Gait & Posture 113 (2024) 519–527

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Load increases IMU signal attenuation per step but reduces IMU signal attenuation per kilometre

AuraLea Fain [a,b], Bradley C. Nindl [c,d], Ayden McCarthy [a,b], Joel T. Fuller [a,b], Jodie A. Wills [a,b], Tim L.A. Doyle [a,b,*]

[a] Biomechanics, Physical Performance, and Exercise (BioPPEx) Research Group, Macquarie University, Sydney, Australia
[b] Faculty of Medicine, Health, and Human Sciences, Macquarie University, Sydney, Australia
[c] Neuromuscular Research Lab/Warrior Performance Center, University of Pittsburgh, Pittsburgh, PA, USA
[d] Department of Sports Medicine, University of Pittsburgh, Pittsburgh, PA, USA

ARTICLE INFO

Keywords:
Wearable sensors
Accelerations
Grade
Load carriage

ABSTRACT

Background: Despite deleterious biomechanics associated with injury, particularly as it pertains to load carriage, there is limited research on the association between physical demands and variables captured with wearable sensors. While inertial measurement units (IMUs) can be used as surrogate measures of ground reaction force (GRF) variables, it is unclear if these data are sensitive to military-specific task demands.
Research question: Can wearable sensors characterise physical load and demands placed on individuals in different load, speed and grade conditions?
Methods: Data were collected on 20 individuals who were self-reportedly free from current injury, recreationally active, and capable of donning 23 kg in the form of a weighted vest. Each participant walked and ran on flat, uphill (+6 %) and downhill (-6 %) without and with load (23 kg). Data were collected synchronously from optical motion capture (OMC) and IMUs placed on the distal limb and the pelvis. Data from an 8-second window was used to generate a participant-based mean of OMC and IMU variables of interest. Repeated Measures ANOVA was used to measure main and interaction effects of load, speed, and grade. Simple linear regression was used to elucidate a relationship between OMC measures and estimated metabolic cost (EMC) to IMU measures.
Results: Load reduces foot and pelvic accelerations ($p<0.001$) but elevate signal attenuation per step ($p=0.044$). Conversely, attenuation per kilometre is lowered with the addition of load ($p=0.017$). Uphill had the lowest attenuation per step ($p=0.003$) and kilometre ($p\leq0.033$) in walking, while downhill had the greatest attenuation per step ($p\leq0.002$) and per kilometre ($p\leq0.004$). Attenuation measures are inconsistently moderately related to limb negative work ($R\leq0.57$). EMC is moderately positively related to unloaded running ($R\geq0.39$), and moderately negatively related to walking with and without load ($R\leq-0.52$).
Significance: While load reduces peak accelerations at both the pelvis and foot. However, it may increase demand on the lower extremity to attenuate the signal between the two sensors with each step, while attenuation over time reduces with load.

1. Introduction

Lower extremity musculoskeletal injuries pose a significant threat to health and readiness in military populations globally [1]. These injuries are cited as the primary reason for attrition, resulting in up to a quarter and a third of individuals dropping out of training and combat scenarios, respectively [2–4]. The most common aetiology cited for such injuries is outdoor marching, which often necessitates the addition of body borne load while navigating over variable terrain and grade [1]. In training, loaded ruck marches are often long in duration, wherein the average load carried is around 23 kg [5].

Laboratory-based biomechanics studies have endeavoured to elucidate three-dimensional (3D) kinematic and kinetic variables that may be deleterious in these scenarios. Studies report heightened ground reaction force (GRF) variables, altered spatiotemporal parameters, and lower extremity motions and loads that may increase risk of injury

* Correspondence to: Department of Health Sciences, Macquarie University, 75 Talavera Road, Macquarie Park 2113, Australia.
E-mail address: tim.doyle@mq.edu.au (T.L.A. Doyle).

https://doi.org/10.1016/j.gaitpost.2024.08.003
Received 8 February 2024; Received in revised form 27 June 2024; Accepted 6 August 2024
Available online 21 August 2024

compared to unloaded conditions [6–8]. Altered biomechanics that occur during load carriage are also paired with an increase in metabolic cost as quantified by oxygen consumption [9]. Specifically, the already high metabolic cost of load carriage may be increased further in uphill and/or increased speed conditions, while downhill conditions may reduce metabolic cost [9,10]. Though biomechanical and physiological studies have supplied invaluable information on maladaptive movement patterns and altered metabolic demands, these metrics necessitate a laboratory environment outfitted with extensive 3D optical motion capture (OMC) and physiological measurement systems. Further, these environments are free from physical and cognitive perturbations experienced in training and operational scenarios. In an effort to bridge the gap between laboratory and field-based metrics, wearable sensors have been developed to quantify metrics that may be indicative of physical demands in organic environments for both athletic and tactical populations [11–13]. However, it is unclear how these metrics are influenced by military-specific demands, such as the addition of body-borne load and grade, and if they are associated with estimated physiological demands.

Evidence suggests there is an incremental increase in GRF with increasing load carried, which elevates the demand on the lower extremity to dissipate elevated forces compared to unloaded conditions [6,8,14]. Increased GRF in a loaded condition compared to an unloaded condition requires the lower extremity to attenuate these potentially deleterious forces up the limb; however, load reportedly results in a stiff limb [8]. A stiff limb reduces the amount of energy actively dissipated by the musculature of the lower extremity and shifts the attenuation demands to other bodily structures (e.g., bone). Another metric of the lower extremity actively attenuating energy up the kinetic chain is total limb negative work, which may be a surrogate measure of the demand placed on the soft tissues of the lower extremity to dissipate energy opposed to the bone. Both increased stiffness and reduced negative work increase the transfer of forces to bony structures of the lower extremity compared to a flexed limb and consequently may increase risk of overuse bone and joint injuries. Muscular demands change in response to the addition of body borne load and/or variable terrain; these task demands increase muscular activity to both generate and dissipate energy, consequently resulting in the observed altered metabolic costs of load carriage [15,16]. Researchers have developed and validated linear models of estimated metabolic cost (EMC) to estimate physiological demands placed on individuals participating in field-based load carriage tasks (between 10 and 40 kg) and various speeds [9,10,17]. Being able to transfer knowledge on the consequences of load carriage on biomechanical and physiological cost to an organic environment may help practitioners more readily quantify training load and risk of injury.

