to run at a fast pace, were particularly designed to capture a wide range of axial and resultant PTA. Tibial acceleration signals which exceed the operating range of the LowOR accelerometer (i.e., ±16 *g*) were required for comparison with the HighOR accelerometer and validation of the restoration algorithm.

7.2.3 Correction algorithm to restore clipped signals

A correction algorithm was adopted from one described in a previous study,[74] it was modified for use with our higher sampling frequency and translated to another programming language (Python to MATLAB). A detailed description of the algorithm is provided in Appendix V. To ensure the algorithm used in this study was comparable to the one developed by Ruder *et al.*, a validation was conducted using the same method as described.[74] More specifically, an artificially clipped signal was generated by replacing the data points between 15 to 16 *g* in the axial component of the LowOR signal with a value of 15 g. For this validation, we compared 1,240 pairs of axial PTA. The reconstructed axial PTA was found to be underestimated by 0.03 ± 0.29 g, as compared to the results obtained in Ruder *et al.*'s validation (underestimation of 0.02 ± 0.24 g). Based on comparable performance, we determined that the adoption of the algorithm for use in the current study was successful.

7.2.4 Data processing and analysis

Tri-axial acceleration data were downloaded from the sensor after each data collection session. Custom-written MATLAB scripts were used to process data in this study. The LowOR acceleration signals were duplicated as LowOR-raw and LowOR-restored, and the LowOR-restored signal was processed with the restoration algorithm. The acceleration signals collected through the High-G sensor were labeled as HighOR. Identical MATLAB scripts were used to filter and extract axial and resultant PTA from each set of acceleration signals (LowOR-raw, LowOR-restored and HighOR).

Each set of acceleration signals was filtered using a 2nd order Butterworth low-pass filter with a cut-off frequency of 85 Hz. The resultant acceleration (AccR) was computed as the square root of the sum of the squared acceleration of the x-, y- and z-axes (i.e., $AccR = \sqrt{AccX^2 + AccY^2 + AccZ^2}$). Time of initial contact was defined as the local minimum which occurred within 0.075 s prior to a local maximum identified in the AccR signal. Peak axial (i.e., positive y-axis) and peak resultant acceleration were extracted within the first 40% of each stride as axial and resultant PTA.[100] A data processing flow diagram is presented in Figure 7.2. The axial and resultant PTA of the LowOR-raw and LowOR-restored conditions were matched with the corresponding peaks obtained from the HighOR signal within a 0.2 s window, which was considered the ground truth.

Two-sample independent *t*-tests were used for intra-subject comparison of axial and resultant PTA obtained from the conditions LowOR-raw and LowOR-restored against the ground truth. The global level of significance for the *t*-tests was set at 0.05. To evaluate the measurement agreement of axial and resultant PTA obtained from the LowOR-restored signal and the ground truth, Bland-Altman plots combined with the calculation of 95% confidence intervals ($CI_{95\%}$), also known as the limit of agreement, were used.[174,175] The agreement was considered acceptable if the difference in the outer limits (upper $CI_{95\%}$ – lower $CI_{95\%}$) was within the MDD obtained in Study 4.

Based on the observation of the pilot data, it was hypothesized that the performance of the correlation algorithm might vary depending on the magnitude of the peaks. The axial and resultant PTA were segregated into different ranges based on the ground truth, in increments of 4 *g*. The differences in the outer limits were calculated and compared accordingly.

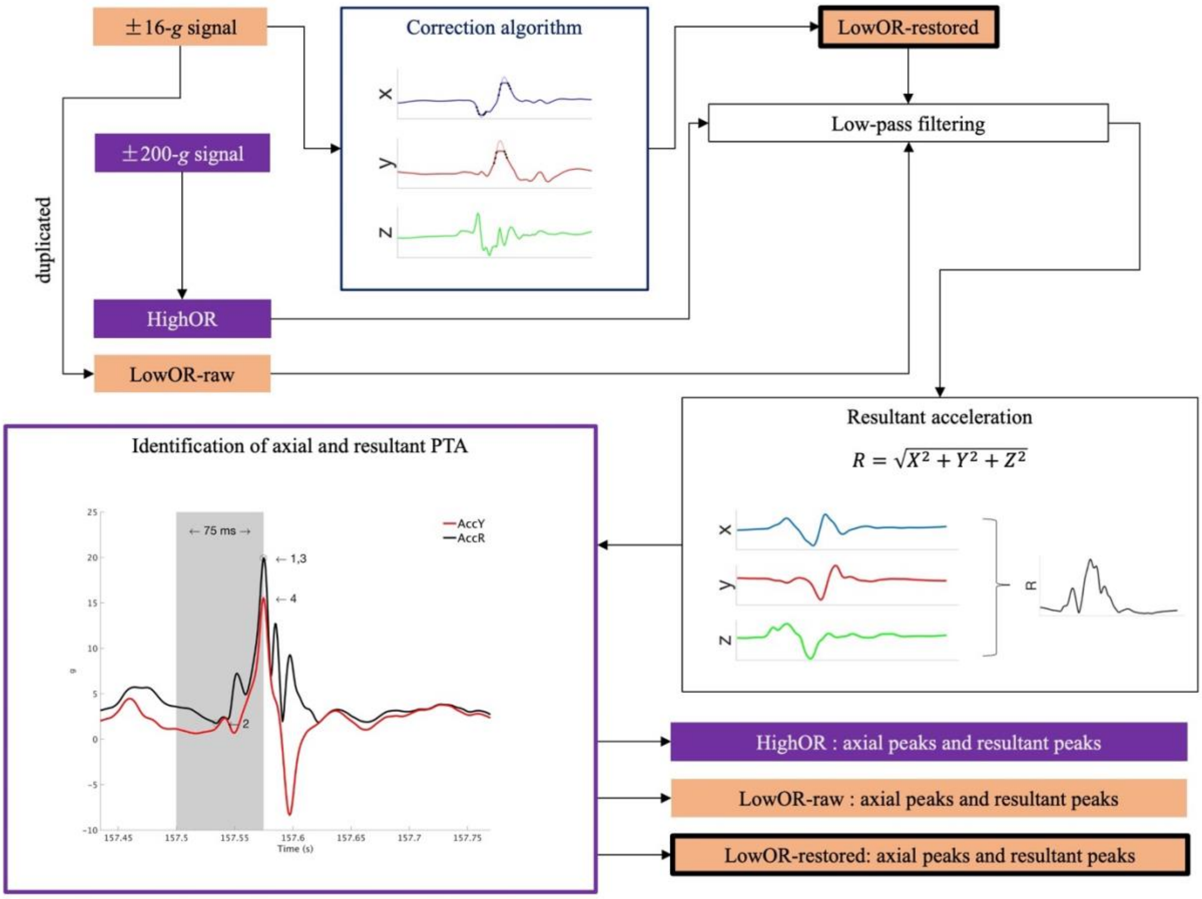


Figure 7.2. Data processing flow diagram.

AccY, axial acceleration signal; AccR, resultant acceleration signal; PTA, peak tibial acceleration.

7.3 Results

Based on our pilot data, we estimated that each participant will contribute 850 – 950 steps for validation, and a minimum of 20 participants were needed to obtain 18,000 steps for validation. Twenty-four physically active individuals (13 males and 11 females; mass: 68.4 ± 12.7 kg; height: 1.7 ± 0.1 m) participated in this study. A total of 40,855 peaks were obtained.

The HighOR acceleration signal did not show any clipping in any of the axes. The axial and resultant PTA were found to be 14.38 ± 3.07 *g* and 20.33 ± 5.83 *g* respectively. Significant differences ($p < 0.05$) were found in all 24 participants for both axial and resultant PTA derived from the LowOR-raw signal and the HighOR signal (Figure 7.3). On average, the PTA values were underestimated by 2.75 ± 1.76 *g* for axial PTA, and 4.56 ± 3.48 *g* for resultant PTA.

Comparing the peaks derived from the LowOR-restored signal with that obtained from the HighOR signal, significant differences ($p < 0.05$) were found in 71% (n = 17) and 50% (n = 12) of the participants for axial and resultant PTA. While peaks from the restoration signal were still underestimated, a smaller difference of 0.75 ± 0.67 *g* and 0.63 ± 0.76 *g* were found for axial and resultant PTA.

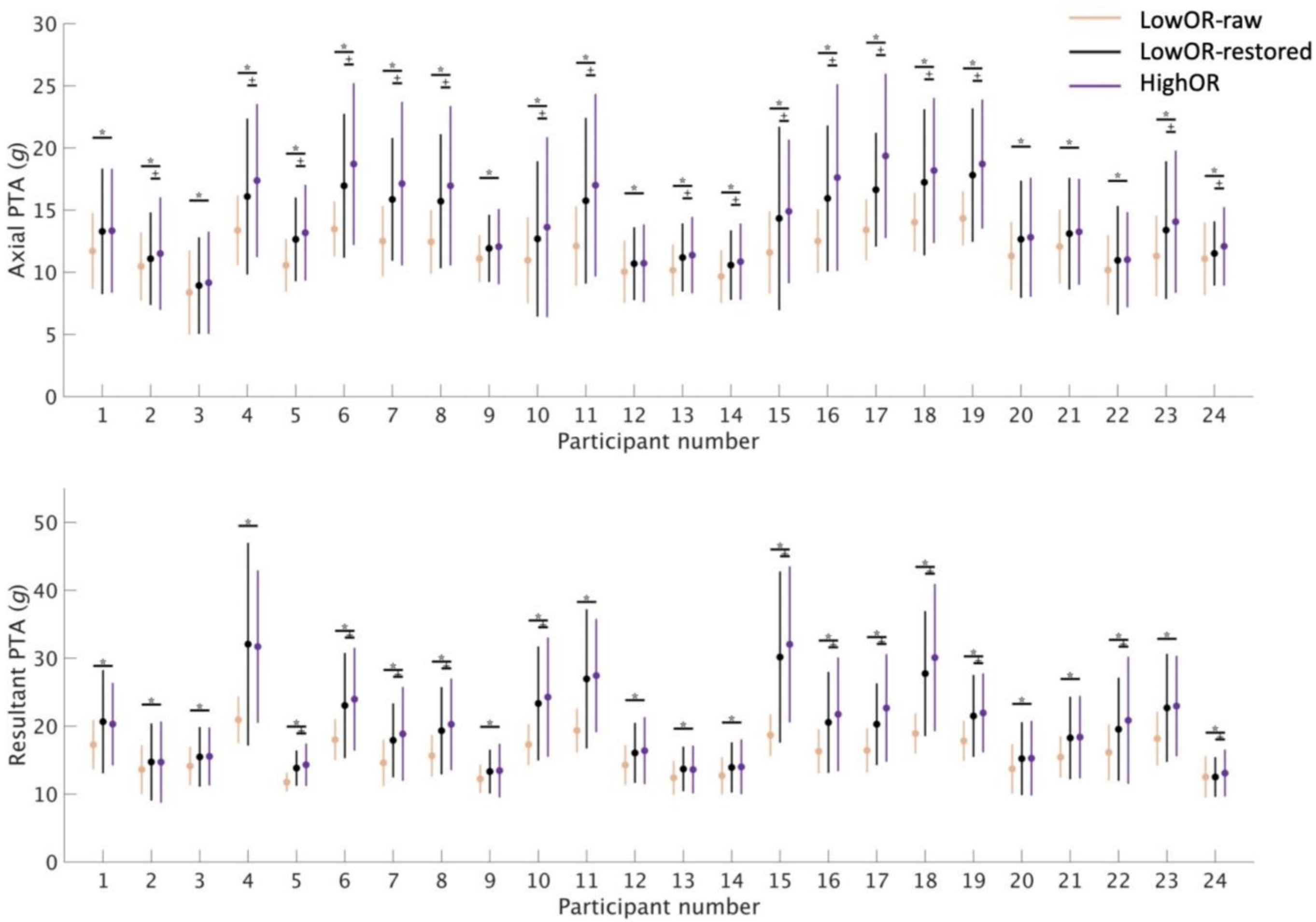


Figure 7.3. Within-subject mean and standard deviation of axial (top) and resultant (bottom) peak tibial acceleration (PTA) derived from different signals.

PTA is presented in unit of *g* (gravitational constant, $g = 9.81$ m/s^2). * denotes $p < 0.05$ between Low-G-raw and High-G; + denotes $p < 0.05$ between Low-G-restored and High-G.

Regarding axial PTA, a total of 13,805 peaks exceeding 16 *g* were identified from the HighOR signal. The algorithm rejected 18.6% of the peaks from the LowOR-restored signal for having a value below 16 *g* (Table 7.1). A total of 11,235 pairs of peaks were compared. On average, the restored peaks were lower than the ground truth by 1.40 ± 4.53 *g*. The Bland-Altman plot for axial PTA is presented in Figure 7.4a. The number of peaks, rejection rate, mean difference and $CI_{95\%}$ of each range (i.e., 16 – 20, 20 – 24, 24 – 28 and > 28 *g*) are presented in Table 7.1.

We observed signal clipping in each of the three axes in the LowOR-raw signal. Most clippings were observed in the y-axis, with signals from 14,388 foot-strikes across the 24 participants being restored. Within the x- and z-axis, 13,031 and 5,789 foot-strikes were being restored by the correction algorithm respectively. Resultant PTA obtained from foot-strikes with at least one of the axes restored were compared against the resultant PTA obtained from the HighOR signal. A total of 20,614 pairs of resultant PTA were compared. On average, the restored resultant PTA values were lower than the ground truth by 1.23 ± 5.48 *g*. The Bland-Altman plot for resultant PTA is presented in Figure 7.4b. The number of peaks, mean difference and $CI_{95\%}$ of each range (i.e., 16 – 20, 20 – 24, 24 – 28, 28 - 32 and > 32 *g*) are presented in Table 7.1.

	n	High-G (g)	Low-G-restored (g)	Mean difference (g)	95% confidence intervals Lower bound	Upper bound	Rejection rate (%)
Axial PTA							
All	**13805**	**21.55 ± 4.54**	**20.15 ± 4.72**	**-1.40 ± 4.53**	**-10.28**	**7.47**	**18.62**
16 – 20 g	7205	18.11 ± 1.05	17.90 ± 2.22	-0.21 ± 2.06	-4.25	3.82	27.84
20 – 24 g	3759	21.80 ± 1.15	20.75 ± 3.91	-1.04 ± 3.89	-8.67	6.58	8.09
24 – 28 g	1708	25.69 ± 1.12	23.12 ± 6.04	-2.57 ± 6.02	-14.37	9.23	8.49
> 28 g	1133	31.92 ± 4.55	25.01 ± 6.98	-6.92 ± 7.58	-21.77	7.93	10.15
Resultant PTA							
All	**20614**	**26.66 ± 7.94**	**25.43 ± 8.51**	**-1.23 ± 5.48**	**-11.97**	**9.52**	
16 – 20 g	3923	18.42 ± 1.07	17.96 ± 1.44	-0.45 ± 0.93	-2.28	1.38	
20 – 24 g	5598	21.93 ± 1.13	21.43 ± 2.12	-0.50 ± 1.84	-4.11	3.11	
24 – 28 g	4144	25.88 ± 1.14	25.05 ± 3.40	-0.83 ± 3.23	-7.17	5.51	
28 – 32 g	2706	29.84 ± 1.16	28.42 ± 5.30	-1.42 ± 5.23	-11.66	8.83	
> 32 g	4243	39.26 ± 6.62	36.09 ± 11.07	-3.17 ± 10.39	-23.53	17.19	

n, number of peaks; g (gravitational constant, g = 9.81 m/s^2); PTA, peak tibial acceleration.

Table 7.1. Measurement agreement for axial and resultant PTA derived from the restored signal (Low-G-restored) and true value (high-G).