Capturing laboratory-based, valid biomechanical data have often eluded researchers who aim to implement wearable sensors to measure performance and indicators of musculoskeletal injury risk in real-world scenarios. Inertial measurement units (IMUs) are small, wearable sensors that capture data related to 3D accelerations. The most common application of IMUs is quantification of tibial accelerations, which reportedly have an extremely close relationship with vertical GRF and loading rate both with and without the addition of load [12,18]. Beyond this singular IMU approach to quantifying mechanical load, adding an additional IMU proximally up the limb (pelvis, cervical spine, or head) can serve to quantify the amount of energy being dissipated between the two sensor locations [19–21]. This measure may supply meaningful information that may be related to injuries that are commonplace in populations that sustain lower extremity overuse injuries that occur as a consequence of repetitive loading, including military personnel [1]. Despite the fact that signal attenuation can supply valuable information regarding the ability of the lower extremity to effectively dissipate energy, only 7 % of studies utilizing IMUs examined this variable of shock attenuation [20]. Not only is there a dearth of this metric, but it is also unclear how shock attenuation may be influenced by speed, load, and grade and if altered metabolic cost profiles change due to these military-specific demands. Additionally, it is important to understand how these factors of speed, load and terrain may impact the total demands placed on the lower extremity over a set distance march, rather than a single step. Given that these task factors can influence both spatiotemporal (e.g., step length and frequency) and accelerations associated with each step, it is possible that the total effect can be magnified or reduced based on changes to the number of steps required to cover a set distance over varied terrain. For example, individuals may reduce attenuation metrics per step, but a simultaneous increase in stride frequency may result in greater attenuation over the course of a kilometre or more. Knowledge of alterations in IMU acceleration metrics per step and per kilometre may enable practitioners to monitor and prescribe marching workload so positive adaptation is facilitated and deleterious strain on the musculoskeletal system is avoided.

Despite the close relationship reported between IMU and vertical GRF measures, and the ability of IMUs to estimate mechanical loading through signal attenuation, it is not clear how these IMU measures are influenced by military-specific demands. Therefore, the purpose of this study was to elucidate if: 1) IMU acceleration variables were sensitive to changes in speed, load, and grade often experienced in tactical scenarios and 2) IMU acceleration variables are related to lower extremity biomechanics. Supplemental post-hoc analysis exploring the estimated physiological cost was included in an effort to begin exploration into the relationship between IMU variables and physiological demands.

2. Methods

2.1. Participants

This study recruited 20 participants (10 M, 10 F; 26.5 ± 7.1 years, 1.7 ± 8.7 m, 70.6 ± 14.8 kg) based off a power analysis (80 % statistical power with an α-level of 0.05). All participants were above the age of 18, and self-reportedly active, free from current injury, and capable of carrying 23 kg in the form of a weighted vest. Each participant signed a written informed consent prior to commencement of data collection. This study received institutional ethics approval (HREC Project Number: 522022787737978). One participant dropped out due to injury unrelated to the study, so the data presented includes 19 individuals.

2.2. Procedures

A single data collection included walking and running (*1.25* $m{\cdot}s^{-1}$ and *2.25* $m{\cdot}s^{-1}$, respectively) on flat (1 %), uphill (+6 %) and downhill (-6 %) grades both without and with load on an instrumented treadmill with two 6-axis force platforms (AMTI, Watertown, MA). It is reported that a treadmill grade of 1 % most accurately represents outdoor locomotion [22]. As such, this grade was used for our "flat" condition. The order of speed and grade was standardized for each participant, while the load condition was randomized. If a participant was completing the loaded condition, a 23 kg weighted vest with evenly distributed weights was placed on their torso. The grades were completed in the following order: flat, uphill, and downhill. Walking was completed first for every grade, followed by running before moving on to the next grade (e.g., flat walking, flat running; uphill walking, uphill running; downhill walking, downhill running).

2.3. IMU data and variables

Tri-axial IMU acceleration data (Low-g: ± 16 g, 1125 Hz; High-g: ± 200 g, 1600 Hz; BlueTrident, iMeasureU, Vicon Ltd., London, UK) were collected simultaneously with optical and Analog motion capture signals via Bluetooth connection and sensor integration in Vicon Nexus (v.2.12.1 Vicon Ltd., Oxford, UK). IMUs were placed on the lateral aspect of the right distal foot, and the midpoint between the posterior superior iliac spines of the pelvis. Distal limb IMUs are often placed on the tibia, however military personnel are required to wear footwear that prevents this placement. As such, the distal limb IMUs were placed on footwear, just distal to lateral malleoli. Raw IMU signals from the sensors were automatically up-sampled to 1250 Hz with a simple linear interpolation within the Vicon Nexus software to maintain time-synchronization between the optical motion capture and force platform data collection system when exported to text files. IMU data were analysed using a custom MatLab script (v.R2021b, The Mathworks, Inc., Natick, MA) that filtered the data (4th Order Butterworth, 12 Hz), and identified peak vertical accelerations of each stride during a 10,000 frame (8 second) window. For the purposes of this study, vertical accelerations were in alignment with the direction of peak bone loading during locomotion. As such, the vertical axis was defined according to the local coordinate system of each segmental IMU. For both the pelvis and the foot (placed orthogonally to the pelvis IMU), it is defined as the axis in alignment with gravity when in anatomical neutral position (Fig. 1). Data were collected for approximately 30 seconds, this region of interest was chosen as this gave ample time to ensure participants were moving at, and comfortable with, the required speed for the given condition. Foot and pelvis peak vertical accelerations were identified, normalized to system weight (*unloaded:* participant weight, *loaded:* participant weight + 23 kg), and averaged across the 8-second window. Peak vertical signal attenuation ($vAtten_{sig}$) was defined as the difference between peak foot vertical acceleration ($vAcc_{Foot}$) and pelvis vertical acceleration ($vAcc_{Pelvis}$) (Eq. 1). Additionally, the relative peak vertical attenuation ($vAtten_{rel}$) at the pelvis was defined as the result of dividing the peak vertical acceleration at the pelvis by that of the foot (Eq. 2).

$$vAtten_{sig} = vAcc_{Foot} - vAcc_{Pelvis} \tag{1}$$

$$vAtten_{rel} = \frac{vAcc_{Pelvis}}{vAcc_{Foot}} \tag{2}$$

To estimate attenuation normalized to system weight per kilometre ($vAtten_{km}$), the number of steps taken for the data analysis window was calculated for each condition for all participants. Condition-specific step rates and the standardized speed (S; $km{\cdot}hr^{-1}$) for walking and running were used to estimate steps per kilometre (Eq. 3).

$$Atten_{km} = \left[\left(\frac{steps}{min} \times 60\right) \div S\right] \times Atten_{sig} \tag{3}$$

2.4. Motion capture and GRF variables

Ten high-speed optical motion capture cameras (MX-T40 S, 250 Hz, Vicon Motion Systems Ltd., Oxford, UK) tracked trajectories of 40 retroreflective markers and clusters placed on participants anatomical landmarks. A participant-based kinematic model was generated with a trunk, bilateral feet, shank, thigh, and CODA pelvis (Visual3D, C-Motion v.2021.11.3). Each joint (ankle, knee and hip) had 6 degrees of freedom, and ankle joint centre was defined according to ISB recommendations [23], while the knee and hip were defined according to Schwartz and Rozumalski [24]. Each participant's centre of mass (COM) was automatically calculated within the Visual3D software. The weight is equally distributed circumventing the torso, negligibly altering the subject COM. As such, it was not added to the Visual3D model. Analog signals from the instrumented treadmill were collected synchronously with optical motion capture measurements (1250 Hz). Force platform signals were filtered using a 4th order Butterworth Filter and a cut-off frequency of 20 Hz. To estimate vertical stiffness, the peak vertical ground reaction force ($vGRF_{pk}$, normalized to system mass in kg), was divided by the change in centre of mass (ΔCOM, metres) for each stride [25]. Vertical stiffness was calculated for the same strides used for IMU data using Visual3D software defined gait events and averaged across strides to calculate a participant-based mean ($SW{\cdot}m^{-1}$). Traditional inverse dynamics were used to calculate net joint powers in Visual3D. Joint work was calculated using the trapezoid rule, finding the area under the curve between negative and positive joint power for the hip, knee, and ankle. This was calculated for the same frames used for IMU metrics and summed across joints to calculate total limb negative work normalized to system weight ($J{\cdot}kg^{-1}$) during the same frames analysed for IMU data.