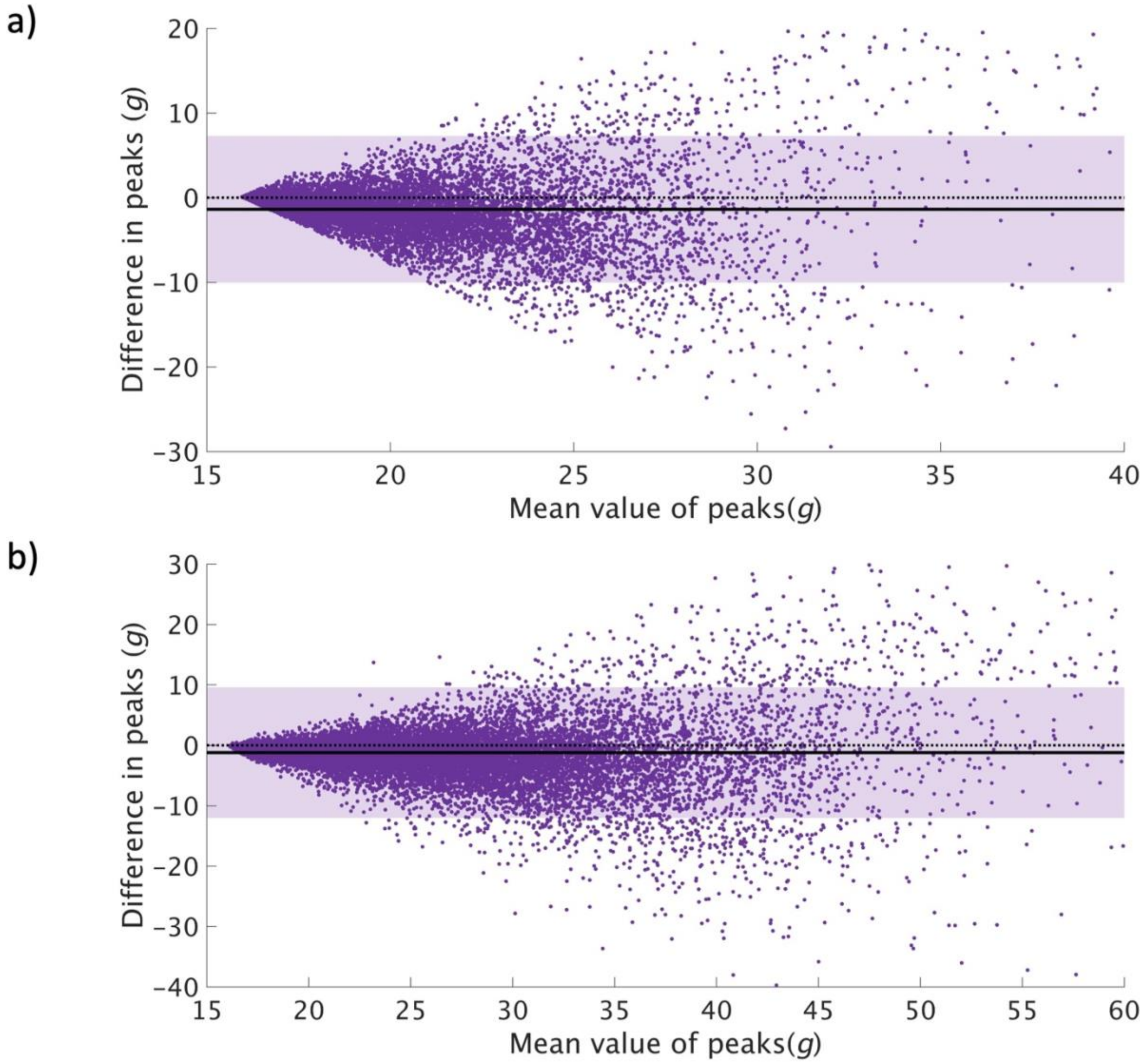


Figure 7.4. Bland-Altman plots for a) axial and b) resultant peak tibial acceleration. Solid line represents the mean difference (HighOR – LowOR-restored). A negative difference indicates a smaller value obtained using the restored signal. The dotted line represent perfect agreement (i.e., difference = 0). The shaded region represents the 95% confidence interval. PTA is presented in unit of *g* (gravitational constant, *g* = 9.81 m/s^2).

7.4 Discussion

The first aim of the current study was to examine the accuracy of using a ±16-*g* accelerometer to measure axial and resultant PTA. In general, significantly lower values were obtained from the low operating range accelerometer as compared to the ground truth. Based on our results, accelerometers of ±16-*g* are not sufficient for tibial acceleration measurements. The second aim was to evaluate the performance of a correction algorithm designed to restore signals that exceed the operating range. The axial and resultant peaks obtained from the restored signals have a smaller error compared to the raw signals, and yet, the peaks obtained were still found to be significantly lower. We also observed a wider $CI_{95\%}$ at peaks of higher magnitude, which confirmed the hypothesis about the correction algorithm performing better with peaks of lower magnitude. Overall, regardless of whether the correction algorithm is applied, using a low operating range accelerometer to measure axial and resultant PTA should be avoided.

A large database of tibial acceleration data is required for further understanding of high PTA as a biomechanical risk factor of injury.[17,21,27] To overcome logistic limitations, we proposed the reconfiguration of runner-owned devices with lower operating ranges to measure tibial acceleration in real-world conditions. However, based on the findings of this study, the use of ±16-*g* accelerometers to measure axial and resultant PTA could result in errors beyond the acceptable range. Compared to lab environments, axial and resultant PTA were found to be higher in real-world conditions.[95] Besides, there are situations where runners might choose to run at higher speeds (e.g., during a competition) or downhill, which can further increase axial and resultant PTA.[46] Tibial acceleration measured in real-world conditions is very likely to exceed ±16-*g* and would result in clipped signals.

According to a recent systematic review of wearable sensors used for gait analysis, ±16-*g* is the most common operating range for accelerometers,[61] which is also the minimum operating range recommended by Mitschke *et al.*[73] Systems with ±16-*g* accelerometers, for example, the IMeasureU (Vicon, New Zealand)[40,74,176] and the AcelSystem (Blautic, Spain),[177,178] have been used in research studies to measure tibial acceleration. We observed systematically skewed data when using the LowOR sensor without the correction algorithm. A large number of axial PTA obtained from the LowOR-raw signal were found at 16 *g*. This is likely explained by the LowOR accelerometer recording the maximum value (i.e., 16 *g*) when the signal exceeds the operating range, resulting in consecutive data points of 16 *g* as shown in Figure 7.1. The MATLAB script we used to identify peaks returns the maximum value within the first 40% of each foot-strike stride. The value of these falsely identified "peaks" therefore skewed the data. Based on such findings, we recommend studies which have or will use a ±16-*g* accelerometer to inspect the tibial acceleration signal for potential clipping, especially when the distribution of the peaks obtained is skewed toward the maximum operating range.

The correction algorithm was able to restore the clipped signal and reduce the error. However, an overall underestimation of the peak values was still found after the signal was restored. With our pilot data, we observed that the data points in the Bland-Altman plot spreading out wider as the magnitude of the peaks get larger. Based on this observation, we hypothesized that the correlation algorithm might be better at reconstructing peaks of lower values. This hypothesis was confirmed by the increase of mean difference across the ranges of peak values in both axial and resultant PTA. The differences between the $CI_{95\%}$ limits were compared against the MDD found in Study 4, 2.70 *g* for axial and 2.38 *g* for resultant PTA, to interpret if the error range was acceptable. Unfortunately, the limits for both axial and resultant PTA in all ranges were outside of the acceptable range. We do not recommend the use of the correction algorithm since it is highly possible that the error would mask the true difference of interventions and lead to invalid interpretations. Nevertheless, the acceptable limit used in our study may not reflect all conditions. A study-specific range may be determined based on clinical or research necessity, with reference to the peak values and effect size estimated specific to each study.[179] In the case where accelerometers with higher operating ranges are not available, and researchers have the intention to use the correction algorithm, the limits of agreement reported in the current study should be consulted. Based on our results, we strongly discourage the use of the correction algorithm on clipped tibial acceleration signals if the peak values are expected to be large and the estimated effect size is within the corresponding limits of agreement.

The correction algorithm was designed to reject reconstructed peak values of below 16.0 *g*. A similar criterion was set in Ruder *et al.* algorithm.[74] In principle, the peak value within the reconstructed signal should not be lower than the limit of the operating range as clipping would not have occurred in such a case. In our study, over 18% of axial PTA were rejected. For peaks within the range of 16 – 20 *g*, more than 27% of the reconstructed peaks were rejected. Since rejecting peaks using a set threshold could have a higher potential of removing peaks with lower values, the data set could be skewed towards higher values. Unfortunately, a comparison of the rejection rate was not possible since the two previous studies using the correction algorithm did not report a percentage.[74,75] We recommend future studies using this algorithm to report the rejection rate.

A few limitations of the current study should be noted. The assessment protocol was designed to obtain high axial and resultant PTA values, which is expected within real-world running. However, the mean error and limits reported may not be directly applicable to other conditions. Future studies may refer to the results as a more conservative estimation of error when deciding if a higher operating range is necessary. Secondly, position data was not collected in this study, therefore we were not able to perform sub-group analysis on peaks obtained under different surface inclinations. The tibial acceleration signal profile has been found to vary between uphill, level and downhill running,[105,111] which could influence the magnitude and timing of the peaks in both the axial and resultant acceleration signals. Based on the current analysis, it is unsure if the performance of the restoration algorithm varies under different slope conditions. Lastly, the sampling frequency and the resolution are different between the LowOR and the HighOR accelerometer and could result in minor differences in the PTA measurements.

7.5 Conclusion and implications for future studies

In conclusion, both axial and resultant PTA obtained from the ±16-*g* sensor were smaller than the true value. Accelerometers of this range are not suitable for tibial acceleration measurements. A thorough inspection for clipping should be conducted prior to filtering the acceleration signals to avoid invalid PTA measurements. The correction algorithm was able to restore peaks and reduce error, however, the axial and resultant PTA values were still found to be smaller than the true value. For axial PTA, a larger error was found at higher peak magnitudes. Based on the findings, it is not recommended to use ±16-*g* accelerometers, with or without the correction algorithm, for the measurement of axial and resultant PTA. The reported the limits of agreement should be taken into consideration for future investigations.

Based on the findings, we have chosen a wide-range (±200-*g*) accelerometer for training and assessment of our proposed training protocol which was conducted within similar conditions (outdoor running route with changes in surface gradient) of the current study.

CHAPTER 8

GENERAL DISCUSSION AND CONCLUSION

8.1 General discussion

The main objective of this thesis is to develop a gait retraining protocol for runners to modify their gait under real-world conditions. To address this objective, we have identified the limitations of treadmill-based gait retraining, examined the habitual grade-specific adaptations in real-world running and established the technical specifications for using wearable IMU sensors to assess PTA outside of the lab.

Based on our findings, we propose the use of an adaptive biofeedback protocol for gait retraining along overground trails with elevation changes.

8.1.1 Limitations of conventional gait retraining protocols

For the past 13 years, over 20 gait retraining studies have been published.[7,12,20,56] These studies have demonstrated promising results in alternating running biomechanics in injured and healthy runners.[7,12,56] Even though the intervention type varies considerably, a common goal of the training was injury prevention among runners with a high risk of RRI. Two gait retraining studies measured the injury outcome directly,[14,15] and other studies measured surrogate variables that associate running biomechanics and injury risk.[12] Common surrogate variables include the vertical GRF loading rates (i.e., VALR and VILR), which have been measured and compared to assess the effectiveness of gait retraining in at least seven RCTs.[12,14,15,34,41,42,180,181] Although the evidence between high loading rates and injury risk is limited, a reduction in VALR and VILR is generally considered preventive of common RRIs.[4,17]

A major component of gait retraining is real-time biofeedback. Modification to the gait pattern requires practice, and extrinsic feedback provided during practice could enhance this process.[6] Feedback used in previous gait retraining studies is either directly related to impact loading rates or indirectly through cadence or foot-strike pattern.[12] Immediate reductions in VALR and VILR have been demonstrated by increasing 10% of cadence or switching from a RFS to MFS/FFS.[57] Examples of direct impact-related feedback are visual displays of the GRF[14] or tibial acceleration,[35] and indirect feedback includes the visual display of cadence[42] or foot-strike pattern.[31] Providing real-time feedback requires lab-based equipment, such as wired accelerometers, instrumented treadmills and/or motion capture systems, making treadmill-based training practically easier than overground.[13] Furthermore, treadmill-based training allows researchers to supervise the training and control running speed. Structured, supervised training on a treadmill eliminates extraneous variables that might affect the outcome of gait retraining studies.[79]

The popularity of training overground versus on a treadmill may differ slightly across regions, yet large-scale surveys on recreational runners found less than 6% of runners predominantly train on a treadmill.[60,82,83] Most runners prefer the road, trail or track.[83] From an injury prevention perspective, most runners would need to adopt the modified running pattern during overground running for optimal clinical benefits. A few gait retraining studies have assessed the training effect overground after a course of treadmill-based training, which suggested partial transfer,[30,34] and yet, the transferred training effect to overground might be reduced.[40,44]

Study 1 was designed to examine to what extent the biomechanical effects of gait retraining on a treadmill could be carried-over to overground running. Twelve habitual RFS runners completed an eight-session gait retraining on a treadmill, modifying their foot-strike pattern to MFS with a real-time visual display of their foot-strike pattern. While changes in foot-strike pattern were carried-over to overground running, the reduction in VILR during treadmill running was not observed in overground running. This suggested an incomplete carry-over and established a need for gait retraining to be conducted overground.

Slopes are often found within real-world outdoor running environments. Many outdoor running courses, including the race course of the Boston Marathon,[182] have a mix of uphill and downhill segments. Study 2 was designed to examine the carry-over of training effects from a level surface gait retraining to sloped running conditions. Similar to Study 1, habitual RFS runners completed a gait retraining which modifies their foot-strike pattern. Runners were trained along an oval running track. Level surface running tracks have been used by another research group for overground gait retraining.[36,37] Training along the tracks allows runners to train in a condition that is closer to their regular training condition, while still allowing researchers to control training speed and supervise the training. After the gait retraining in Study 2, VALR and VILR were reduced in both level and uphill running conditions, but not during downhill. Further inspection of the results revealed that runners demonstrate natural adaptation in foot-strike patterns when running on slopes, and these adaptations might have interacted with the training effect.

The incomplete carry-over from traditional training conditions (i.e., treadmill and level running surface) to conditions that resemble real-world running (i.e., overground and slopes) established the need for a change in the training protocol. These limitations can be overcome by incorporating gait retraining within the runners' routine training, allowing practice within a natural training environment. Simple wireless equipment that can provide real-time feedback outside of the lab and a protocol designed for training outdoors with changes in elevation are fundamental for gait retraining under real-world conditions.

8.1.2 Wearables for gait retraining

Apart from the incomplete carry-over to real-world running conditions, low accessibility has been a major shortcoming for lab-based gait retraining.[9] Runners with a high risk of RRI may not have access to expensive, specialized, research-grade equipment required for gait assessment and real-time biofeedback. Fortunately, the recent development in wearable technology has created the possibility for conducting gait retraining outside of the lab.

Wearable devices, such as GPS-enabled smartwatches, were used by over 60% of runners.[58] While accuracy can be affected by the device model and running location,[183] GPS-enabled smartwatches are generally considered valid tools for measuring distance and speed outside of the lab.[184] Other commonly used wearables in running gait analysis are IMUs, accelerometers and pressure insoles, providing outcome measures including cadence, stride length, ground contact time and tibial acceleration.[61] Wearable systems have been used in gait retraining to provide real-time feedback. In Willy *et al.*'s study, a smartwatch paired with an accelerometer at the shoe provided cadence information to runners training to increase their cadence by 7.5%.[42] Another research group used a system consisted tibia-mounted accelerometers, a tablet and headphones to provide audio feedback on PTA when training along an overground running track.[36,37] In study 2, we used a foot-strike detection insole and provided real-time audio feedback to runners regarding their foot-strike pattern.[70] These applications have demonstrated the capability of wearable systems for use in gait retraining outside of the lab.

In addition to the accessibility and availability of wearable systems, the reliability and validity of using such systems are also crucial for running gait assessments and gait retraining. A series of studies by Benson *et al.* have suggested differences in biomechanics and variability between treadmill/indoor running and under real-world conditions.[159,185,186] External factors within the real world should be considered when designing assessment and training protocols. Reliability should also be established for measuring running biomechanics outside of the lab. Study 4 aimed to examine the between-session reliability of measuring PTA, a surrogate measure of loading rates,[50,51] during overground running. A considerable difference in reliability was found between our study and that previously reported for treadmill running.[47,171] Overground running has been associated with higher variability,[103] and could lead to lower reliability if adopting the same protocol as treadmill running. Results of Study 4 suggested a larger number of strides should be averaged from each session to improve the reliability in overground running.

The validity of PTA measurements can also be affected by the technical specifications of the accelerometer or IMUs used, such as the operating range.[73] Study 5 examined the accuracy of PTA measurements using an accelerometer with a low operating range of ±16-*g*. Based on a recent systematic review, this operating range is the most commonly used range in running gait analysis,[61] and was previously recommended as the minimum operating range required.[73] Results of Study 5 showed data skewing when using a ± 16-*g* accelerometer, and values of the peaks were found to be significantly lower than their true value. We have also validated a correction algorithm adopted from a previous study to restore peak values,[74] yet, peak values were still found to be underestimated. Based on the findings, we recommend using accelerometers with a higher operating range for accurate measurements of PTA.

8.1.3 Gait retraining under real-world conditions

The fundamental differences between supervised lab-based training and free-form real-world training should be taken into account when designing the training protocol. The structure of training (i.e., number of sessions), ease of implementation, feedback modality, and external factor within real-world conditions should be considered.

Lab-based training often has a set schedule, it ranges from three sessions in one week[187] to 36 sessions over 12 weeks.[41] The length of each session is also predetermined, some are consistent throughout the training,[36,187] while some progressively increase from 10 to 30 minutes.[38] However, it might be difficult to standardize training in the real world. The training frequency and length of each session would depend on the runner's regular training routine. Structured lab-based training allows researchers to objectively assess the effectiveness of one standardized training protocol in a group of runners based on biomechanical outcomes, but this may not be necessary or practical in real-world training. The duration of the training should be determined on an as-needed basis.[188] Training in the real world could make use of continuous monitoring, assessing the performance within sessions, and allowing adjustments to the overall duration of training. In fact, in a recent gait retraining study, Goss *et al.* introduced checkpoints at six, eight and 10 weeks to assess the modification.[189] Gait assessments were conducted after the first six weeks of training, and runners with an incomplete RFS to non-RFS transition were asked to continue training for another 2 weeks before reassessment. Among the 19 runners who participated, 79% transitioned within six weeks, and another 5% and 11% transitioned within eight and 10 weeks. The rate of motor learning could vary between runners of a different age or skill level. Real-world training should incorporate more frequent assessments and/or continuous monitoring, and adjust the training protocol based on individual needs.[188] Clansey *et al.* found that VALR and VILR in some trained runners trended back to their baseline value after a month and proposed the need for additional sessions (i.e., refresher sessions) to promote retention.[34] Continuous monitoring could detect changes in the running gait post-training and indicate if refresher sessions are required.