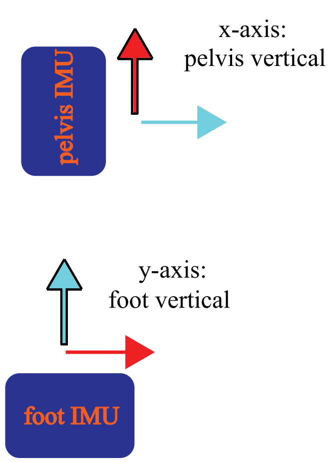

Fig. 1. Orientation of foot and pelvis IMUs and the associated axes that were used to define vertical acceleration for each IMU.

2.5. Physiological cost

Equations initially developed by Pandolf et al. and corrections by Santee et al. were used to calculated estimated metabolic cost (EMC) for all load and grade conditions [26,27]. These equations (Eqs. 4 and 5) have been shown to accurately estimate physiological cost, with a predictive error of 7–11 % for a 25 kg load and speeds of 4.8 and 5.5 kph, which are within 6 and 14 % of the current study's locomotor speed, respectively [9]. Flat and uphill walking was calculated using body mass (M_{bw}, kg), total mass of body mass and additional 23 kg load (M_t, kg), additional 23 kg load (L), locomotor velocity (V, $m{\cdot}s^2$), treadmill grade (G, %).

$$W_{Flat/Uphill} = 1.5M_{bw} + 2.0(M_t)(L/M_{bw})^2 + (M_t)(1.5V^2 + 0.35VG) \tag{4}$$

$$W_{down} = (W_{Flat/Uphill}) - G\left(\frac{M_t V}{3.5}\right) - \left(\frac{M_t (G+6)^2}{M_{bw}}\right) + 25 - V^2 \tag{5}$$

Statistical Analyses: A repeated measures ANOVA was used to explore the main and interaction effects of speed, grade, and load on: vertical signal attenuation per step ($m/s^2{\cdot}kg^{-1}$) and per kilometre ($m/s^2{\cdot}kg^{-1}{\cdot}km^{-1}$), relative attenuation (percentage), foot and pelvis peak vertical accelerations ($m/s^2{\cdot}kg^{-1}$), and estimated physiological cost (Watts). Where there were significant interactions or main effects, simple pairwise comparisons with a Bonferroni correction were used. Cohen's D was used to measure effect size [28] and was considered trivial ($d < 0.2$), small ($d < 0.5$), moderate ($d < 0.8$) or large ($d \geq 0.8$) [29]. Simple linear regressions measured the relationship between OMC variables (*limb negative work, vertical stiffness*) and: relative attenuation ($vAtten_{rel}$), attenuation ($vAtten_{sig}$), and pelvis accelerations ($vAccel_{pelvis}$). Simple linear regressions were also used to measure the relationship between the independent variables of 1) attenuation ($vAtten_{sig}$) and 2) attenuation per kilometre ($vAtten_{km}$), and the dependent variables of 1) EMC and 2) limb negative work. Relationships were defined as: Very strong (≥ 0.8), strong ($0.8 > r \geq 0.6$), moderate ($0.6 > r \geq 0.4$), weak ($0.4 > r \geq 0.2$), very weak ($r < 0.2$) [30]. For all analyses, *a priori* was defined as ≤ 0.05. All analyses were conducted using SPSS (IMB, v29.0.0).

3. Results

3.1. Attenuation per step

All main and interaction effects, associated p values and Cohen's D effect sizes for $vAtten_{Sig}$ are presented in Table 1. Interactions for load and speed, load and grade, and speed and grade were observed. $vAtten_{Sig}$ was higher with load and in walking compared to unloaded and running in main effects analyses with large and small effects, respectively. However, separating by speeds revealed that load reduced $vAtten_{Sig}$ while walking, while the inverse was true for running with $vAtten_{Sig}$ increasing with the addition of load, both with large effect sizes. When separating by load, walking only had higher $vAtten_{Sig}$ than running in an unloaded condition with a large effect.

The addition of load only increased $vAtten_{Sig}$ uphill with large effects. Though $vAtten_{Sig}$ did not differ between grades in the unloaded condition, downhill $vAtten_{Sig}$ was lower than both uphill and flat with load with large effects. Walking had greater $vAtten_{Sig}$ than running in flat and downhill grades with large effects, while the speeds did not differ in $vAtten_{Sig}$ in uphill. In walking, flat had greater $vAtten_{Sig}$ than uphill with a large effect, while uphill had greater $vAtten_{Sig}$ than both flat and downhill in running with moderate and large effects, respectively.

3.2. Attenuation per kilometer

All main and interaction effects, associated p values and Cohen's D effect sizes for $vAtten_{Km}$ are presented in Table 2. $vAtten_{km}$ was higher for unloaded compared to loaded when only the load main effect was considered. Separating data by speed demonstrated that the addition of load reduced $vAtten_{km}$ in walking with a large effect, but not running.

Walking had higher $vAtten_{km}$ compared to running based on a speed main effect, but the effect of speed was large for unloaded and trivial for loaded when separating data by load. Walking had higher $vAtten_{km}$ than running in all grades with large effects.

There was no significant grade main effect, but separating data by speed and load revealed differences. Uphill walking had lower $vAtten_{km}$ than both flat and downhill with small effects. Conversely, uphill running had higher $vAtten_{km}$ than both flat and downhill with moderate effects. In the loaded condition, uphill had higher $vAtten_{km}$ than downhill but the effect was trivial.

3.3. Relative attenuation

All main and interaction effects, associated p values and Cohen's D effect sizes for $vAtten_{Rel}$ are presented in Table 3, and visually represented in Fig. 2. There were interaction effects of load and speed, as well as speed and grade. Load increased $vAtten_{rel}$ based on main effects analysis. However, when separating data by speed, the addition of load elevated the $vAtten_{rel}$ while running, but not walking. Considering main effects of speed, running reduced $vAtten_{rel}$. In both load conditions, $vAtten_{rel}$ was higher with large effects in walking compared to running.

Main effects were observed for grade, with $vAtten_{rel}$ significantly higher in downhill compared to uphill with moderate effects. Separating data by speed, walking had a greater $vAtten_{rel}$ in flat compared to uphill, though the effect was small. Uphill had greater $vAtten_{rel}$ than both flat and downhill while running, with moderate and large effects, respectively. Considering grades independently, $vAtten_{rel}$ was higher in walking than running for all grades, all with large effects.

Table 1
Mean (SD) of vertical acceleration at the foot ($vAtten_{Sig}$), and significant main and interaction effects, associated p-values (P) and effect sizes (Cohen's D).

			$vAtten_{Sig}^{a, b, c, a*b, a*c, b*c}$		
Unloaded	**Walk**	**Flat**	424.98 (146.62)		
		Up	357.64 (159.42)		
		Down	426.55 (139.89)		
	Run	**Flat**	60.20 (285.15)		
		Up	133.79 (329.30)		
		Down	55.34 (248.76)		
Loaded	**Walk**	**Flat**	331.37 (101.92)		
		Up	293.50 (112.87)		
		Down	329.12 (90.19)		
	Run	**Flat**	248.47 (255.23)		
		Up	370.28 (243.91)		
		Down	169.94 (226.70)		
Effect	**P**	**Condition**	**Pairwise Comparison**	**P**	**Cohen's D**
Load	0.044				0.98
Speed	0.005				0.44
Grade	0.001		Uphill, Downhill	0.018	0.93
Load * Speed	< 0.001	Unloaded	Walking, Running	< 0.001	4.27
		Walking	Unloaded, Loaded	< 0.001	2.07
		Running		0.002	2.03
Load * Grade	0.019	Uphill	Unloaded, Loaded	0.019	10.78
		Loaded	Flat, Downhill	0.037	0.81
			Uphill, Downhill	0.004	1.63
Speed * Grade	< 0.001	Flat	Walking, Running	0.003	3.28
		Uphill		< 0.001	1.07
		Walking	Flat, Uphill	0.004	1.20
		Running	Flat, Uphill	0.002	0.50
			Uphill, Downhill	< 0.001	1.69

Effect of load[a], speed[b], grade[c]

Table 2
Mean (SD) of vertical acceleration at the foot ($vAtten_{Km}$), and significant main and interaction effects, associated p-values (P) and effect sizes (Cohen's D).

			$vAtten_{Km}^{a, b, a*b, a*c, b*c}$		
Unloaded	**Walk**	**Flat**	308431.89 (111024.19)		
		Up	249948.59 (10551.872)		
		Down	316564.84 (103467.89)		
	Run	**Flat**	40547.44 (174224.45)		
		Up	91344.81 (199797.09)		
		Down	54333.18 (164630.12)		
Loaded	**Walk**	**Flat**	179176.86 (59843.85)		
		Up	160281.51 (63148.15)		
		Down	176448.99 (52483.52)		
	Run	**Flat**	111348.82 (118859.29)		
		Up	158524.46 (102744.86)		
		Down	81403.25 (121035.60)		
Effect	**P**	**Condition**	**Pairwise Comparison**	**P**	**Cohen's D**
Load	0.017				0.86
Speed	< 0.001				2.84
Load * Speed	< 0.001	Unloaded	Walking, Running	< 0.001	2.93
		Loaded		0.033	0.17
Load * Grade	0.001	Flat	Unloaded, Loaded	< 0.001	1.84
		Uphill		0.040	0.88
		Downhill		< 0.001	1.96
		Loaded	Uphill, Downhill	0.025	0.12
Speed * Grade	< 0.001	Walking	Uphill, Flat	0.004	0.54
			Uphill, Downhill	0.002	0.63

Effect of load[a], speed[b], grade[c]

Table 3
Significant main and interaction effects, associated p-values (P) and effect sizes (Cohen's D) of relative attenuation ($vAtten_{Rel}$) between the foot and the pelvis IMUs.

Effect	P-value	Condition	Pairwise Comparison	P-value	Cohen's D
Load	< 0.001				1.58
Speed	< 0.001				3.72
Grade	< 0.001		Uphill, Downhill	0.001	0.71
Load * Speed	<0.001	Running	Unloaded, Loaded	<0.001	2.11
		Unloaded	Walking, Running	<0.001	4.51
		Loaded		<0.001	2.55
Speed * Grade	< 0.001	Walking	Flat, Uphill	0.007	0.22
		Running	Flat, Uphill	0.005	0.57
			Uphill, Downhill	<0.001	0.84
		Flat	Walking, Running	<0.001	2.26
		Uphill		0.002	1.47
		Downhill		<0.001	2.45

3.4. *Vertical foot acceleration*

All main and interaction effects, associated p values and Cohen's D effect sizes for $vAccel_{Foot}$ are presented in Table 4. An interaction effect was observed between speed and grade. A main effect to of load was observed, with $vAccel_{Foot}$ lower in the loaded condition compared to unloaded with small effects. When considering main effect of speed, $vAccel_{Foot}$ were higher in running compared to walking, though the effect was small. Considering speeds independently, $vAccel_{Foot}$ were significantly higher than walking for all grades. In walking, uphill $vAccel_{Foot}$ were lower than both flat and downhill but with small effects, while flat and downhill did not significantly differ. $vAccel_{Foot}$ were higher running uphill compared to flat with small effects, while downhill did not differ from flat or uphill.

3.5. *Vertical pelvis acceleration*

All main and interaction effects, associated p values and Cohen's D effect sizes for $vAccel_{Pelvis}$ are presented in Table 5. Interaction effects were observed for load and speed as well as speed and grade. The addition of load lowered $vAccel_{Pelvis}$ with large effects, and running $vAccel_{Pelvis}$ were greater than walking with large effects. When separating data by speed, load lowered $vAccel_{Pelvis}$ for both walking and running with small and large effects, respectively. Running $vAccel_{Pelvis}$ were higher than walking when separating data by load conditions.

Main effect analysis of grade revealed downhill $vAccel_{Pelvis}$ were higher than flat and uphill with small and moderate effects, respectively. Flat $vAccel_{Pelvis}$ were higher than uphill with small effects. Running $vAccel_{Pelvis}$ were higher than walking all grades when grades were examined independently, all with large effects. When separating data by speed, downhill $vAccel_{Pelvis}$ were higher than both flat and uphill while walking, but with trivial effects. However, uphill and flat did not differ. In running, uphill $vAccel_{Pelvis}$ were lower than both flat and downhill, with small and moderate effects, respectively. $vAccel_{Pelvis}$ were not different in flat compared to downhill.

3.6. *Regression analyses*

Regression analyses (Table 6) revealed moderate positive relationships between the dependent variable of limb negative work and independent variable $vAtten_{Sig}$ in flat walking and running and uphill running. In the loaded condition, $vAtten_{Sig}$ in uphill walking, and downhill running had moderate relationships to limb negative work. $vAtten_{Sig}$ had a moderate relationship with vertical stiffness un unloaded running in all grades, and only downhill loaded walking.