Monitoring performance within training sessions could also help the early detection of non-responders. Runners respond differently to training.[24,31,40,44] In studies that promote non-RFS in habitual RFS runners, including studies 1 and 2, the response rate ranged from 40 to 75%.[31] Sheerin *et al.* and Zhang *et al.* reported the response rate of 61 and 80% for their gait retraining to reduce PTA.[40,44] For runners training by themselves in the real world, it would save both effort and time if they could be advised of alternative training targets once identified as a non-responder. Besides, monitoring performance within sessions might also promote progressive learning. Lab-based gait retraining studies often use a fixed training target (i.e., increasing cadence by 7.5%,[41,42] reducing PTA to 50 or 80% of baseline[32,35,44]). Training targets could be flexible and adjusted based on previous performance to ensure a challenging yet attainable goal.[40,188] Runners are not supervised in real-world training, and they are not prescribed structured training. A challenging and attainable goal might help keep runners motivated and enhance compliance.[190]

The training set-up for real-world conditions should be simple, and equipment should be accessible to the general running population. While not commercially available, the manufacturing cost of the foot-strike detection insole used for training in Study 2 is low,[70] and the feedback system set-up was simple. A similar system also exists within the market.[191] Recent gait retraining studies have used commercially available foot-pod and a smartwatch to provide cadence information,[42] a simple metronome application on the smartphone to cue runners to run at the targeted cadence[41] and instrumented socks with textile pressure sensors to provide real-time foot-strike pattern and cadence information.[189] Runners in these studies were given simple instructions on the set-up before their first training and their training sessions were unsupervised, with no researcher to assist in the set-up. These applications have demonstrated the possibility of using wearable systems to provide biofeedback for training in the real world, and simple systems with intuitive interfaces are preferred.

Common types of feedback modality for biofeedback gait retraining include visual, audio and haptic. Visual display through a screen placed in front of a treadmill is one the most popular form of visual delivery within the lab.[192] Study 1 displayed foot-strike pattern information to runners using this method. Other studies have displayed real-time GRF curves,[14] hip adduction angle curves,[193] and tibial acceleration curves.[35,44,194] Visual display allows for more complex information, yet screens might not be possible in overground training. The use of audio feedback might be preferred when running in real-world conditions. Audio feedback allows better focus on the task compared to visual feedback.[33,195,196] Runners training in the real world would be able to receive feedback without losing focus on the road condition. In Study 2, foot-strike information was represented by audio cues of a different pitch. Compared to using a graphical presentation of the location of the foot that strikes the ground at initial contact in Study 1, the audio cue in Study 2 might be less intuitive. Runners have to associate a certain pitch with the corresponding foot-strike pattern, fortunately, this process did not seem to affect the training effect. Audio cues as feedback should be simple and informative, for example, a simple "beep" or buzzer noise could be used when performance differs from the training target,[129,147] alerting the runner to focus on the modification. Another type of audio cue like metronomes[38,41] or music with beat frequency synced to the target cadence[151] can also be used to assist runners in changing their cadence. Haptic feedback, often in the form of vibrations, was seldom used in running gait retraining. Sheerin *et al.* used a haptic feedback watch, providing vibration pulses when PTA exceeded the training target.[40] While it is possible to use haptic feedback in real-world training, the feedback should be easy to perceive and the vibrations should not hinder movement.[197]

In treadmill-based studies, runners would train at a constant speed. The speed was usually determined by the runner[14,38] or picked between a few choices.[40] In study 1, the speed was set to match a typical 30-minute training session. Speed control is simple on treadmills, overground supervised training would require a positioning system and/or feedback from researchers.[37] In study 2, a researcher timed each lap during the training sessions and asked the runner to adjust their speed accordingly. While speed can be controlled in overground running, the reason for such a design was unclear. One plausible explanation is to allow comparable results to treadmill-based studies. Brake *et al.* let runners train at their own pace, and even though they asked them to run roughly to the same speed as the baseline, they found a slight reduction in speed after the training.[151] In real-world overground training, speed control might lack practical significance, runners should be able to adjust their speed within the training. Speed can be provided as feedback,[41,42] but not necessarily controlled.

Lastly, there are external factors within real-world running conditions that may not be considered when training in a controlled environment, such as weather, running surface and surface inclination. A study has shown subject-specific biomechanical changes under different weather conditions (-10ºC *vs*. +6ºC).[198] Studies have reported differences in PTA along different overground running surfaces.[95,100,117] When compared to lab runways, axial PTA and resultant PTA were higher when running on grass and sidewalk.[95] The other study observed higher axial PTA in grass than concrete at the level condition, and higher axial PTA in grass than asphalt at the incline condition.[117] Difference in foot-strike pattern has also been observed between slope conditions.[107,108] In Study 2, we observed a RFS to MFS transition among some runners during uphill running. In Study 3, we observed changes in speed and cadence when runners were running uphill and downhill, compared to a level surface. These natural biomechanical adaptations should be considered when designing a training protocol. The training target should be adapted to the running condition to ensure that the feedback is relevant during real-world training. A proof-of-study has been designed to explore the potential of using adaptive feedback in gait retraining; this study is presented in section 8.2.

8.2 A proof-of-concept study

8.2.1 Rationale and objective

The training target of biofeedback gait retraining should be relevant to the running condition. Runners demonstrate grade-specific biomechanical adaptations, including changes in speed, cadence and PTA. A target set based on level running may not be suitable for training along slopes. We propose the use of adaptive feedback, with the target set based on each slope condition. As a proof-of-concept, this study was designed to establish the feasibility of adaptive feedback in a structured gait retraining to lower PTA along an outdoor route with elevation changes. The axial and resultant PTA before and after the training were compared on various slope conditions.

8.2.2 Methods

An experienced female runner with no musculoskeletal injury within 6 months prior to participation completed this study. She trained 30 – 40 km per week and was a self-reported MFS/FFS runner. She was given a detailed explanation of the experiment prior to signing an informed consent approved by the Conjoint Health Research Ethics Board of The University of Calgary (REB22-0931).

The participant completed an initial screening session at the 200-m track located in the Fitness Centre at the University of Calgary. A self-paced run at her regular training speed confirmed her average axial PTA to be higher than 8 *g* and eligible for training.[44]

The baseline assessment was separated into two parts, the outdoor field-based assessment and the treadmill assessment. The participant wore the same pair of shoes throughout the entire experiment.

To establish a stable running pattern under sloped running conditions,[159] the participant completed three outdoor field-based tests. The tests were completed between 8 to 10 AM within eight days, with at least 24 hours in between tests. Each testing session included a warm-up on a level surface and a 5-km trail run along the Weaselhead regional pathway, Calgary. The runner was asked to run at a pace that was suitable for a 5-km run. The elevation profile and the details of the route are provided in Appendix VI. The route was divided into a 1600-m level, two 800-m uphill and two 800-m downhill sections. An IMU (± 200-*g*, Blue Trident IMU, Vicon, New Zealand) was strapped to the anteromedial aspect of the right distal tibia of the participant, with the y-axis aligned to the long axis of the tibia. [50] Tri-axial accelerations were recorded simultaneously at 1,600 Hz. The participant was also fitted with a GPS-enabled smartwatch (Garmin, vívoactive® HR; Garmin International Inc, KS, United States) to record elevation and distance at 1 Hz.

Following the outdoor field-based assessment, the runner completed a lab-based treadmill assessment on a separate day. The instrumented treadmill (Bertec, OH, United States) inside the Human Performance Laboratory at the University of Calgary was used. The IMU was fitted and strapped to record tri-axial acceleration. There were three inclination conditions matching the outdoor running tests: level (0%), uphill (+4%) and downhill (-4%). The speed for each condition was set to match the average speed recorded for each slope condition (level, uphill and downhill) during the outdoor tests. Data were recorded for three minutes.

The participant completed six training sessions over two weeks, with at least 48 hours in between sessions. The training was conducted along the same route as that of the outdoor assessment, but instead of running in a loop, the runner was first trained along the 1600-m level section, then repeated the uphill and downhill segments three times each. The IMU device was strapped to the participant's right leg and connected wirelessly to her own iOS device through the Capture.U app (Vicon, New Zealand). The axial PTA collected for the level, uphill and downhill sections during the baseline field-based tests were averaged. The threshold values for retraining under each condition were set at 80% of the corresponding baseline values.[33,44] An audio beep was provided through headphones in real-time when the training threshold was exceeded. The participant was told to land softer and avoid the audio cue. Tri-axial accelerations were measured and recorded at 500 Hz.

After the training, the outdoor field-based assessment and the treadmill assessment were repeated. The treadmill assessment was conducted once at the speed used during the baseline assessment and once at the speed that matched the outdoor runs after training.

The IMU and Garmin smartwatch data were synchronized to the nearest second. Tibial acceleration was filtered and processed using identical MATLAB scripts from Study 5 to extract axial and resultant PTA. Distance and elevation were extracted from the Garmin device and smoothed with a 10 s moving average.[159] Speed was calculated using the elevation change and horizontal distance traveled over time for each section. The position data (latitude and longitude) from the Garmin device were used to identify segments of the data to be removed from the analysis. The runner crossed a bridge during the level section of the outdoor run and did a U-turn at the half-way point, data within 15 seconds prior to and after the bridge and the U-turn were removed from analysis. For each condition, 200 steps were taken from the middle and averaged.

The change in speed was analyzed using the method described in Brake *et al.*'s study for within-subject changes across time.[151] A 2-SD band was calculated using the average speed under each slope condition recorded during the three field-based assessments at baseline. The average speeds for each slope condition during training and post-training assessment sessions were calculated. Two or more successive data points (i.e., average speed) outside of the 2-SD band indicate a significant change from baseline.[151]

The reliable change index (RCI)[44] was calculated for axial and resultant PTA between baseline and post-training assessments for each sloped condition using the following equation:

$$RCI = \frac{PTA_1 - PTA_2}{\sqrt{2 \times \left(SD_1 \times \sqrt{1 - r_{xx}}\right)^2}}$$

where PTA_1 and PTA_2 represent the average PTA value measured during the baseline and post-training assessments for each condition; SD_1 represents the SD in the baseline assessment. r_{xx} represents the reliability coefficient. The reliability coefficients for each condition are listed in Table 8.1. The RCI has been used to assess the effect of gait retraining on an individual level,[44] a value greater than 1.96 indicates a significant change with 95% confidence.

Condition	Variable	Reliability coefficient (r_{xx})
Outdoor field-based	Axial PTA	0.81+
	Resultant PTA	0.93+
Treadmill	Axial PTA	0.81[171]
	Resultant PTA	0.92[47]

+ reliability coefficient obtained in Study 4.

Table 8.1. Reliability coefficients used for the calculation of reliable change index for different conditions and variables.

8.2.3 Results

At baseline, the running speed for level, uphill and downhill conditions were 3.62 ± 0.13, 3.47 ± 0.08 and 3.82 ± 0.07 m/s respectively. Slower running speeds were recorded during the post-training field-based assessment, the adjusted speeds were 2.96 ± 0.05, 3.08 ± 0.08 and 3.10 ± 0.14 m/s. The average speed recorded during the six training sessions and the post-training assessment were all outside of the 2-SD band. A significant reduction in speed was observed during training sessions and during the post-training assessment for all sloped conditions (Figure 8.1).

The RCIs are presented in Table 8.2. Significant changes in axial PTA were observed in all slope conditions, for field-based (RCI > 2.4) and treadmill (RCI > 3.2) assessments, when running at the adjusted speed. The largest change in axial PTA was observed during field-based downhill running with a 57% (-9.17*g*) reduction. However, when running on the treadmill at the baseline speed (i.e., faster speed), no significant reduction was observed in axial and resultant PTA at the level and uphill conditions. The only significant change was observed in the downhill condition (RCI = 4.0) with a 15.7% (-1.9 *g*) reduction in axial PTA.

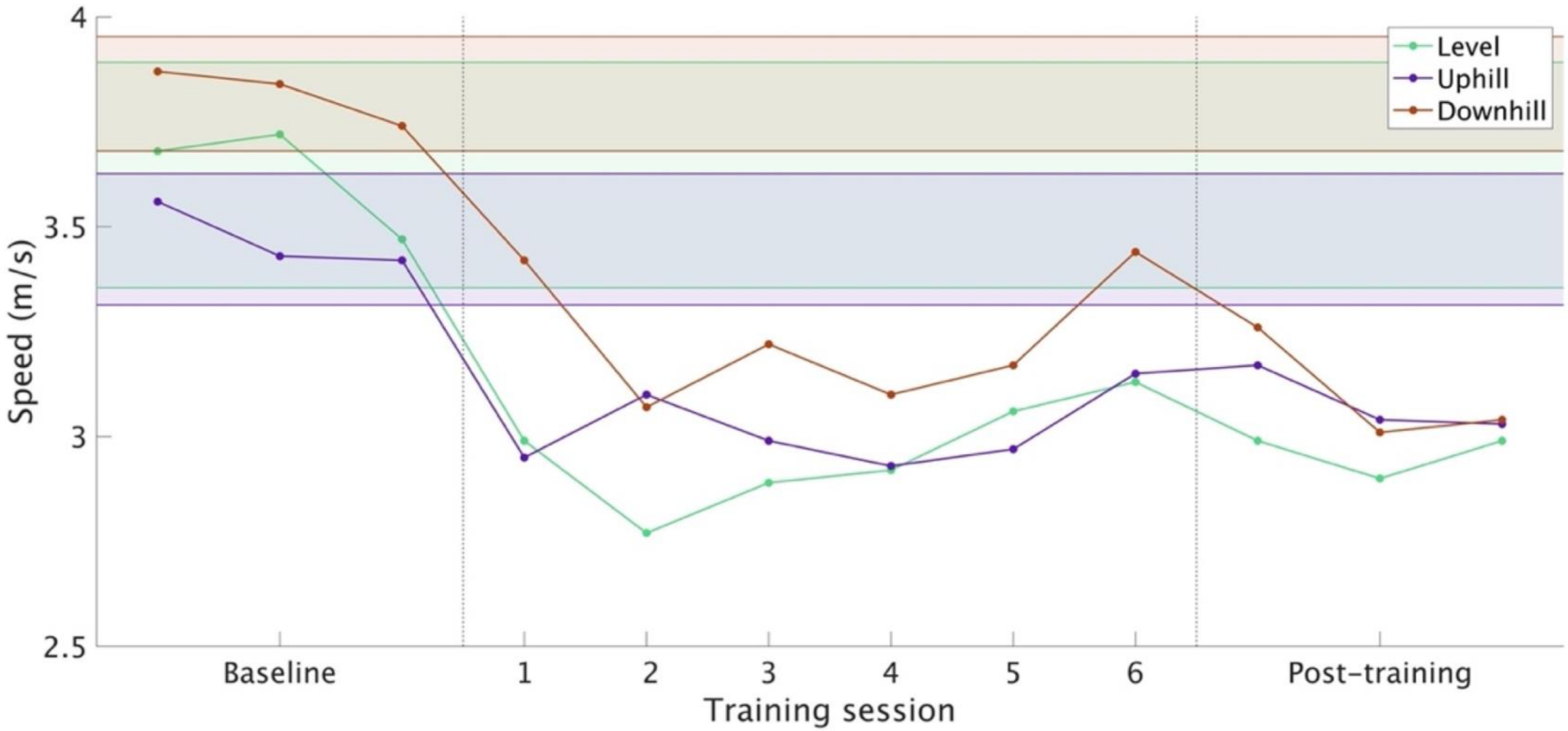


Figure 8.1. Average speed for the level, uphill and downhill conditions for each session.

Data points are color coded to represent different slope conditions. Shaded regions in corresponding colors represent the two standard deviation band calculated using the average speed under of each slope condition at baseline.

		Baseline PTA (g)	Post-training PTA (g)		RCI	
			Adjusted Speed	Same Speed	Adjusted Speed	Same Speed
Outdoor field-based						
Axial	Level	8.56 ± 0.13	5.31 ±0.12		5.14*	
	Uphill	6.33 ± 0.41	4.10 ± 0.16		2.42*	
	Downhill	16.20 ± 0.15	7.04 ± 0.35		5.83*	
Resultant	Level	18.16 ± 3.06	14.05 ± 0.33		3.76*	
	Uphill	18.70 ± 1.66	14.65 ± 0.50		3.85*	
	Downhill	35.46 ± 5.25	18.62 ± 0.44		4.18*	
Treadmill						
Axial	Level	7.90 ± 0.74	5.70 ± 0.47	7.71 ± 0.58	5.86*	0.49
	Uphill	5.19 ± 0.78	3.90 ± 0.43	5.16 ± 0.46	3.22*	0.07
	Downhill	11.93 ± 0.92	7.19 ± 0.65	10.06 ± 0.85	10.05*	3.97*
Resultant	Level	17.17 ± 0.85	13.32 ± 1.02	16.77 ± 1.12	11.38*	1.18
	Uphill	14.42 ± 1.06	13.28 ± 1.13	14.58 ± 1.20	2.69*	-0.36
	Downhill	22.84 ± 1.03	18.71 ± 1.13	23.12 ± 1.12	10.02*	-0.67

PTA, peak tibial acceleration; g (gravitational constant, g = 9.81 m/s^2); RCI, reliable change index.
* significant difference (RCI > 1.96) between baseline and post-training assessments

Table 8.2. Mean ± standard deviation of axial and resultant PTA at baseline and post-training assessments.