Regression between $vAtten_{sig}$ and EMC are presented in Fig. 3. There was a moderate negative relationship between EMC and $vAtten_{sig}$ in flat and uphill walking in both unloaded and loaded conditions. A moderate positive relationship was observed between EMC and $vAtten_{sig}$ for unloaded running at all grades

3.7. *Post-hoc analysis*

Supplemental RM ANOVA was completed on EMC and is presented in supplementary files.

4. Discussion

In general, peak acceleration and signal attenuation quantified by foot and pelvis IMUs are sensitive to demands of military-specific tasks. Accelerations at the foot and pelvis both increase with speed but decrease with the addition of load. While both foot and pelvis accelerations are sensitive to changes in grade while walking, only pelvic accelerations significantly differ between nearly all grades while running. Both signal attenuation and attenuation per kilometre have a consistent relationship with EMC. Specifically, higher values of signal attenuation in walking are related to reduced EMC, whereas the inverse is true for running – an increase in attenuation is associated with increased EMC, opposed to a moderate relationship between attenuation and lower extremity OMC variables.

Signal attenuation increased by 14 % with the addition of load, but load reduced both foot and pelvis accelerations by 24 % and 50 %, respectively. These changes may be a consequence of altered

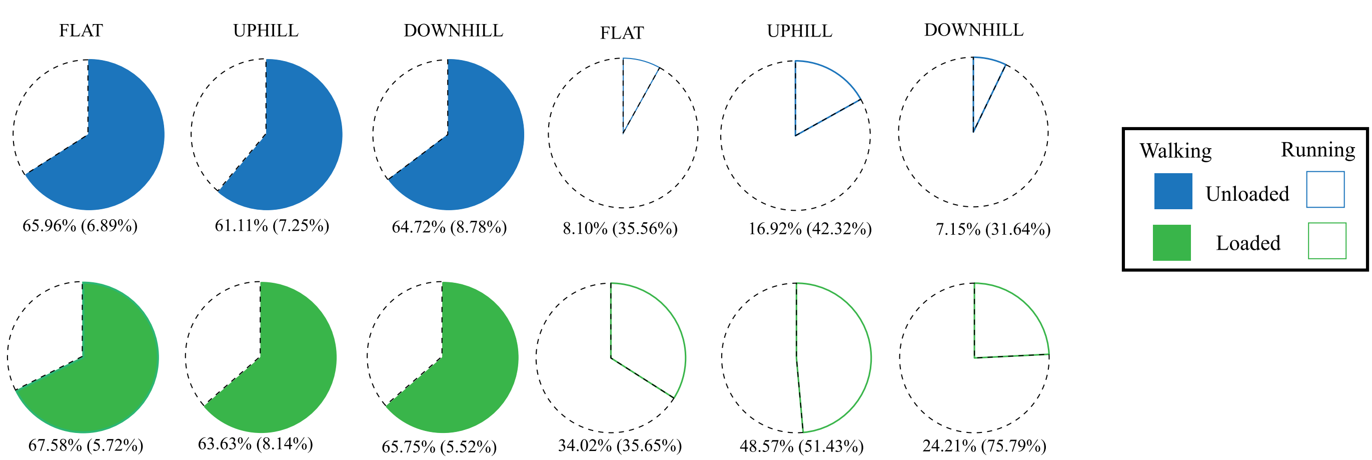

Fig. 2. Relative Attenuation ($vAtten_{rel}$) of all loads, speeds, and grades represented as Mean % (SD).

Table 4
Mean (SD) of vertical acceleration at the foot ($vAccel_{Foot}$), and significant main and interaction effects, associated p-values (P) and effect sizes (Cohen's D).

			$vAccel_{Foot}^{a,\ b,\ b*c}$		
Unloaded	**Walk**	**Flat**	633.77 (170.94)		
		Up	567.64 (186.25)		
		Down	650.14 (156.11)		
	Run	**Flat**	775.53 (169.31)		
		Up	816.38 (158.93)		
		Down	802.57 (198.48)		
Loaded	**Walk**	**Flat**	483.41 (116.84)		
		Up	447.75 (132.56)		
		Down	496.58 (110.43)		
	Run	**Flat**	657.00 (179.13)		
		Up	729.20 (212.80)		
		Down	631.18 (125.15)		
Effect	**P**	**Condition**	**Pairwise Comparison**	**P**	**Cohen's D**
Load	< 0.001				0.29
Speed	< 0.001				0.33
Speed * Grade	< 0.001	Flat	Walking, Running	<0.001	1.22
		Uphill		<0.001	2.06
		Downhill		<0.001	1.11
		Walking	Flat, Uphill	0.003	0.40
			Uphill, Downhill	0.018	0.51
		Running	Flat, Uphill	0.006	0.44

Effect of load[a], speed[b], grade[c]

Table 5
Mean (SD) of vertical acceleration at the foot ($vAccel_{Pelvis}$), and significant main and interaction effects, associated p-values (P) and effect sizes (Cohen's D).

			$vAccel_{Pelvis}^{a,\ b,\ c,\ a*b,\ b*c}$		
Unloaded	**Walk**	**Flat**	208.79 (43.82)		
		Up	210.01 (39.78)		
		Down	223.59 (52.58)		
	Run	**Flat**	715.33 (328.09)		
		Up	682.59 (354.78)		
		Down	747.23 (298.08)		
Loaded	**Walk**	**Flat**	152.04 (24.43)		
		Up	154.25 (27.18)		
		Down	167.46 (33.02)		
	Run	**Flat**	408.53 (185.24)		
		Up	358.92 (147.30)		
		Down	461.24 (183.99)		
Effect	**P-value**	**Condition**	**Pairwise Comparison**	**P-value**	**Cohen's D**
Load	< 0.001				4.49
Grade	< 0.001		Flat, Uphill	0.019	0.25
			Flat, Downhill	0.012	0.36
			Uphill, Downhill	<0.001	0.60
Load * Speed	<0.001	Walking	Unloaded, Loaded	<0.001	0.44
		Running		<0.001	2.38
		Unloaded	Walking, Running	<0.001	3.90
		Loaded		<0.001	1.96
Speed * Grade	< 0.001	Walking	Flat, Downhill	0.005	0.10
			Uphill, Downhill	0.042	0.09
		Running	Flat, Uphill	0.025	0.28
			Uphill, Downhill	0.004	0.56
		Flat	Walking, Running	<0.001	2.58
		Uphill		<0.001	2.23
		Downhill		<0.001	2.76

Effect of load[a], speed[b], grade[c]

spatiotemporal parameters that occur due to load. While literature reports load-induced reductions of step and stride length, the current participants increased stride length with the addition of load – similar to what occurs following fatiguing protocols [31,32]. In unloaded running, it's reported that increasing stride length by 15 % increases signal attenuation [33]. As such, adopting a longer stride may have increased the ability of individuals to attenuate the signal between the foot and the pelvis. It is not immediately clear why load reduced foot and pelvis accelerations, especially as it is reported that both increasing stride length and the addition of load, as performed by the current participants, increases IMU accelerations – particularly at the distal limb [33–35].

As attenuation per step and per kilometre increased in walking in unloaded and loaded conditions and in both flat and uphill, there was a reduction in EMC. This implies that participants who attenuated more of the impact accelerations during loaded walking were the most economical. It may be that participants who were able to attenuate the forces adequately adapted lower extremity biomechanics (e.g., joint motions, muscle activation patterns) to execute the flat and uphill walking tasks, both without and with load, to optimize their walking economy [36]. Interestingly, the opposite relationship was observed in running. Specifically, increased EMC was associated with increased attenuation for unloaded running in all grades. The relationship between unloaded running and EMC is in line with what has been found when inducing fatigue in a running protocol. Fatigue reduces the ability of individuals to attenuate the acceleration signal between the distal limb and the head, as well as increases oxygen consumption [31]. However, the lack of a similar relationship was not observed with the addition of load. Load increases signal attenuation while running, which may be indicative of the reported attempt of individuals to maintain centre of mass vertical oscillation while running by increasing limb stiffness. While a stiff limb may be indicative of reduced attenuation, the addition of load reportedly increases joint flexion at initial contact but results in stiffer joints during stance [7,8,37]. As such, participants may have landed with a more flexed limb, attenuating more of the force up the limb during weight acceptance. As such, those adept at attenuation may have incurred increased physiological cost by actively attenuating higher forces that occur as a result of load carriage [6,8].

In opposition to what was observed with the addition of load, increasing speed from walking to running increased foot accelerations by nearly 75 % and pelvic accelerations by a factor of three. This was paired with reduced attenuation per step and kilometre by 52 % and 62 %, respectively, when going from walking to running. The reduction in attenuation with speed is unsurprising given it is reported that there is a reduction in the amount of shock attenuated as an individual's speed increases [34]. The foot was sensitive to grade while walking, with uphill having significantly lower accelerations than both flat and uphill, but only uphill was significantly lower than flat while running. This is in line with previous literature reporting tibial accelerations in downhill and uphill running [38,39]. In fact, tibial shock reportedly increases by 30 % in a nearly −9 % downhill grade [40]. The current participants saw a change in attenuation per step of ± 10 % in both uphill (increase) and downhill (decrease). Given the −6 % grade employed in the current study, it stands to reason that the altered attenuation profiles would be similar to these changes in tibial accelerations previously reported by Hamill et al. [40]. When examining attenuation per kilometre, uphill attenuation per kilometre was nearly a fifth lower in uphill walking compared to flat and downhill. An uphill grade of 10 % reportedly increases walking cadence 7 %, while flat and downhill only differ by 3 % [41]. An elevated cadence in an uphill walking may reduce the amount attenuated over a prescribed distance compared to flat and downhill cadences. Conversely, uphill running increases attenuation compared to flat and downhill, but flat and downhill do not differ. In general, uphill running results in individuals presenting with greater measures of joint flexion at initial contact, which may increase one's ability to attenuate the signal when compared to flat and downhill running [38]. If practitioners are aware of the grade profile of the intended activity (e.g., prescribed march or run), they may be able to adequately understand the physical demands being placed on individuals and how it differs on terrain over the course of a long-duration task. This may enable refinement of training prescription to further minimize risk of injury by putting an individual under deleterious task demands.

The local axes used on the IMUs to define vertical is susceptible to

Table 6
Simple linear regression between the independent variable of attenuation per step (vAttensig) and dependent variables of: vertical stiffness, limb negative work, and estimated metabolic cost (EMC).

			Vertical Stiffness	Limb Negative Work	EMC
			R Value		
Unloaded	***Walking***	***Flat***	0.25	**0.39**	−0.52
		Up	0.18	0.13	−0.58
		Down	−0.04	0.00	−0.24
	Running	***Flat***	**0.40**	**0.50**	**0.44**
		Up	**0.37**	**0.46**	**0.40**
		Down	**0.40**	0.05	**0.41**
Loaded	***Walking***	***Flat***	−0.05	0.24	−0.53
		Up	−0.30	**0.45**	−0.55
		Down	**0.47**	0.32	−0.28
	Running	***Flat***	0.01	0.26	−0.03
		Up	0.02	0.14	−0.15
		Down	0.02	**0.83**	0.04

Bold is indicative of at least a moderate linear relationship.