8.2.4 Discussion

This proof-of-concept study demonstrated the feasibility of using adaptive feedback, training targets set based on different slope conditions (i.e., level, uphill and downhill), in outdoor gait retraining. Reductions in axial and resultant PTA were observed in all slope conditions. Running speed was reduced during the training sessions and during the assessment sessions after the training.

Several gait retraining studies have adopted a free-form training protocol, where runners completed the biofeedback gait retraining within their usual training conditions.[39,41,42,151] These studies were designed to increase cadence, where the training target and feedback were set as +7.5 – 10% of the baseline value. While the exact training conditions were not reported in these studies, their participants could have trained on treadmills, overground running tracks or even trails with elevation changes. The training target remained constant during the whole training, regardless of slope conditions. None of the mentioned studies tested the training effects on slopes,[39,41,42,151] therefore it is unsure if using a constant training target was effective for sloped running. However, based on our understanding of natural gait adaptations in sloped running, as shown in previous studies[46,105] and Study 3, running speed, cadence and PTA were different across slope conditions. Using a training target set for running along a level surface might not be suitable for training along slopes. To ensure relevant feedback during training, we proposed the use of adaptive feedback, with the training targets set based on the baseline value obtained from each condition. To our knowledge, this proof-of-concept study is the first to use a different training target for different slope conditions during gait retraining. We tested this protocol along an overground running route with changes in elevation.

Although the training sessions in this study were supervised, it has demonstrated the feasibility of using a commercially available system to conduct gait retraining to reduce

PTA under real-world conditions. This wearable system was chosen for its ease of use, and it meets the technical requirements as recommended in Study 5. Before each training session, the participant fixed the sensor onto her right lower leg and connected the sensor to her own hand-held device. The system is commercially available, and the setup for providing audio feedback is simple. The participant did not encounter any technical difficulties during the training sessions.

The acceleration signals during the training sessions were recorded. As mentioned previously in section 8.1.3, this information could be used to identify non-responders and determine if the runner requires more training sessions.[188] The axial PTA values obtained from the data collected during her training sessions were processed and presented in Figure 8.2. Using data obtained during the training session, metrics such as the mean, median or upper/lower quartile PTA, and percentage of strides meeting the training target are available and could be used to determine if the training is complete based on pre-set criteria. This proof-of-concept study has demonstrated the feasibility of obtaining training data. Future research can utilize training data from a larger group of participants to define the criteria for the detection of non-responders and to determine if more sessions are required based on an individual's training performance.

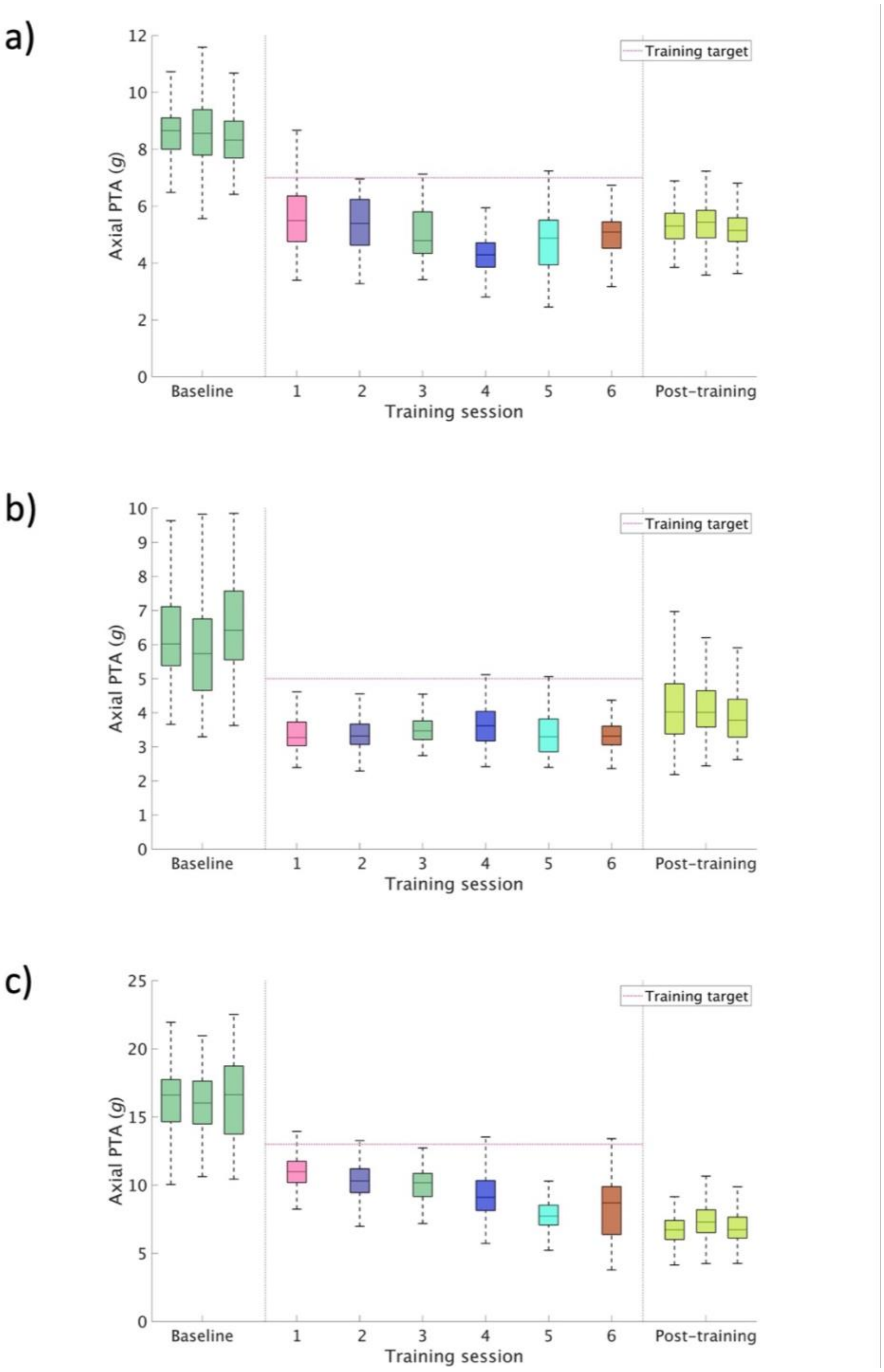


Figure 8.2. Box plots of axial peak tibial acceleration (PTA) during a) level, b) uphill and c) downhill conditions across sessions.

PTA is presented in unit of *g* (gravitational constant, $g = 9.81$ m/s^2).

In this study, we observed a reduction in running speed during and after the training. This observation suggested that when speed was not strictly controlled during training, as in treadmill-based training, the runner make adjustments to their speed. A similar observation was reported in Brake *et al.*'s study.[151] In the current study, the runner completed the post-training treadmill assessment at both the baseline speed and the adjusted speed. While a significant reduction in PTA was observed in all conditions at the adjusted speed, only axial PTA during downhill running was reduced at the baseline speed. Reducing speed appears to be the main strategy used by this runner to reduce axial PTA across conditions, and other subsidiary changes might only be carried-over to the downhill condition. There may also be a floor effect on the ability to reduce PTA on the treadmill at the baseline speed with lower baseline axial PTA ($< 8\ g$).[44] A larger study with kinematic data is required to confirm these speculations. Furthermore, continuous remote monitoring of the runner is warranted. If the runner's regular training speed trends back to the baseline value, additional training sessions with speed control might be needed.

8.3 Limitations and future directions

The proof-of-concept study is designed to explore the potential of gait retraining in conditions that resemble real-world training. The proposed training protocol was tested to demonstrate its feasibility and to provide information for future large-scale studies. One limitation is that we have only considered three slope conditions. Runners training in the real world can run across a wide range of gradients, and a subtle uphill slope could affect the gait differently than a steep slope. Based on the results of Study 3, the grade-specific adaptations are expected to be subject-specific and non-linear. Future studies could assess runners on common retraining targets (i.e., cadence and PTA) across a range of gradients to create a subject-specific model. Real-time position (i.e., GPS data) obtained from wearable devices could be used during real-world training to obtain the exact gradient, and an appropriate threshold could be set based on the subject-specific model developed.

The runner in this proof-of-concept study reduced her speed during and after the training. While reducing speed appears to be a viable strategy to reduce PTA during overground gait retraining, this strategy may not be preferred or adopted by competitive runners who prioritize their performance. Real-time visual[42] or audio feedback[189] on speed can be provided to runners who would prefer to maintain their speed during training. Future investigations are required to examine the effect of speed-controlled overground training along slopes.

The focus of this study was on modifying gait in real-world conditions. While there are theoretical links between biomechanical changes and injury risk,[12,17] large-scale RCTs which analyze injury prevalence after training are needed to fully comprehend the clinical implications of gait retraining in real-world conditions.

Lastly, this proof-of-concept study focused on gait training along overground slopes. Compared to lab-based training, our training was conducted in an environment that better resembles the natural training conditions of the general running population. And yet, the training was conducted within two weeks and the running course was controlled. There are other external factors within real-world training that could interact with the training effect, such as weather and change in running surface. Future studies should assess these factors to further optimize the training protocol for use in real-world conditions.

8.4 Conclusion

This thesis presented the limitations of lab-based gait retraining. The training effect of lab-based training was not fully transferred to conditions that resemble runners' natural training conditions, including overground and sloped running. The results of this thesis support the need for training in conditions that matches real-world conditions. Furthermore, runners demonstrated biomechanical adaptations when running along slopes in the real world. External factors within real-world running conditions should be considered when optimizing the training protocol.

Gait retraining under real-world conditions would require systems that can provide feedback outside of the lab. With the existing wearable technology, tibial acceleration, a common training parameter, can be measured and real-time biofeedback can be provided to runners. Some technical specifications should be considered for accurate and reliable tibial acceleration measurements, including the use of accelerometers with an operating range wider than ±16-*g*, and at least 100 strides should be measured during each session.

REFERENCE

1. Hulteen RM, Smith JJ, Morgan PJ, et al. Global participation in sport and leisure-time physical activities: A systematic review and meta-analysis. *Prev Med*. 2017;95:14-25. doi:10.1016/j.ypmed.2016.11.027

2. Messier SP, Martin DF, Mihalko SL, et al. A 2-Year Prospective Cohort Study of Overuse Running Injuries: The Runners and Injury Longitudinal Study (TRAILS). *Am J Sports Med*. 2018;46(9):2211-2221. doi:10.1177/0363546518773755

3. van Gent RN, Siem D, van Middelkoop M, et al. Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. *Br J Sports Med*. 2007;41(8):469-480. doi:10.1136/bjsm.2006.033548

4. Davis IS, Bowser BJ, Mullineaux DR. Greater vertical impact loading in female runners with medically diagnosed injuries: a prospective investigation. *Br J Sports Med*. 2016;50(14):887-892. doi:10.1136/bjsports-2015-094579

5. Napier C, Cochrane CK, Taunton JE, Hunt MA. Gait modifications to change lower extremity gait biomechanics in runners: a systematic review. *Br J Sports Med*. 2015;49(21):1382-1388. doi:10.1136/bjsports-2014-094393

6. Davis IS, Futrell E. Gait Retraining: Altering the Fingerprint of Gait. *Phys Med Rehabil Clin N Am*. 2016;27(1):339-355. doi:10.1016/j.pmr.2015.09.002

7. Agresta C, Brown A. Gait Retraining for Injured and Healthy Runners Using Augmented Feedback: A Systematic Literature Review. *J Orthop Sports Phys Ther*. 2015;45(8):576-584. doi:10.2519/jospt.2015.5823

8. Giggins OM, Persson UM, Caulfield B. Biofeedback in rehabilitation. *Journal Neuroeng Rehabil*. 2013;10(1):60. doi:10.1186/1743-0003-10-60

9. Napier C, Esculier JF, Hunt MA. Gait retraining: out of the lab and onto the streets with the benefit of wearables. *Br J Sports Med*. 2017;51(23):1642-1643. doi:10.1136/bjsports-2017-098637

10. Winstein CJ. Knowledge of results and motor learning--implications for physical therapy. *Phys Ther*. 1991;71(2):140-149.

11. Hodges NJ, Williams AM. Understanding the role of augmented feedback: The good, the bad and the ugly. In: *Skill Acquisition in Sport*. Routledge; 2004:145-168. doi:10.4324/9780203646564-13

12. Doyle E, Doyle TLA, Bonacci J, Fuller JT. The Effectiveness of Gait Retraining on Running Kinematics, Kinetics, Performance, Pain, and Injury in Distance Runners: A Systematic Review With Meta-analysis. *J Orthop Sports Phys Ther*. 2022;52(4):192-A5. doi:10.2519/jospt.2022.10585

13. Barton CJ, Bonanno DR, Carr J, et al. Running retraining to treat lower limb injuries: a mixed-methods study of current evidence synthesised with expert opinion. *Br J Sports Med*. 2016;50(9):513-526. doi:10.1136/bjsports-2015-095278

14. Chan ZYS, Zhang JH, Au IPH, et al. Gait Retraining for the Reduction of Injury Occurrence in Novice Distance Runners: 1-Year Follow-up of a Randomized Controlled Trial. *Am J Sports Med*. 2018;46(2):388-395. doi:10.1177/0363546517736277

15. Letafatkar A, Rabiei P, Farivar N, Alamouti G. Long-term efficacy of conditioning training program combined with feedback on kinetics and kinematics in male runners. *Scand J Med Sci Sports*. 2020;30(3):429-441. doi:10.1111/sms.13587

16. Jauhiainen S, Pohl AJ, Äyrämö S, Kauppi JP, Ferber R. A hierarchical cluster analysis to determine whether injured runners exhibit similar kinematic gait patterns. *Scand J Med Sci Sports*. 2020;30(4):732-740. doi:10.1111/sms.13624

17. Willwacher S, Kurz M, Robbin J, et al. Running-Related Biomechanical Risk Factors for Overuse Injuries in Distance Runners: A Systematic Review Considering Injury Specificity and the Potentials for Future Research. *Sports Med*. 2022;52(8):1863-1877. doi:10.1007/s40279-022-01666-3

18. Raghunandan A, Charnoff JN, Matsuwaka ST. The Epidemiology, Risk Factors, and Nonsurgical Treatment of Injuries Related to Endurance Running. *Curr Sports Med Rep*. 2021;20(6):306-311. doi:10.1249/JSR.0000000000000852

19. Kakouris N, Yener N, Fong DTP. A systematic review of running-related musculoskeletal injuries in runners. *J Sport Health Sci*. 2021;10(5):513-522. doi:10.1016/j.jshs.2021.04.001

20. Gaudette LW, Bradach MM, de Souza Junior JR, et al. Clinical Application of Gait Retraining in the Injured Runner. *J Clin Med*. 2022;11(21):6497. doi:10.3390/jcm11216497

21. Milner CE, Ferber R, Pollard CD, Hamill J, Davis IS. Biomechanical Factors Associated with Tibial Stress Fracture in Female Runners. *Med Sci Sports Exerc*. 2006;38(2):323-328. doi:10.1249/01.mss.0000183477.75808.92

22. Futrell EE, Jamison ST, Tenforde AS, Davis IS. Relationships between Habitual Cadence, Footstrike, and Vertical Loadrates in Runners. *Med Sci Sports Exerc*. 2018;50(9):1837-1841. doi:10.1249/MSS.0000000000001629

23. Schmida EA, Wille CM, Stiffler-Joachim MR, Kliethermes SA, Heiderscheit BC. Vertical Loading Rate Is Not Associated with Running Injury, Regardless of Calculation Method. *Med Sci Sports Exerc*. 2022;54(8):1382-1388. doi:10.1249/MSS.0000000000002917

24. Crowell HP, Milner CE, Hamill J, Davis IS. Reducing Impact Loading During Running With the Use of Real-Time Visual Feedback. *J Orthop Sports Phys Ther*. 2010;40(4):206-213. doi:10.2519/jospt.2010.3166

25. Johnson CD, Tenforde AS, Outerleys J, Reilly J, Davis IS. Impact-Related Ground Reaction Forces Are More Strongly Associated With Some Running Injuries Than Others. *Am J Sports Med*. 2020;48(12):3072-3080. doi:10.1177/0363546520950731

26. Zifchock RA, Davis I, Hamill J. Kinetic asymmetry in female runners with and without retrospective tibial stress fractures. *J Biomech*. 2006;39(15):2792-2797. doi:10.1016/j.jbiomech.2005.10.003

27. Pohl MB, Mullineaux DR, Milner CE, Hamill J, Davis IS. Biomechanical predictors of retrospective tibial stress fractures in runners. *J Biomech*. 2008;41(6):1160-1165. doi:10.1016/j.jbiomech.2008.02.001

28. Pohl MB, Hamill J, Davis IS. Biomechanical and Anatomic Factors Associated with a History of Plantar Fasciitis in Female Runners. *Clin J Sport Med*. 2009;19(5):372-376. doi:10.1097/JSM.0b013e3181b8c270

29. Ribeiro AP, Sacco ICN, Dinato RC, João SMA. Relationships between static foot alignment and dynamic plantar loads in runners with acute and chronic stages of plantar fasciitis: a cross-sectional study. *Braz J Phys Ther*. 2016;20(1):87-95. doi:10.1590/bjpt-rbf.2014.0136

30. Bowser BJ, Fellin R, Milner CE, Pohl MB, Davis IS. Reducing Impact Loading in Runners: A One-Year Follow-up. *Med Sci Sports Exerc*. 2018;50(12):2500-2506. doi:10.1249/MSS.0000000000001710

31. Chan ZYS, Zhang JH, Ferber R, Shum G, Cheung RTH. The effects of midfoot strike gait retraining on impact loading and joint stiffness. *Phys Ther Sport*. 2020;42:139-145. doi:10.1016/j.ptsp.2020.01.011

32. Cheung RTH, An WW, Au IPH, Zhang JH, Chan ZYS, MacPhail AJ. Control of impact loading during distracted running before and after gait retraining in runners. *J Sports Sci*. 2018;36(13):1497-1501. doi:10.1080/02640414.2017.1398886

33. Ching E, An WW, Au IPH, et al. Impact Loading During Distracted Running Before and After Auditory Gait Retraining. *Int J Sports Med*. 2018;39(14):1075-1080. doi:10.1055/a-0667-9875

34. Clansey AC, Hanlon M, Wallace ES, Nevill A, Lake MJ. Influence of tibial shock feedback training on impact loading and running economy. *Med Sci Sports Exerc*. 2014;46(5):973-981. doi:10.1249/MSS.0000000000000182

35. Crowell HP, Davis IS. Gait retraining to reduce lower extremity loading in runners. *Clin Biomech*. 2011;26(1):78-83. doi:10.1016/j.clinbiomech.2010.09.003

36. Derie R, Van den Berghe P, Gerlo J, et al. Biomechanical adaptations following a music-based biofeedback gait retraining program to reduce peak tibial accelerations. *Scand J Med Sci Sports*. 2022;32(7):1142-1152. doi:10.1111/sms.14162

37. Van den Berghe P, Derie R, Bauwens P, et al. Reducing the peak tibial acceleration of running by music-based biofeedback: A quasi-randomized controlled trial. *Scand J Med Sci Sports*. 2022;32(4):698-709. doi:10.1111/sms.14123

38. Futrell EE, Gross KD, Reisman D, Mullineaux DR, Davis IS. Transition to forefoot strike reduces load rates more effectively than altered cadence. *J Sport Health Sci*. 2020;9(3):248-257. doi:10.1016/j.jshs.2019.07.006

39. Hafer JF, Brown AM, deMille P, Hillstrom HJ, Garber CE. The effect of a cadence retraining protocol on running biomechanics and efficiency: a pilot study. *J Sports Sci*. 2015;33(7):724-731. doi:10.1080/02640414.2014.962573

40. Sheerin KR, Reid D, Taylor D, Besier TF. The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners. *Phys Ther Sport*. 2020;43:173-180. doi:10.1016/j.ptsp.2020.03.001

41. Wang J, Luo Z, Dai B, Fu W. Effects of 12-week cadence retraining on impact peak, load rates and lower extremity biomechanics in running. *PeerJ*. 2020;8:e9813. doi:10.7717/peerj.9813

42. Willy RW, Buchenic L, Rogacki K, Ackerman J, Schmidt A, Willson JD. In-field gait retraining and mobile monitoring to address running biomechanics associated with tibial stress fracture. *Scand J Med Sci Sports*. 2016;26(2):197-205. doi:10.1111/sms.12413

43. Yang Y, Zhang X, Luo Z, Wang X, Ye D, Fu W. Alterations in Running Biomechanics after 12 Week Gait Retraining with Minimalist Shoes. *Int J Environ Res Public Health*. 2020;17(3):818. doi:10.3390/ijerph17030818

44. Zhang JH, Chan ZYS, Au IPH, An WW, Cheung RTH. Can runners maintain a newly learned gait pattern outside a laboratory environment following gait retraining? *Gait Posture*. 2019;69:8-12. doi:10.1016/j.gaitpost.2019.01.014

45. Zhang JH, Chan ZYS, Au IPH, An WW, Shull PB, Cheung RTH. Transfer Learning Effects of Biofeedback Running Retraining in Untrained Conditions. *Med Sci Sports Exerc*. 2019;51(9):1904-1908. doi:10.1249/MSS.0000000000002007

46. Sheerin KR, Reid D, Besier TF. The measurement of tibial acceleration in runners—A review of the factors that can affect tibial acceleration during running and evidence-based guidelines for its use. *Gait Posture*. 2019;67:12-24. doi:10.1016/j.gaitpost.2018.09.017

47. Sheerin KR, Besier TF, Reid D, Hume PA. The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners. *Sports Biomech*. 2018;17(4):531-540. doi:10.1080/14763141.2017.1371214

48. Lafortune MA. Three-dimensional acceleration of the tibia during walking and running. *J Biomech*. 1991;24(10):877-886. doi:10.1016/0021-9290(91)90166-k

49. Zhang JH, An WW, Au IPH, Chen TL, Cheung RTH. Comparison of the correlations between impact loading rates and peak accelerations measured at two different body sites: Intra- and inter-subject analysis. *Gait Posture*. 2016;46:53-56. doi:10.1016/j.gaitpost.2016.02.002

50. Van den Berghe P, Six J, Gerlo J, Leman M, De Clercq D. Validity and reliability of peak tibial accelerations as real-time measure of impact loading during over-ground rearfoot running at different speeds. *J Biomech*. 2019;86:238-242. doi:10.1016/j.jbiomech.2019.01.039

51. Fortune E, Morrow MMB, Kaufman KR. Assessment of Gait Kinetics Using Tri-Axial Accelerometers. *J Appl Biomech*. 2014;30(5):668-674. doi:10.1123/jab.2014-0037

52. Laughton CA, Davis IM, Hamill J. Effect of Strike Pattern and Orthotic Intervention on Tibial Shock during Running. *J Appl Biomech*. 2003;19(2):153-168. doi:10.1123/jab.19.2.153

53. Sheerin KR, Besier TF, Reid D. The influence of running velocity on resultant tibial acceleration in runners. *Sports Biomech*. 2020;19(6):750-760. doi:10.1080/14763141.2018.1546890

54. Cheung RTH, Zhang JH, Chan ZYS, et al. Shoe-mounted accelerometers should be used with caution in gait retraining. *Scand J Med Sci Sports*. 2019;29(6):835-842. doi:10.1111/sms.13396

55. Altman AR, Davis IS. A kinematic method for footstrike pattern detection in barefoot and shod runners. *Gait Posture*. 2012;35(2):298-300. doi:10.1016/j.gaitpost.2011.09.104

56. Li X, Yu J, Bai J, et al. The Effect of Real-Time Tibial Acceleration Feedback on Running Biomechanics During Gait Retraining: A Systematic Review and Meta-Analysis. *J Sport Rehabil*. 2023;32(4):449-461. doi:10.1123/jsr.2022-0279

57. Huang Y, Xia H, Chen G, Cheng S, Cheung RTH, Shull PB. Foot strike pattern, step rate, and trunk posture combined gait modifications to reduce impact loading during running. *J Biomech*. 2019;86:102-109. doi:10.1016/j.jbiomech.2019.01.058

58. Janssen M, Scheerder J, Thibaut E, Brombacher A, Vos S. Who uses running apps and sports watches? Determinants and consumer profiles of event runners' usage of running-related smartphone applications and sports watches. Guilhem G, ed. *PLoS One*. 2017;12(7):e0181167. doi:10.1371/journal.pone.0181167

59. Clermont CA, Duffett-Leger L, Hettinga BA, Ferber R. Runners' Perspectives on 'Smart' Wearable Technology and Its Use for Preventing Injury. *Int J Hum-Comput Interact*. 2020;36(1):31-40. doi:10.1080/10447318.2019.1597575

60. Chan ZYS, Peeters R, Cheing G, Ferber R, Cheung RTH. Evaluation of COVID-19 Restrictions on Distance Runners' Training Habits Using Wearable Trackers. *Front Sports Act Living*. 2022;3:812214. doi:10.3389/fspor.2021.812214

61. Mason R, Pearson LT, Barry G, et al. Wearables for Running Gait Analysis: A Systematic Review. *Sports Med*. 2023;53(1):241-268. doi:10.1007/s40279-022-01760-6

62. Hasegawa H, Yamauchi T, Kraemer WJ. Foot strike patterns of runners at the 15-km point during an elite-level half marathon. *J Strength Cond Res*. 2007;21(3):888-893.

63. Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. *J Biomech*. 1980;13(5):397-406. doi:10.1016/0021-9290(80)90033-0

64. Cheung RTH, Wong RYL, Chung TKW, Choi RT, Leung WWY, Shek DHY. Relationship between foot strike pattern, running speed, and footwear condition in recreational distance runners. *Sports Biomech*. 2017;16(2):238-247. doi:10.1080/14763141.2016.1226381

65. Giandolini M, Poupard T, Gimenez P, et al. A simple field method to identify foot strike pattern during running. *J Biomech*. 2014;47(7):1588-1593. doi:10.1016/j.jbiomech.2014.03.002

66. Farina KA, Needle AR, van Werkhoven H. Continuous Tracking of Foot Strike Pattern during a Maximal 800-Meter Run. *Sensors*. 2021;21(17):5782. doi:10.3390/s21175782

67. van Werkhoven H, Farina KA, Langley MH. Using A Soft Conformable Foot Sensor to Measure Changes in Foot Strike Angle During Running. *Sports*. 2019;7(8):184. doi:10.3390/sports7080184

68. Shiang TY, Hsieh TY, Lee YS, et al. Determine the Foot Strike Pattern Using Inertial Sensors. *J Sens*. 2016;2016:1-6. doi:10.1155/2016/4759626

69. Mo S, Chan ZYS, Lai KKY, et al. Effect of minimalist and maximalist shoes on impact loading and footstrike pattern in habitual rearfoot strike trail runners: An in-field study. *Eur J Sport Sci*. 2021;21(2):183-191. doi:10.1080/17461391.2020.1738559

70. Cheung RTH, An WW, Au IPH, et al. Measurement agreement between a newly developed sensing insole and traditional laboratory-based method for footstrike pattern detection in runners. Gard SA, ed. *PLoS One*. 2017;12(6):e0175724. doi:10.1371/journal.pone.0175724

71. Mo S, Chow DHK. Accuracy of three methods in gait event detection during overground running. *Gait Posture*. 2018;59:93-98. doi:10.1016/j.gaitpost.2017.10.009

72. Benson L, Clermont C, Watari R, Exley T, Ferber R. Automated Accelerometer-Based Gait Event Detection During Multiple Running Conditions. *Sensors*. 2019;19(7):1483. doi:10.3390/s19071483

73. Mitschke C, Kiesewetter P, Milani TL. The Effect of the Accelerometer Operating Range on Biomechanical Parameters: Stride Length, Velocity, and Peak Tibial Acceleration during Running. *Sensors*. 2018;18(1):130. doi:10.3390/s18010130

74. Ruder M, Jamison ST, Tenforde A, Mulloy F, Davis IS. Relationship of Foot Strike Pattern and Landing Impacts during a Marathon. *Med Sci Sports Exerc*. 2019;51(10):2073-2079. doi:10.1249/MSS.0000000000002032

75. Johnson CD, Outerleys J, Jamison ST, Tenforde AS, Ruder M, Davis IS. Comparison of Tibial Shock during Treadmill and Real-World Running. *Med Sci Sports Exerc*. 2020;52(7):1557-1562. doi:10.1249/MSS.0000000000002288

76. Aubol KG, Milner CE. Minimum Sampling Frequency for Accurate and Reliable Tibial Acceleration Measurements During Rearfoot Strike Running in the Field. *J Appl Biomech*. Published online March 31, 2023:1-6. doi:10.1123/jab.2022-0069

77. Sara LK, Outerleys J, Johnson CD. The Effect of Sensor Placement on Measured Distal Tibial Accelerations During Running. *J Appl Biomech*. 2023;39(3):199-203. doi:10.1123/jab.2022-0249

78. Aubol KG, Hawkins JL, Milner CE. Tibial Acceleration Reliability and Minimal Detectable Difference During Overground and Treadmill Running. *J Appl Biomech*. 2020;36(6):457-459. doi:10.1123/jab.2019-0272

79. Van Hooren B, Fuller JT, Buckley JD, et al. Is Motorized Treadmill Running Biomechanically Comparable to Overground Running? A Systematic Review and Meta-Analysis of Cross-Over Studies. *Sports Med*. 2020;50(4):785-813. doi:10.1007/s40279-019-01237-z

80. Benson LC, Räisänen AM, Clermont CA, Ferber R. Is This the Real Life, or Is This Just Laboratory? A Scoping Review of IMU-Based Running Gait Analysis. *Sensors*. 2022;22(5):1722. doi:10.3390/s22051722

81. Catalá-Vilaplana I, Encarnación-Martínez A, Camacho-García A, Sanchis-Sanchis R, Pérez-Soriano P. Influence of surface condition and prolonged running on impact accelerations. *Sports Biomech*. Published online May 19, 2023:1-15. doi:10.1080/14763141.2023.2214519

82. DeJong AF, Fish PN, Hertel J. Running behaviors, motivations, and injury risk during the COVID-19 pandemic: A survey of 1147 runners. *PLoS One*. 2021;16(2):e0246300. doi:10.1371/journal.pone.0246300

83. Taunton JE. A prospective study of running injuries: the Vancouver Sun Run "In Training" clinics. *Br J Sports Med*. 2003;37:239-244. doi:10.1136/bjsm.37.3.239

84. Lafferty L, Wawrzyniak J, Chambers M, et al. Clinical Indoor Running Gait Analysis May Not Approximate Outdoor Running Gait Based on Novel Drone Technology. *Sports Health*. 2022;14(5):710-716. doi:10.1177/19417381211050931

85. Wank V, Frick U, Schmidtbleicher D. Kinematics and Electromyography of Lower Limb Muscles in Overground and Treadmill Running. *Int J Sports Med*. 1998;19(07):455-461. doi:10.1055/s-2007-971944

86. Chambon N, Delattre N, Guéguen N, Berton E, Rao G. Shoe drop has opposite influence on running pattern when running overground or on a treadmill. *Eur J Appl Physiol*. 2015;115(5):911-918. doi:10.1007/s00421-014-3072-x

87. Elliott BC, Blanksby BA. A cinematographic analysis of overground and treadmill running by males and females. *Med Sci Sports*. 1976;8(2):84-87.

88. Riley PO, Dicharry J, Franz J, Croce UD, Wilder RP, Kerrigan DC. A Kinematics and Kinetic Comparison of Overground and Treadmill Running. *Med Sci Sports Exerc*. 2008;40(6):1093-1100. doi:10.1249/MSS.0b013e3181677530

89. Tao H, Joyce L, Kozak B, Luiken J, Wendt N. Spatiotemporal comparison of overground and treadmill running with pressure sensor insoles in division I collegiate runners. *Int J Sports Phys Ther*. 2019;14(5):731-739.

90. Nelson RC, Dillman CJ, Lagasse P, Bickett P. Biomechanics of overground versus treadmill running. *Med Sci Sports*. 1972;4(4):233-240. doi:10.1249/00005768-197200440-00029

91. Hong Y, Wang L, Li JX, Zhou JH. Comparison of plantar loads during treadmill and overground running. *J Sci Med Sport*. 2012;15(6):554-560. doi:10.1016/j.jsams.2012.01.004

92. Kong PW, Koh TMC, Tan WCR, Wang YS. Unmatched perception of speed when running overground and on a treadmill. *Gait Posture*. 2012;36(1):46-48. doi:10.1016/j.gaitpost.2012.01.001

93. Ueberschär O, Fleckenstein D, Warschun F, Kränzler S, Walter N, Hoppe MW. Measuring biomechanical loads and asymmetries in junior elite long-distance runners through triaxial inertial sensors. *Sports Orthop Traumatol*. 2019;35(3):296-308. doi:10.1016/j.orthtr.2019.06.001

94. Dillon S, Burke A, Whyte EF, O'Connor S, Gore S, Moran KA. Are impact accelerations during treadmill running representative of those produced overground? *Gait Posture*. 2022;98:195-202. doi:10.1016/j.gaitpost.2022.09.076

95. Milner CE, Hawkins JL, Aubol KG. Tibial Acceleration during Running Is Higher in Field Testing Than Indoor Testing: *Med Sci Sports Exerc*. 2020;52(6):1361-1366. doi:10.1249/MSS.0000000000002261

96. García-Pérez JA, Pérez-Soriano P, Llana Belloch S, Lucas-Cuevas AG, Sánchez-Zuriaga D. Effects of treadmill running and fatigue on impact acceleration in distance running. *Sports Biomech*. 2014;13(3):259-266. doi:10.1080/14763141.2014.909527

97. Fu W, Fang Y, Liu DMS, Wang L, Ren S, Liu Y. Surface effects on in-shoe plantar pressure and tibial impact during running. *J Sport Health Sci*. 2015;4(4):384-390. doi:10.1016/j.jshs.2015.09.001

98. Montgomery G, Abt G, Dobson C, Smith T, Ditroilo M. Tibial impacts and muscle activation during walking, jogging and running when performed overground, and on motorised and non-motorised treadmills. *Gait Posture*. 2016;49:120-126. doi:10.1016/j.gaitpost.2016.06.037

99. Oliveira AS, Gizzi L, Ketabi S, Farina D, Kersting UG. Modular Control of Treadmill vs Overground Running. *PLoS One*. 2016;11(4):e0153307. doi:10.1371/journal.pone.0153307

100. Garcia MC, Gust G, Bazett-Jones DM. Tibial acceleration and shock attenuation while running over different surfaces in a trail environment. *J Sci Med Sport*. 2021;24(11):1161-1165. doi:10.1016/j.jsams.2021.03.006