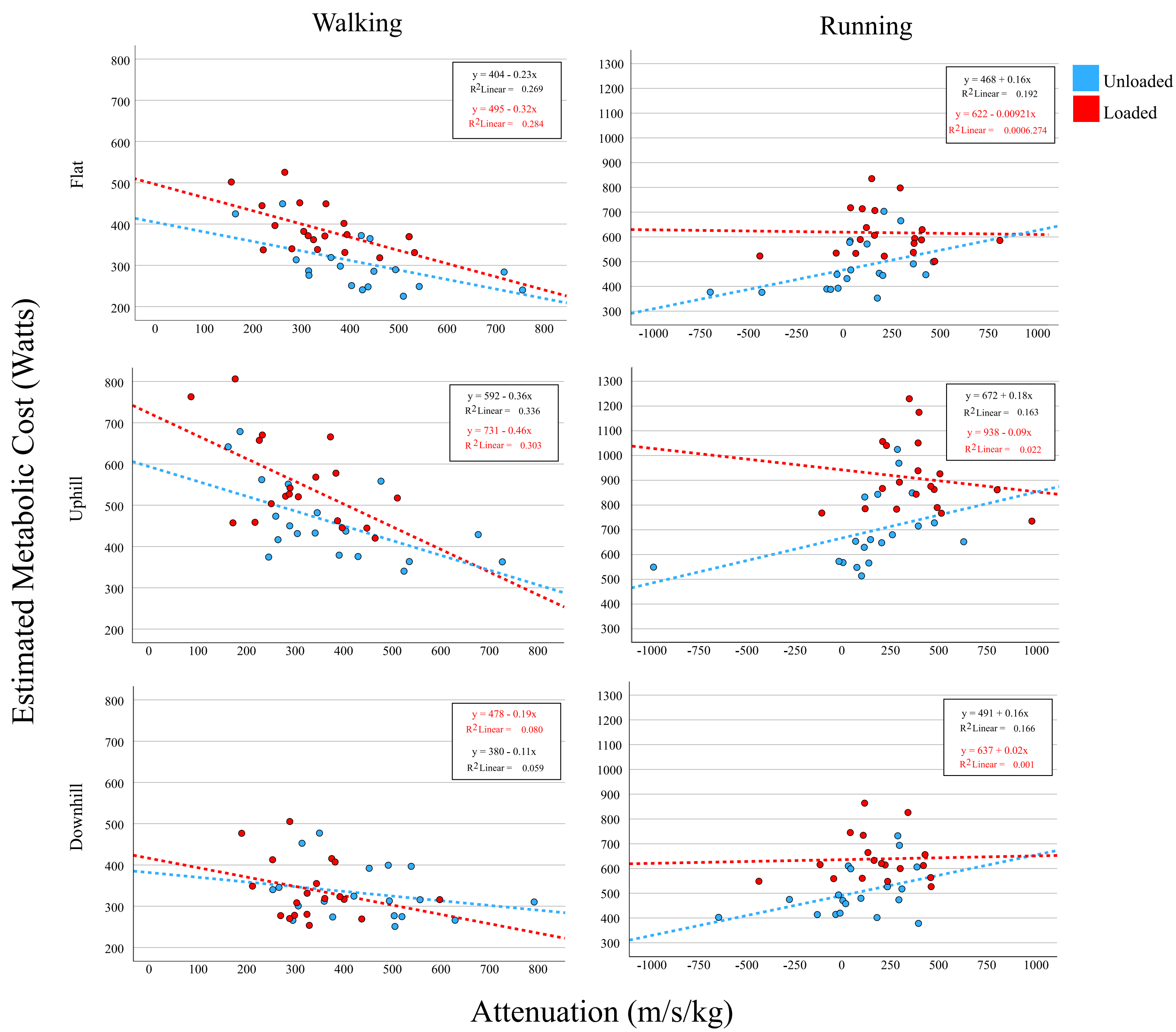

Fig. 3. Linear regression of the relationship between attenuation (x-axis; m/s/kg) and Estimated Metabolic Cost (y-axis; Watts) for walking and running, with and without load for flat, uphill and downhill.

segment orientation in space. While the vertical axis of the IMU is in line with the loading forces that act up the kinetic chain and in line with previous literature, there are accelerations that also occur out-of-plane. As such, altered or lowered accelerations that occurred due to load and/or grade may have been a result of accelerations shifting distribution to other planes of motion. The use of the EMC equation to examine the relationship between physiological cost and IMU acceleration metrics has limitations insofar that physiological cost was not directly measured. While these equations have been previously validated and employed, it is a metric that relies on prediction. Consequently, interpretation of regression results between EMC and IMU metrics should be done with caution as not all individuals will fall within the assumption parameters of these EMC equations. Future literature should look to examine measured physiological cost to IMU metrics, including attenuation.

Foot and pelvis IMUs accelerations and signal attenuation between these sensors, are altered with changes to speed and the addition of load. Attenuation between a distally placed IMU and an IMU on the pelvis may be indicative of negative limb work and estimated physiological cost in unloaded locomotion. Given that loaded ruck marches are cited as the leading-most mechanism of injury in a training scenario, practical application of this information enables appropriate task prescription to minimize the risk of injury. Specifically, it may be that individuals are capable of attenuating elevated forces that occur during load carriage in a single step, but altered spatiotemporal profiles result in reduced attenuation over time (e.g., over a kilometre). It may be that load reduces cumulative attenuation over time, resulting in what may be injurious transfer of energy up the kinetic chain as the duration of the loaded task increases. This is particularly of interest in scenarios wherein practitioners seek to manage or mitigate risk of injury that occurs as a consequence of total load over time opposed to the magnitude of attenuation per step. The use of two commercially available wearable sensors can supply practitioners with estimated physical demands being placed on individuals, and how these demands may change during a long duration course with variable grade and speed, both with and without required body borne load.

CRediT authorship contribution statement

Tim Doyle: Writing – review & editing, Supervision, Resources, Methodology, Funding acquisition, Conceptualization. **Jodie Wills:** Writing – review & editing, Supervision, Resources, Methodology, Funding acquisition, Conceptualization. **AuraLea Fain:** Writing – review & editing, Writing – original draft, Project administration, Methodology, Investigation, Formal analysis, Conceptualization. **Brad Nindl:** Writing – review & editing, Supervision, Conceptualization. **Joel Fuller:** Writing – review & editing, Supervision, Methodology, Formal analysis. **Ayden McCarthy:** Writing – review & editing, Methodology, Investigation, Conceptualization.

Declaration of Competing Interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests: Dr. Timothy L.A. Doyle reports financial support was provided by Office of Naval Research Global. If there are other authors, they declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

This research was funded by the Office of Naval Research Global (N62909–21–1–2015).

Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.gaitpost.2024.08.003.