101. Boey H, Aeles J, Schütte K, Vanwanseele B. The effect of three surface conditions, speed and running experience on vertical acceleration of the tibia during running. *Sports Biomech*. 2017;16(2):166-176. doi:10.1080/14763141.2016.1212918

102. Mileti I, Serra A, Wolf N, et al. Muscle Activation Patterns Are More Constrained and Regular in Treadmill Than in Overground Human Locomotion. *Front Bioeng Biotechnol*. 2020;8:581619. doi:10.3389/fbioe.2020.581619

103. Lindsay TR, Noakes TD, McGregor SJ. Effect of Treadmill versus Overground Running on the Structure of Variability of Stride Timing. *Percept Mot Skills*. 2014;118(2):331-346. doi:10.2466/30.26.PMS.118k18w8

104. Fohrmann D, Hamacher D, Sanchez-Alvarado A, et al. Reliability of Running Stability during Treadmill and Overground Running. *Sensors*. 2022;23(1):347. doi:10.3390/s23010347

105. Vernillo G, Giandolini M, Edwards WB, et al. Biomechanics and Physiology of Uphill and Downhill Running. *Sports Med*. 2016;47:615-629. doi:10.1007/s40279-016-0605-y

106. World Athletics - Trail Running. Trail Running. Published 2021. Accessed June 4, 2023. http://worldathletics.org/disciplines/trail-running/trail-running

107. Gottschall JS, Kram R. Ground reaction forces during downhill and uphill running. *J Biomech*. 2005;38(3):445-452. doi:10.1016/j.jbiomech.2004.04.023

108. Lussiana T, Fabre N, Hébert-Losier K, Mourot L. Effect of slope and footwear on running economy and kinematics. *Scand J Med Sci Sports*. 2013;23(4):e246-e253. doi:10.1111/sms.12057

109. Horvais N, Giandolini M. Foot strike pattern during downhill trail running. *Footwear Sci*. 2013;5(sup1):S26-S27. doi:10.1080/19424280.2013.799535

110. Giandolini M, Pavailler S, Samozino P, Morin JB, Horvais N. Foot strike pattern and impact continuous measurements during a trail running race: proof of concept in a world-class athlete. *Footwear Sci*. 2015;7(2):127-137. doi:10.1080/19424280.2015.1026944

111. Giandolini M, Horvais N, Rossi J, Millet GY, Samozino P, Morin JB. Foot strike pattern differently affects the axial and transverse components of shock acceleration and attenuation in downhill trail running. *J Biomech*. 2016;49(9):1765-1771. doi:10.1016/j.jbiomech.2016.04.001

112. Telhan G, Franz JR, Dicharry J, Wilder RP, Riley PO, Kerrigan DC. Lower Limb Joint Kinetics During Moderately Sloped Running. *J Athl Train*. 2010;45(1):16-21. doi:10.4085/1062-6050-45.1.16

113. Padulo J, Annino G, Migliaccio GM, D'ottavio S, Tihanyi J. Kinematics of running at different slopes and speeds. *J Strength Cond Res*. 2012;26(5):1331-1339. doi:10.1519/JSC.0b013e318231aafa

114. Townshend AD, Worringham CJ, Stewart IB. Spontaneous pacing during overground hill running. *Med Sci Sports Exerc*. 2010;42(1):160-169. doi:10.1249/MSS.0b013e3181af21e2

115. Chu JJ, Caldwell GE. Stiffness and Damping Response Associated with Shock Attenuation in Downhill Running. *J Appl Biomech*. 2004;20(3):291-308. doi:10.1123/jab.20.3.291

116. Hamill CL, Clarke IE, Frederick EG, Goodyear LJ, Howley ET. Effects of grade running on kinematics and impact force. *Med Sci Sports Exerc*. 1984;16(2):184.

117. Waite N, Goetschius J, Lauver JD. Effect of Grade and Surface Type on Peak Tibial Acceleration in Trained Distance Runners. *J Appl Biomech*. 2020;37(1):2-5. doi:10.1123/jab.2020-0096

118. Clarke TE, Cooper LB, Hamill CL, Clark DE. The effect of varied stride rate upon shank deceleration in running. *J Sports Sci*. 1985;3(1):41-49. doi:10.1080/02640418508729731

119. Ferber R, Noehren B, Hamill J, Davis IS. Competitive female runners with a history of iliotibial band syndrome demonstrate atypical hip and knee kinematics. *J Orthop Sports Phys Ther*. 2010;40(2):52-58. doi:10.2519/jospt.2010.3028

120. Morris JB, Goss DL, Miller EM, Davis IS. Using real-time biofeedback to alter running biomechanics: A randomized controlled trial. *Transl Sports Med*. 2020;3(1):63-71. doi:10.1002/tsm2.110

121. Chan PPK, Chan ZYS, Au IPH, Lam BMF, Lam WK, Cheung RTH. Biomechanical effects following footstrike pattern modification using wearable sensors. *J Sci Med Sport*. 2021;24(1):30-35. doi:10.1016/j.jsams.2020.05.019

122. Zhang JH, McPhail AJC, An WW, et al. A new footwear technology to promote non-heelstrike landing and enhance running performance: Fact or fad? *J Sports Sci*. 2017;35(15):1533-1537. doi:10.1080/02640414.2016.1224915

123. Alvim F, Cerqueira L, Netto AD, Leite G, Muniz A. Comparison of Five Kinematic-Based Identification Methods of Foot Contact Events During Treadmill Walking and Running at Different Speeds. *J Appl Biomech*. 2015;31(5):383-388. doi:10.1123/jab.2014-0178

124. Blackmore T, Willy RW, Creaby MW. The high frequency component of the vertical ground reaction force is a valid surrogate measure of the impact peak. *J Biomech*. 2016;49(3):479-483. doi:10.1016/j.jbiomech.2015.12.019

125. Cohen J. *Statistical Power Analysis for the Behavioral Sciences*. 2nd ed. L. Erlbaum Associates; 1988.

126. Lai YJ, Chou W, Chu IH, et al. Will the Foot Strike Pattern Change at Different Running Speeds with or without Wearing Shoes? *Int J Environ Res Public Health*. 2020;17(17):6044. doi:10.3390/ijerph17176044

127. Santos AF dos, Nakagawa TH, Nakashima GY, Maciel CD, Serrão F. The Effects of Forefoot Striking, Increasing Step Rate, and Forward Trunk Lean Running on Trunk and Lower Limb Kinematics and Comfort. *Int J Sports Med*. 2016;37(5):369-373. doi:10.1055/s-0035-1564173

128. Au IPH, Ng L, Davey P, et al. Impact Sound Across Rearfoot, Midfoot, and Forefoot Strike During Overground Running. *J Athl Train*. 2021;56(12):1362-1366. doi:10.4085/1062-6050-0708.20

129. Cheung RTH, Davis IS. Landing Pattern Modification to Improve Patellofemoral Pain in Runners: A Case Series. *J Orthop Sports Phys Ther*. 2011;41(12):914-919. doi:10.2519/jospt.2011.3771

130. Lin JZ, Chiu WY, Tai WH, Hong YX, Chen CY. Ankle Muscle Activations during Different Foot-Strike Patterns in Running. *Sensors*. 2021;21(10):3422. doi:10.3390/s21103422

131. Wei Z, Zhang Z, Jiang J, Zhang Y, Wang L. Comparison of plantar loads among runners with different strike patterns. *J Sports Sci*. 2019;37(18):2152-2158. doi:10.1080/02640414.2019.1623990

132. Nishida K, Hagio S, Kibushi B, Moritani T, Kouzaki M. Comparison of muscle synergies for running between different foot strike patterns. Alway SE, ed. *PLOS ONE*. 2017;12(2):e0171535. doi:10.1371/journal.pone.0171535

133. Yong JR, Silder A, Delp SL. Differences in muscle activity between natural forefoot and rearfoot strikers during running. *J Biomech*. 2014;47(15):3593-3597. doi:10.1016/j.jbiomech.2014.10.015

134. Ahn AN, Brayton C, Bhatia T, Martin P. Muscle activity and kinematics of forefoot and rearfoot strike runners. *J Sport Health Sci*. 2014;3(2):102-112. doi:10.1016/j.jshs.2014.03.007

135. Darendeli A, Ertan H, Enoka RM. Comparison of EMG Activity in Leg Muscles between Overground and Treadmill Running. *Med Sci Sports Exerc*. 2023;55(3):517. doi:10.1249/MSS.0000000000003055

136. Dixon SJ, Collop AC, Batt ME. Surface effects on ground reaction forces and lower extremity kinematics in running. *Med Sci Sports Exerc*. 2000;32(11):1919-1926.

137. Goss DL, Gross MT. Relationships among self-reported shoe type, footstrike pattern, and injury incidence. *US Army Med Dep J*. 2012;2012 Oct-Dec:25-30.

138. Davis IS, Chen TLW, Wearing SC. Reversing the Mismatch With Forefoot Striking to Reduce Running Injuries. *Front Sports Act Living*. 2022;4:794005. doi:10.3389/fspor.2022.794005

139. Schmidt RA, Lee TD. *Motor Learning and Performance: From Principles to Application*. Fifth edition. Human Kinetics; 2014.

140. *A Special Report on Trail Running*. Outdoor Industry Foundation; 2010:1-12. Accessed January 13, 2017. http://www.outdoorfoundation.org/research.trailrunning.html

141. ITRA Discover Trail Running. Accessed May 25, 2022. https://itra.run/About/DiscoverTrailRunning

142. Vincent HK, Brownstein M, Vincent KR. Injury Prevention, Safe Training Techniques, Rehabilitation, and Return to Sport in Trail Runners. *Arthrosc Sports Med Rehabil*. 2022;4(1):e151-e162. doi:10.1016/j.asmr.2021.09.032

143. Worp H van der, Vrielink JW, Bredeweg SW. Do runners who suffer injuries have higher vertical ground reaction forces than those who remain injury-free? A systematic review and meta-analysis. *Br J Sports Med*. 2016;50(8):450-457. doi:10.1136/bjsports-2015-094924

144. Chen TL, An WW, Chan ZYS, Au IPH, Zhang ZH, Cheung RTH. Immediate effects of modified landing pattern on a probabilistic tibial stress fracture model in runners. *Clin Biomech*. 2016;33:49-54. doi:10.1016/j.clinbiomech.2016.02.013

145. Nordin AD, Dufek JS, Mercer JA. Three-dimensional impact kinetics with foot-strike manipulations during running. *J Sport Health Sci*. 2017;6(4):489-497. doi:10.1016/j.jshs.2015.11.003

146. Baggaley M, Willy RW, Meardon SA. Primary and secondary effects of real-time feedback to reduce vertical loading rate during running. *Scand J Med Sci Sports*. 2017;27(5):501-507. doi:10.1111/sms.12670

147. Whittier T, Willy RW, Sandri Heidner G, et al. The Cognitive Demands of Gait Retraining in Runners: An EEG Study. *J Mot Behav*. 2020;52(3):360-371. doi:10.1080/00222895.2019.1635983

148. Breine B, Malcolm P, Galle S, Fiers P, Frederick EC, De Clercq D. Running speed-induced changes in foot contact pattern influence impact loading rate. *Eur J Sport Sci*. 2019;19(6):774-783. doi:10.1080/17461391.2018.1541256

149. Lower body modeling with Plug-in Gait - Nexus 2.13 Documentation - Vicon Documentation. Lower body modeling with Plug-in Gait. Published 2023. Accessed May 5, 2023. https://docs.vicon.com/display/Nexus213/Lower+body+modeling+with+Plug-in+Gait

150. Hobara H, Sato T, Sakaguchi M, Sato T, Nakazawa K. Step Frequency and Lower Extremity Loading During Running. *Int J Sports Med*. 2012;33(04):310-313. doi:10.1055/s-0031-1291232

151. Brake M te, Stolwijk N, Staal B, Van Hooren B. Using beat frequency in music to adjust running cadence in recreational runners: A randomized multiple baseline design. *Eur J Sport Sci*. 2022;0(0):1-10. doi:10.1080/17461391.2022.2042398

152. Swanson SC, Caldwell GE. An integrated biomechanical analysis of high speed incline and level treadmill running. *Med Sci Sports Exerc*. 2000;32(6):1146-1155. doi:10.1097/00005768-200006000-00018

153. Park SK, Jeon HM, Lam WK, Stefanyshyn D, Ryu J. The effects of downhill slope on kinematics and kinetics of the lower extremity joints during running. *Gait Posture*. 2018;68:181-186. doi:10.1016/j.gaitpost.2018.11.007

154. Nardello F, Venturini N, Skroce K, Tarperi C, Schena F. Kinematic and mechanical changes during a long half-marathon race: males and females at uphill/downhill slopes. *J Sports Med Phys Fitness*. 2021;61(3):350-358. doi:10.23736/S0022-4707.20.11177-0

155. Chan ZYS, Au IPH, Lau FOY, Ching ECK, Zhang JH, Cheung RTH. Does maximalist footwear lower impact loading during level ground and downhill running? *Eur J Sport Sci*. 2018;18(8):1083-1089. doi:10.1080/17461391.2018.1472298

156. Padulo J, Annino G, Smith L, et al. Uphill Running at Iso-Efficiency Speed. *Int J Sports Med*. 2012;33(10):819-823. doi:10.1055/s-0032-1311588

157. Benson LC, Clermont CA, Bošnjak E, Ferber R. The use of wearable devices for walking and running gait analysis outside of the lab: A systematic review. *Gait Posture*. 2018;63:124-138. doi:10.1016/j.gaitpost.2018.04.047

158. Moore IS, Willy RW. Use of Wearables: Tracking and Retraining in Endurance Runners. *Curr Sports Med Rep*. 2019;18(12):437-444. doi:10.1249/JSR.0000000000000667

159. Ahamed NU, Benson LC, Clermont CA, Pohl AJ, Ferber R. New Considerations for Collecting Biomechanical Data Using Wearable Sensors: How Does Inclination Influence the Number of Runs Needed to Determine a Stable Running Gait Pattern? *Sensors*. 2019;19(11):2516. doi:10.3390/s19112516

160. Diedrich FJ, Warren WH. Why change gaits? Dynamics of the walk-run transition. *J Exp Psychol Hum Percept Perform*. 1995;21(1):183-202. doi:10.1037/0096-1523.21.1.183

161. Ahamed NU, Kobsar D, Benson LC, Clermont CA, Osis ST, Ferber R. Subject-specific and group-based running pattern classification using a single wearable sensor. *J Biomech*. 2019;84:227-233. doi:10.1016/j.jbiomech.2019.01.001

162. Schober P, Boer C, Schwarte LA. Correlation Coefficients: Appropriate Use and Interpretation. *Anesthesia & Analgesia*. 2018;126(5):1763. doi:10.1213/ANE.0000000000002864

163. Vermand S, Ferrari FJ, Cherdo F, et al. Running biomechanics alterations during a 40 km mountain race. *J Sports Med Phys Fitness*. 2022;62(10). doi:10.23736/S0022-4707.22.13049-5

164. Wall-Scheffler CM, Chumanov E, Steudel-Numbers K, Heiderscheit B. Electromyography activity across gait and incline: The impact of muscular activity on human morphology. *Am J Phys Anthropol*. 2010;143(4):601-611. doi:10.1002/ajpa.21356

165. Abe D, Fukuoka Y, Muraki S, Yasukouchi A, Sakaguchi Y, Niihata S. Effects of load and gradient on energy cost of running. *J Physiol Anthropol*. 2011;30(4):153-160. doi:10.2114/jpa2.30.153

166. van der Worp H, Vrielink JW, Bredeweg SW. Do runners who suffer injuries have higher vertical ground reaction forces than those who remain injury-free? A systematic review and meta-analysis. *Br J Sports Med*. 2016;50(8):450-457. doi:10.1136/bjsports-2015-094924

167. Simon SR. Quantification of human motion: gait analysis—benefits and limitations to its application to clinical problems. *J Biomech*. 2004;37(12):1869-1880. doi:10.1016/j.jbiomech.2004.02.047

168. Giandolini M, Gimenez P, Temesi J, et al. Effect of the Fatigue Induced by a 110-km Ultramarathon on Tibial Impact Acceleration and Lower Leg Kinematics. *PLoS ONE*. 2016;11(3):e0151687. doi:10.1371/journal.pone.0151687

169. Brayne L, Barnes A, Heller B, Wheat J. Using a wireless consumer accelerometer to measure tibial acceleration during running: agreement with a skin-mounted sensor. *Sports Eng*. 2018;21(4):487-491. doi:10.1007/s12283-018-0271-4

170. Winter DA. *Biomechanics and Motor Control of Human Movement*. 4th ed. Wiley; 2009.

171. Burke A, Dillon S, O'Connor S, Whyte EF, Gore S, Moran KA. Relative and absolute reliability of shank and sacral running impact accelerations over a short- and long-term time frame. *Sports Biomech*. 2022;0(0):1-16. doi:10.1080/14763141.2022.2086169

172. Churchill SM, Salo AIT, Trewartha G. The effect of the bend on technique and performance during maximal effort sprinting. *Sports Biomech*. 2015;14(1):106-121. doi:10.1080/14763141.2015.1024717

173. Sun X, Lam WK, Zhang X, Wang J, Fu W. Systematic Review of the Role of Footwear Constructions in Running Biomechanics: Implications for Running-Related Injury and Performance. *J Sports Sci Med*. 2020;19(1):20-37.

174. McLaughlin P. Testing agreement between a new method and the gold standard-how do we test? *J Biomech*. 2013;46(16):2757-2760. doi:10.1016/j.jbiomech.2013.08.015

175. Bland JM, Altman DG. Measuring agreement in method comparison studies. *Stat Methods Med Res*. 1999;8(2):135-160. doi:10.1177/096228029900800204

176. Tenforde AS, Borgstrom HE, Outerleys J, Davis IS. Is Cadence Related to Leg Length and Load Rate? *J Orthop Sports Phys Ther*. 2019;49(4):280-283. doi:10.2519/jospt.2019.8420

177. Lucas-Cuevas AG, Quesada JIP, Giménez JV, Aparicio I, Jimenez-Perez I, Pérez-Soriano P. Initiating running barefoot: Effects on muscle activation and impact accelerations in habitually rearfoot shod runners. *Eur J Sport Sci*. 2016;16(8):1145-1152. doi:10.1080/17461391.2016.1197317

178. Lucas-Cuevas AG, Encarnación-Martínez A, Camacho-García A, Llana-Belloch S, Pérez-Soriano P. The location of the tibial accelerometer does influence impact acceleration parameters during running. *Journal of Sports Sciences*. 2017;35(17):1734-1738. doi:10.1080/02640414.2016.1235792

179. Giavarina D. Understanding Bland Altman analysis. *Biochem Med*. 2015;25(2):141-151. doi:10.11613/BM.2015.015

180. Kumar D, McDermott K, Feng H, et al. Effects of Form-Focused Training on Running Biomechanics: A Pilot Randomized Trial in Untrained Individuals. *PM&R*. 2015;7(8):814-822. doi:10.1016/j.pmrj.2015.01.010

181. Esculier JF, Bouyer LJ, Dubois B, et al. Is combining gait retraining or an exercise programme with education better than education alone in treating runners with patellofemoral pain?A randomised clinical trial. *Br J Sports Med*. 2018;52(10):659-666. doi:10.1136/bjsports-2016-096988

182. Boston Marathon Course Information | Boston Athletic Association. Accessed May 23, 2023. https://www.baa.org/races/boston-marathon/enter/course-information

183. Gilgen-Ammann R, Schweizer T, Wyss T. Accuracy of Distance Recordings in Eight Positioning-Enabled Sport Watches: Instrument Validation Study. *JMIR mHealth and uHealth*. 2020;8(6):e17118. doi:10.2196/17118

184. Johansson RE, Adolph ST, Swart J, Lambert MI. Accuracy of GPS sport watches in measuring distance in an ultramarathon running race. *Int J Sports Sci Coach*. 2020;15(2):212-219. doi:10.1177/1747954119899880

185. Benson LC, Clermont CA, Ferber R. New Considerations for Collecting Biomechanical Data Using Wearable Sensors: The Effect of Different Running Environments. *Front Bioeng Biotechnol*. 2020;8. doi:10.3389/fbioe.2020.00086

186. Benson LC, Ahamed NU, Kobsar D, Ferber R. New considerations for collecting biomechanical data using wearable sensors: Number of level runs to define a stable running pattern with a single IMU. *J Biomech*. 2019;85:187-192. doi:10.1016/j.jbiomech.2019.01.004

187. Dunn MD, Claxton DB, Fletcher G, Wheat JS, Binney DM. Effects of running retraining on biomechanical factors associated with lower limb injury. *Hum Mov Sci*. 2018;58:21-31. doi:10.1016/j.humov.2018.01.001

188. Aranki D, Peh GX, Kurillo G, Bajcsy R. The Feasibility and Usability of RunningCoach: A Remote Coaching System for Long-Distance Runners. *Sensors*. 2018;18(1):175. doi:10.3390/s18010175

189. Goss DL, Watson DJ, Miller EM, Weart AN, Szymanek EB, Freisinger GM. Wearable Technology May Assist in Retraining Foot Strike Patterns in Previously Injured Military Service Members: A Prospective Case Series. *Front Sports Act Living*. 2021;3. doi:10.3389/fspor.2021.630937

190. Locke EA. Motivation through conscious goal setting. *Applied and Preventive Psychology*. 1996;5(2):117-124. doi:10.1016/S0962-1849(96)80005-9

191. NURVV Run Smart Insoles. NURVV Run Smart Insoles. Accessed August 9, 2023. https://www.nurvv.com/en-gb/products/nurvv-run-insoles-trackers/

192. Giraldo-Pedroza A, Lee WCC, Lam WK, Coman R, Alici G. Effects of Wearable Devices with Biofeedback on Biomechanical Performance of Running—A Systematic Review. *Sensors*. 2020;20(22):6637. doi:10.3390/s20226637

193. Noehren B, Scholz J, Davis I. The effect of real-time gait retraining on hip kinematics, pain and function in subjects with patellofemoral pain syndrome. *Br J Sports Med*. 2011;45(9):691-696. doi:10.1136/bjsm.2009.069112

194. Creaby MW, Franettovich Smith MM. Retraining running gait to reduce tibial loads with clinician or accelerometry guided feedback. *J Sci Med Sport*. 2016;19(4):288-292. doi:10.1016/j.jsams.2015.05.003

195. Lorenzoni V, Van den Berghe P, Maes PJ, De Bie T, De Clercq D, Leman M. Design and validation of an auditory biofeedback system for modification of running parameters. *J Multimodal User Interfaces*. 2019;13(3):167-180. doi:10.1007/s12193-018-0283-1

196. Walker BN, Nees MA. Theory of Sonification. In: Hermann T, Hunt A, Neuhoff JG, eds. *The Sonification Handbook*. Logos Verlag; 2011.

197. Sigrist R, Rauter G, Riener R, Wolf P. Augmented visual, auditory, haptic, and multimodal feedback in motor learning: a review. *Psychon Bull Rev*. 2013;20(1):21-53. doi:10.3758/s13423-012-0333-8

198. Ahamed NU, Kobsar D, Benson L, et al. Using wearable sensors to classify subject-specific running biomechanical gait patterns based on changes in environmental weather conditions. *PLoS One*. 2018;13(9):e0203839. doi:10.1371/journal.pone.0203839

199. Dmitry K. Knee Point. Accessed May 17, 2023. https://www.mathworks.com/matlabcentral/fileexchange/35094-knee-point

200. On The Go Map - 5.02 km route. Accessed May 27, 2023. http://onthegomap.com/s/vi7grjk3

APPENDIX I

Publications arising from the thesis

Articles published in peer-reviewed journals:

Chan, P.P.K.; Chan, Z.Y.S.; Au, I.P.H.; Lam, B.M.F.; Lam, W.K.; Cheung, R.T.H. Biomechanical Effects Following Footstrike Pattern Modification Using Wearable Sensors. *J Sci Med Sport* **2021**, *24*, 30–35, doi:10.1016/j.jsams.2020.05.019

Chan, Z.Y.S.; Angel, C.; Thomson, D.; Ferber, R.; Tsang, S.M.H.; Cheung, R.T.H. Evaluation of a Restoration Algorithm Applied to Clipped Tibial Acceleration Signals. *Sensors* **2023**, *23*, 4609, doi:10.3390/s23104609

APPENDIX II

Other publications during PhD candidature

Articles published in peer-reviewed journals:

An, W.W.; Ting, K.-H.; Au, I.P.H.; Zhang, J.H.; Chan, Z.Y.S.; Davis, I.S.; So, W.K.Y.; Chan, R.H.M.; Cheung, R.T.H. Neurophysiological Correlates of Gait Retraining With Real-Time Visual and Auditory Feedback. *IEEE Trans Neural Syst Rehabil Eng* **2019**, *27*, 1341–1349, doi:10.1109/TNSRE.2019.2914187.

Chan, Z.Y.S.; MacPhail, A.J.C.; Au, I.P.H.; Zhang, J.H.; Lam, B.M.F.; Ferber, R.; Cheung, R.T.H. Walking with Head-Mounted Virtual and Augmented Reality Devices: Effects on Position Control and Gait Biomechanics. *PLoS One* **2019**, *14*, e0225972, doi:10.1371/journal.pone.0225972.

Chan, Z.Y.S.; Peeters, R.; Cheing, G.; Ferber, R.; Cheung, R.T.H. Evaluation of COVID-19 Restrictions on Distance Runners' Training Habits Using Wearable Trackers. *Front Sports Act Living* **2022**, *3*, 812214, doi:10.3389/fspor.2021.812214.

Cheung, R.T.H.; Ho, K.K.W.; Au, I.P.H.; An, W.W.; Zhang, J.H.W.; Chan, Z.Y.S.; Deluzio, K.; Rainbow, M.J. Immediate and Short-Term Effects of Gait Retraining on the Knee Joint Moments and Symptoms in Patients with Early Tibiofemoral Joint Osteoarthritis: A Randomized Controlled Trial. *Osteoarthritis Cartilage* **2018**, *26*, 1479–1486, doi:10.1016/j.joca.2018.07.011.

Cheung, R.T.H.; Zhang, J.H.; Chan, Z.Y.S.; An, W.W.; Au, I.P.H.; MacPhail, A.; Davis, I.S. Shoe-mounted Accelerometers Should Be Used with Caution in Gait Retraining. *Scand J Med Sci Sports* **2019**, *29*, 835–842, doi:10.1111/sms.13396.

Cheung, V.C.K.; Cheung, B.M.F.; Zhang, J.H.; Chan, Z.Y.S.; Ha, S.C.W.; Chen, C.-Y.; Cheung, R.T.H. Plasticity of Muscle Synergies through Fractionation and Merging during

Development and Training of Human Runners. *Nat Commun* **2020**, *11*, 4356, doi:10.1038/s41467-020-18210-4.

Ching, E.; An, W.; Au, I.; Zhang, J.; Chan, Z.; Shum, G.; Cheung, R. Impact Loading During Distracted Running Before and After Auditory Gait Retraining. *Int J Sports Med* **2018**, *39*, 1075–1080, doi:10.1055/a-0667-9875.

Law, M.H.C.; Choi, E.M.F.; Law, S.H.Y.; Chan, S.S.C.; Wong, S.M.S.; Ching, E.C.K.; Chan, Z.Y.S.; Zhang, J.H.; Lam, G.W.K.; Lau, F.O.Y.; Cheung R.T.H. Effects of Footwear Midsole Thickness on Running Biomechanics. *J Sports Sci* **2018**, 1–7, doi:10.1080/02640414.2018.1538066.

Mo, S.; Chan, Z.Y.S.; Lai, K.K.Y.; Chan, P.P.-K.; Wei, R.X.-Y.; Yung, P.S.-H.; Shum, G.; Cheung, R.T.-H. Effect of Minimalist and Maximalist Shoes on Impact Loading and Footstrike Pattern in Habitual Rearfoot Strike Trail Runners: An in-Field Study. *Eur J Sport Sci* **2021**, *21*, 183–191, doi:10.1080/17461391.2020.1738559.

Mo, S.; Lam, W.-K.; Ching, E.C.K.; Chan, Z.Y.S.; Zhang, J.H.; Cheung, R.T.H. Effects of Heel-Toe Drop on Running Biomechanics and Perceived Comfort of Rearfoot Strikers in Standard Cushioned Running Shoes. *Footwear Sci* **2020**, *12*, 91–99, doi:10.1080/19424280.2020.1734868.

Mo, S.; Lau, F.O.Y.; Lok, A.K.Y.; Chan, Z.Y.S.; Zhang, J.H.; Shum, G.; Cheung, R.T.H. Bilateral Asymmetry of Running Gait in Competitive, Recreational and Novice Runners at Different Speeds. *Hum Mov Sci* **2020**, *71*, 102600, doi:10.1016/j.humov.2020.102600.

Mo, S.; Leung, S.H.S.; Chan, Z.Y.S.; Sze, L.K.Y.; Mok, K.-M.; Yung, P.S.H.; Ferber, R.; Cheung, R.T.H. The Biomechanical Difference between Running with Traditional and 3D Printed Orthoses. *J Sports Sci* **2019**, *37*, 2191–2197, doi:10.1080/02640414.2019.1626069.

Mok, K.-M.; Ha, S.C.W.; Chan, Z.Y.S.; Yung, P.S.H.; Fong, D.T.P. An Inverted Ankle Joint Orientation at Foot Strike Could Incite Ankle Inversion Sprain: Comparison between Injury and Non-Injured Cutting Motions of a Tennis Player. *Foot* **2021**, *48*, 101853, doi:10.1016/j.foot.2021.101853.

Wang, C.; Chan, P.P.K.; Lam, B.M.F.; Wang, S.; Zhang, J.H.; Chan, Z.Y.S.; Chan, R.H.M.; Ho, K.K.W.; Cheung, R.T.H. Real-Time Estimation of Knee Adduction Moment for Gait Retraining in Patients With Knee Osteoarthritis. *IEEE Trans. Neural Syst Rehabil Eng.* **2020**, *28*, 888–894, doi:10.1109/TNSRE.2020.2978537.

Wang, S.; Chan, P.P.K.; Lam, B.M.F.; Chan, Z.Y.S.; Zhang, J.H.W.; Wang, C.; Lam, W.K.; Ho, K.K.W.; Chan, R.H.M.; Cheung, R.T.H. Sensor-Based Gait Retraining Lowers Knee Adduction Moment and Improves Symptoms in Patients with Knee Osteoarthritis: A Randomized Controlled Trial. *Sensors* **2021**, *21*, 5596, doi:10.3390/s21165596.

Wei, R.X.; Au, I.P.H.; Lau, F.O.Y.; Zhang, J.H.; Chan, Z.Y.S.; MacPhail, A.J.C.; Mangubat, A.L.; Pun, G.; Cheung, R.T.H. Running Biomechanics before and after Pose® Method Gait Retraining in Distance Runners. *Sports Biomech* **2021**, *20*, 958–973, doi:10.1080/14763141.2019.1624812.

Wei, R.X.Y.; Chan, Z.Y.S.; Zhang, J.H.W.; Shum, G.L.; Chen, C.-Y.; Cheung, R.T.H. Difference in the Running Biomechanics between Preschoolers and Adults. *Braz J Phys Ther* **2021**, *25*, 162–167, doi:10.1016/j.bjpt.2020.05.003.

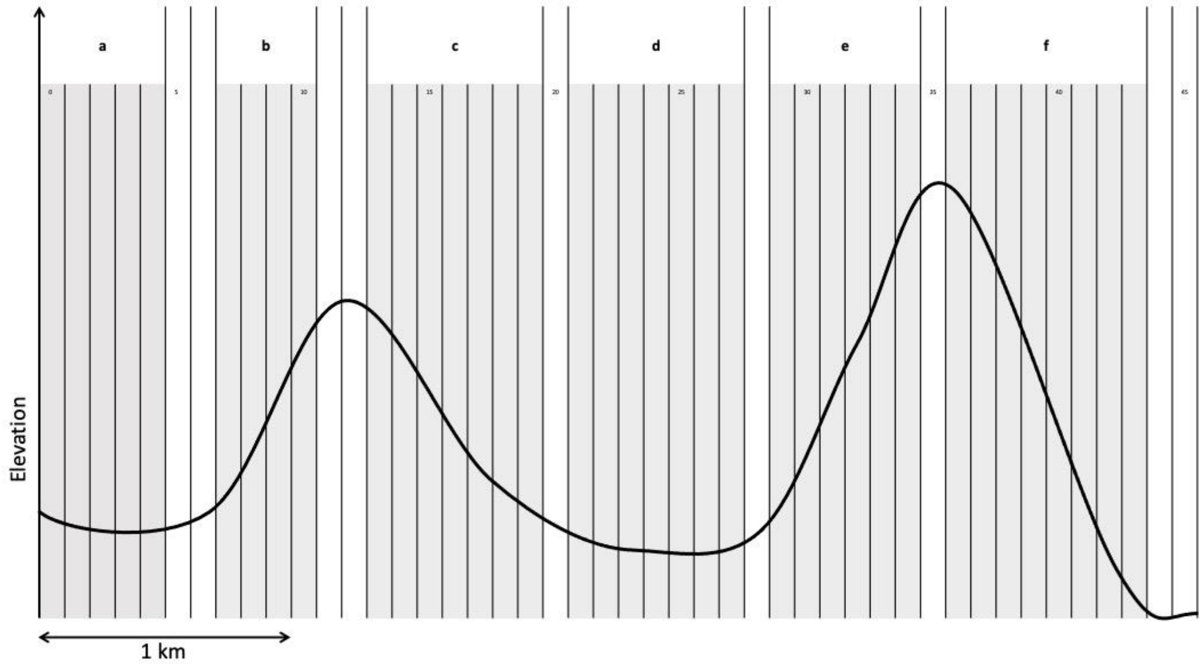
Zhang, J.H.; Chan, Z.Y.S.; Au, I.P.H.; An, W.W.; Cheung, R.T.H. Can Runners Maintain a Newly Learned Gait Pattern Outside a Laboratory Environment Following Gait Retraining? *Gait Posture* **2019**, *69*, 8–12, doi:10.1016/j.gaitpost.2019.01.014.

Zhang, J.H.; Chan, Z.Y.S.; Lau, F.O.Y.; Huang, M.; Wang, A.C.; Wang, S.; Au, I.P.H.; Wang, S.; Lam, B.M.F.; An, W.W.; Cheung R.T.H. How Do Training Experience and Geographical Origin of a Runner Affect Running Biomechanics? *Gait Posture* **2021**, *84*, 209–214, doi:10.1016/j.gaitpost.2020.12.003.

Zhang, J.H.; Chan, Z.Y.-S.; Au, I.P.-H.; An, W.W.; Shull, P.B.; Cheung, R.T.-H. Transfer Learning Effects of Biofeedback Running Retraining in Untrained Conditions. *Med Sci Sports Exerc* **2019**, doi:10.1249/MSS.0000000000002007.

APPENDIX III

An example of running record segmentation



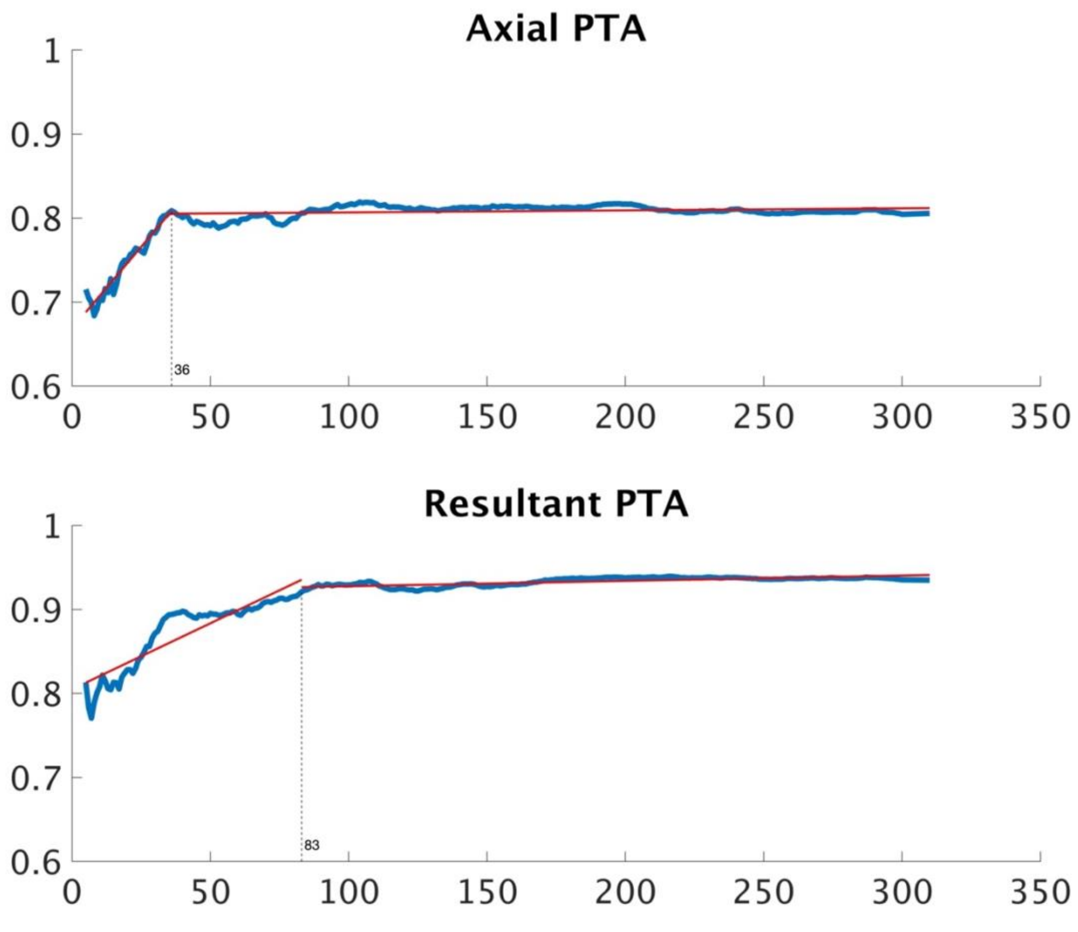
The figure above presents an example of the elevation profile across a recorded run. The record has been segmented into 100 m sections as indicated by the vertical lines. The table below summarized the segments identified for analysis.

Segment	Description	Condition
a	First 500 m of the run: warm-up	Excluded
b	4 consecutive 100-m sections between 3 and 15% This segment is less than 600 m of the same conditions	Excluded
c	7 consecutive 100 m sections between -15 and -3% Sections 2 to 6 (total 500 m) was used for analysis	Downhill
d	7 consecutive 100 m sections between -2 and 2% Sections 2 to 6 (total 500 m) was used for analysis	Level
e	6 consecutive 100 m sections between 3 and 15% Sections 2 to 6 (total 500 m) was used for analysis	Uphill
f	8 consecutive 100 m sections between -15 and -3% This is the second segment of the condition “downhill”	Excluded

APPENDIX IV

Knee-point detection algorithm to detect plateau of ICC

The knee-point detection algorithm was adopted from the *knee_pt()* function developed by Dmitry Kaplan.[199] In short, the bisection point which minimized the sum of errors between two best fit lines (fitting all data points on the left and right of the bisection point) is considered the 'knee point'.



Bisection point: 36 for axial PTA and 83 for resultant PTA.

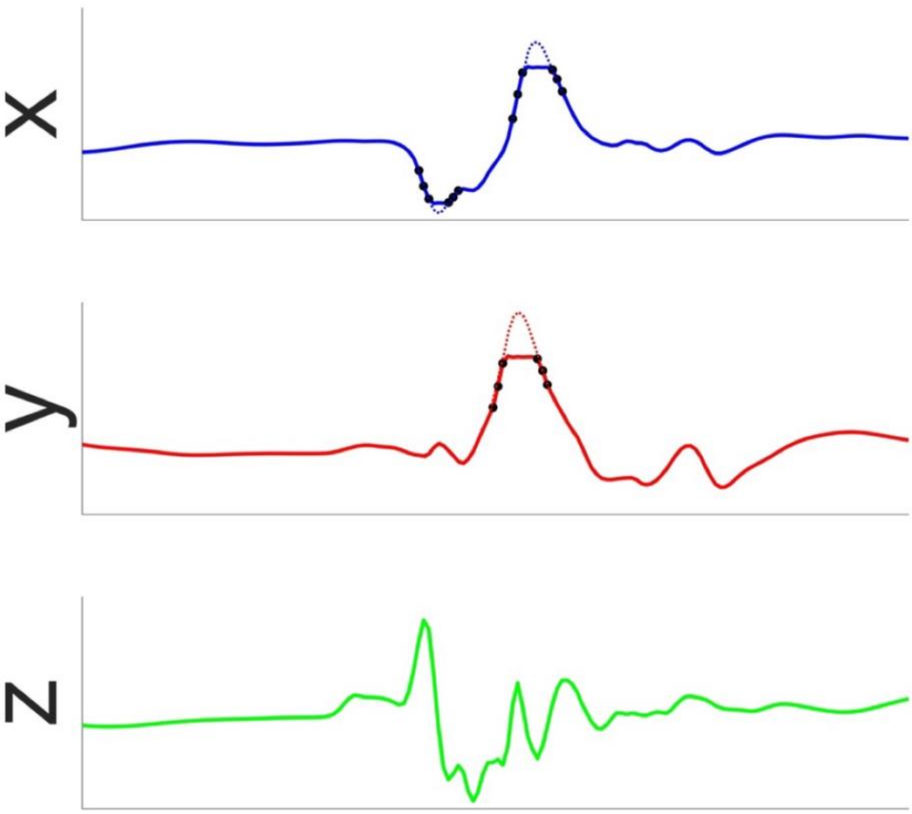
APPENDIX V

Correction algorithm for restoring clipped signals

The correction algorithm was adopted from Ruder *et al*.'s study.[74]

The algorithm is described in steps with reference to the figure below, as follows:

1. The correction algorithm is applied to each axis (x-, y- and z-axis) separately.
2. The clipped portion of the raw signal (solid lines) is identified by consecutive data points (minimum: 2) with the value ±15.985 g (flat portions of the solid lines).
3. Three data points before and after the clipping (solid black dots) are used for reconstruction of the signal by 5th order spline interpolation with MATLAB functions: *spapi()* and *fnval()*.
4. The magnitude of the peak/trough within the reconstructed portion is assessed. The reconstructed signal is rejected if the magnitude is within a pre-set range (i.e., ±16 *g*).
5. If the magnitude of the peak/trough is outside of the range, reconstruction is considered successful. The clipped portion is replaced with the reconstructed signal (dotted line).

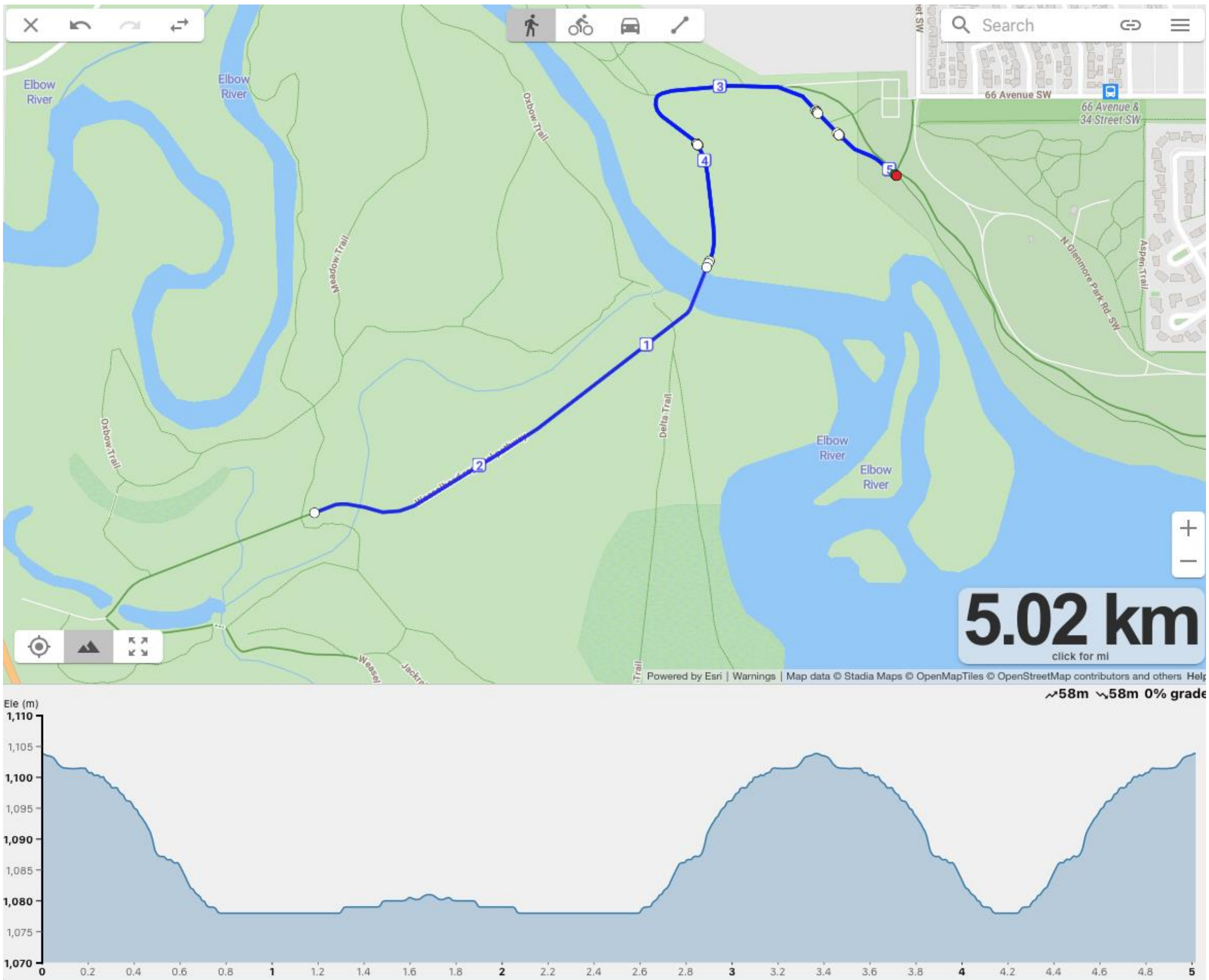


APPENDIX VI

Running route used for outdoor field-based assessment

The map and elevation profile of the 5-km route can be accessed online:[200]

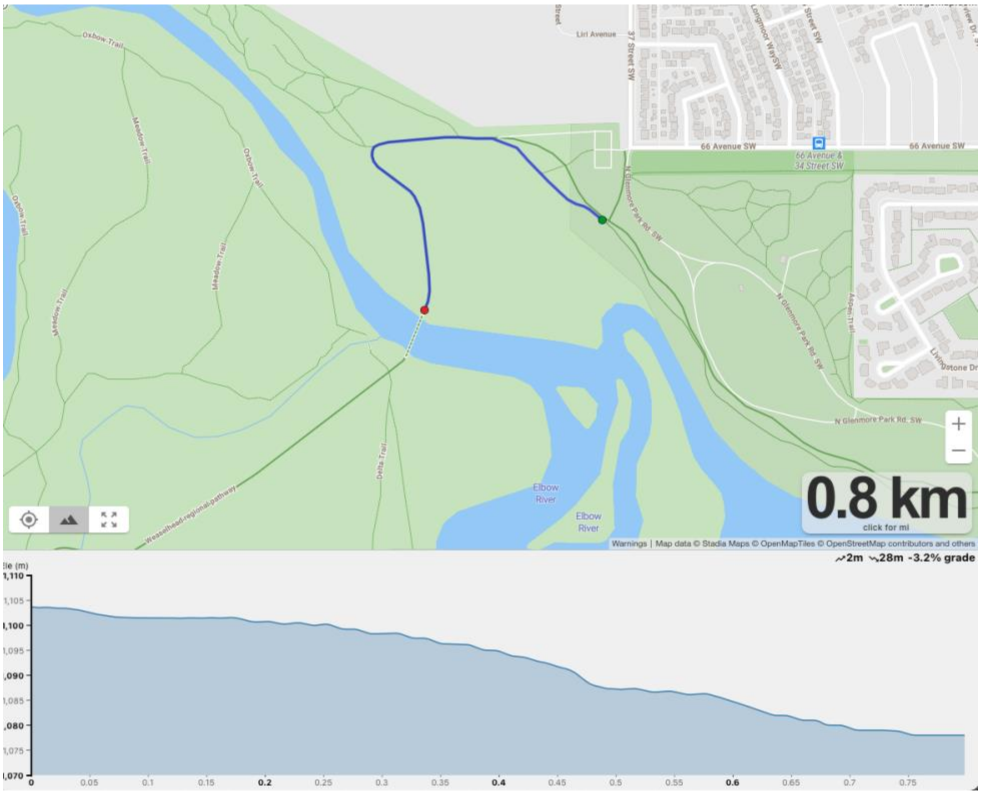
https://onthegomap.com/s/vi7grjk3



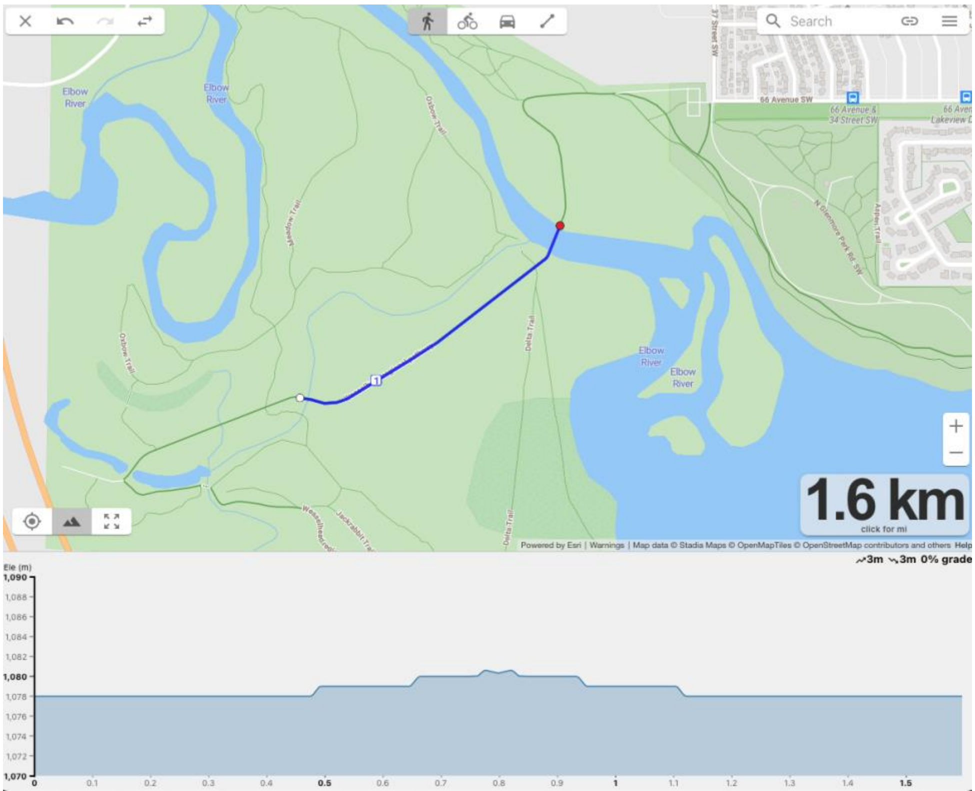
The running course used for the outdoor field-based assessments is made up of 3 sections and the runner completed the sections in the order: A > B > C > A > C.

Section	Gradient (%), min	Gradient (%), max	Length (m)
A	-7.32	-0.64	800
B	-0.87	0.75	1600
C	0.64	7.32	800

Section A: Downhill; length: 0.8 km; average grade -3.2%



Section B: Level; length: 1.6 km; average grade 0%



Section C: Uphill; length: 0.8 km; average grade +3.2%