References

[1] P.D. Grimm, T.C. Mauntel, B.K. Potter, Combat and noncombat musculoskeletal injuries in the US military, Sports Med Arthrosc. Rev. 27 (3) (2019) 84–91. ⟨https://www.ncbi.nlm.nih.gov/pubmed/31361716⟩.
[2] I. Dijksma, W.O. Zimmermann, E.J. Hertenberg, C. Lucas, M.M. Stuiver, One out of four recruits drops out from elite military training due to musculoskeletal injuries in the Netherlands Armed Forces, BMJ Mil. Health 168 (2) (2022) 136–140. ⟨https://www.ncbi.nlm.nih.gov/pubmed/32139408⟩.
[3] U. Gottlieb, D. Kelman, S. Springer, Evaluation of two simple functional tests to predict attrition from combat service in female light infantry soldiers, Med Sci. Monit. 24 (2018) 9334–9341. ⟨https://www.ncbi.nlm.nih.gov/pubmed/30578629⟩.
[4] J.L. Tait, J.R. Drain, S. Bulmer, P.B. Gastin, L.C. Main, Factors predicting training delays and attrition of recruits during basic military training, Int J. Environ. Res Public Health 19 (12) (2022). ⟨https://www.ncbi.nlm.nih.gov/pubmed/35742522⟩.
[5] D.J. Walsh, G. Palevo, M. Polascik, J. Slaton, Physiological differences of US army cadets during a loaded and unloaded 6-mile ruck march, J. Exerc. Physiol. Online 23 (1) (2020) 79–87.
[6] S.A. Birrell, R.H. Hooper, R.A. Haslam, The effect of military load carriage on ground reaction forces, Gait Posture 26 (4) (2007) 611–614. ⟨https://www.ncbi.nlm.nih.gov/pubmed/17337189⟩.
[7] T.N. Brown, M. O'Donovan, L. Hasselquist, B.D. Corner, J.M. Schiffman, Body borne loads impact walk-to-run and running biomechanics, Gait Posture 40 (1) (2014) 237–242. ⟨https://www.ncbi.nlm.nih.gov/pubmed/24794647⟩.
[8] N.J. Lobb, A.C. Fain, K.D. Seymore, T.N. Brown, Sex and stride length impact leg stiffness and ground reaction forces when running with body borne load, J. Biomech. 86 (2019) 96–101. ⟨https://www.ncbi.nlm.nih.gov/pubmed/30755323⟩.
[9] C.A.J. Vine, S.L. Coakley, S.D. Blacker, J. Doherty, B.J. Hale, E.F. Walker, et al., Accuracy of Metabolic Cost Predictive Equations During Military Load Carriage, J. Strength Cond. Res 36 (5) (2022) 1297–1303. ⟨https://www.ncbi.nlm.nih.gov/pubmed/32398631⟩.
[10] D.P. Looney, W.R. Santee, E.O. Hansen, P.J. Bonventre, C.R. Chalmers, A.W. Potter, Estimating energy expenditure during level, uphill, and downhill walking, Med Sci. Sports Exerc 51 (9) (2019) 1954–1960. ⟨https://www.ncbi.nlm.nih.gov/pubmed/30973477⟩.
[11] P.P. Carden, R.M. Izard, J.P. Greeves, J.P. Lake, S.D. Myers, Force and acceleration characteristics of military foot drill: implications for injury risk in recruits, BMJ Open Sport Exerc Med 1 (1) (2015). ⟨https://www.ncbi.nlm.nih.gov/pubmed/27900110⟩.
[12] C.D. Johnson, L.K. Sara, M.M. Bradach, D.R. Mullineaux, S.A. Foulis, J.M. Hughes, et al., Relationships between tibial accelerations and ground reaction forces during walking with load carriage, J. Biomech. 156 (2023) 111693. ⟨https://www.ncbi.nlm.nih.gov/pubmed/37406568⟩.
[13] A.S. Tenforde, T. Hayano, S.T. Jamison, J. Outerleys, I.S. Davis, Tibial Acceleration Measured from Wearable Sensors Is Associated with Loading Rates in Injured Runners, PM R. 12 (7) (2020) 679–684. ⟨https://www.ncbi.nlm.nih.gov/pubmed/31671242⟩.
[14] M. Baggaley, M. Esposito, C. Xu, G. Unnikrishnan, J. Reifman, W.B. Edwards, Effects of load carriage on biomechanical variables associated with tibial stress fractures in running, Gait Posture 77 (2020) 190–194. ⟨https://www.ncbi.nlm.nih.gov/pubmed/32058282⟩.
[15] H. Rice, J. Fallowfield, A. Allsopp, S. Dixon, Influence of a 12.8-km military load carriage activity on lower limb gait mechanics and muscle activity, Ergonomics 60 (5) (2017) 649–656. ⟨https://www.ncbi.nlm.nih.gov/pubmed/27462759⟩.
[16] M.A. Sloniger, K.J. Cureton, B.M. Prior, E.M. Evans, Lower extremity muscle activation during horizontal and uphill running (1985), J. Appl. Physiol. 83 (6) (1997), ⟨https://www.ncbi.nlm.nih.gov/pubmed/9390983⟩, 2073-9.
[17] A.W. Potter, D.P. Looney, L.A. Blanchard, A.P. Welles, W.R. Santee, Accuracy of predictive equations for metabolic cost of locomotion while carrying external load, J. Sport Hum. Perform. 5 (1) (2017).
[18] C. Napier, R.W. Willy, B.C. Hannigan, R. McCann, C. Menon, The effect of footwear, running speed, and location on the validity of two commercially available inertial measurement units during running, Front Sports Act. Living 3 (2021) 643385. ⟨https://www.ncbi.nlm.nih.gov/pubmed/33981991⟩.
[19] J. Reenalda, E. Maartens, J.H. Buurke, A.H. Gruber, Kinematics and shock attenuation during a prolonged run on the athletic track as measured with inertial magnetic measurement units, Gait Posture 68 (2019) 155–160. ⟨https://www.ncbi.nlm.nih.gov/pubmed/30481697⟩.
[20] L.C. Benson, A.M. Raisanen, C.A. Clermont, R. Ferber, Is this the real life, or is this just laboratory? A scoping review of IMU-based running gait analysis, Sens. (Basel) 22 (5) (2022). ⟨https://www.ncbi.nlm.nih.gov/pubmed/35270869⟩.
[21] J.H. Williams, S.F. Rizzuto, B.A. Dillard, E.J. Whitaker, Complementary Use of Wearable Technology 4: Assessing Gait Asymmetry and Shock Attenuation Using Multiple IMU Devices, (2022).

[22] A.M. Jones, J.H. Doust, A 1% treadmill grade most accurately reflects the energetic cost of outdoor running, J. Sports Sci. 14 (4) (1996) 321–327. ⟨https://www.ncbi.nlm.nih.gov/pubmed/8887211⟩.

[23] Gea Wu, ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion–part I: ankle, hip, and spine. International Society of Biomechanics, J. Biomech. 35 (4) (2002) 543–548. ⟨https://www.ncbi.nlm.nih.gov/pubmed/11934426⟩.

[24] M.H. Schwartz, A. Rozumalski, A new method for estimating joint parameters from motion data, J. Biomech. 38 (1) (2005) 107–116. ⟨https://www.ncbi.nlm.nih.gov/pubmed/15519345⟩.

[25] J.-B. Morin, G. Dalleau, H. Kyröläinen, T. Jeannin, A. Belli, A simple method for measuring stiffness during running, J. Appl. Biomech. 21 (2) (2005) 167–180.

[26] K.B. Pandolf, B. Givoni, R.F. Goldman, Predicting energy expenditure with loads while standing or walking very slowly, J. Appl. Physiol. 43 (4) (1977) 577–581.

[27] W.R. Santee, L.A. Blanchard, K.L. Speckman, J.A. Gonzalez, R.F. Wallace, Load carriage model development and testing with field data, Natick, MA: US Army Research Institute of Environmental Medicine ADA 415788 (2003).

[28] J. Cohen, Statistical power analysis for the behavioral sciences, Academic press, 2013.

[29] D. Lakens, Calculating and reporting effect sizes to facilitate cumulative science: a practical primer for t-tests and ANOVAs, Front Psychol. 4 (2013) 863. ⟨https://www.ncbi.nlm.nih.gov/pubmed/24324449⟩.

[30] W.G. Hopkins, Linear models and effect magnitudes for research, clinical and practical applications, Sportscience 14 (2010) 49–59.

[31] J.A. Mercer, B.T. Bates, J.S. Dufek, A. Hreljac, Characteristics of shock attenuation during fatigued running, J. Sports Sci. 21 (11) (2003) 911–919. ⟨https://www.ncbi.nlm.nih.gov/pubmed/14626370⟩.

[32] V.G. Bode, P.N. Frykman, N.I. Smith, R.E. Fellin, J.F. Seay, Spatiotemporal and kinematic comparisons between anthropometrically paired male and female soldiers while walking with heavy loads, Mil. Med. 186 (3-4) (2021) 387–392. ⟨https://www.ncbi.nlm.nih.gov/pubmed/33215681⟩.

[33] J.A. Mercer, P. Devita, T.R. Derrick, B.T. Bates, Individual effects of stride length and frequency on shock attenuation during running, Med Sci. Sports Exerc 35 (2) (2003) 307–313. ⟨https://www.ncbi.nlm.nih.gov/pubmed/12569221⟩.

[34] J.A. Mercer, J. Vance, A. Hreljac, J. Hamill, Relationship between shock attenuation and stride length during running at different velocities, Eur. J. Appl. Physiol. 87 (4-5) (2002) 403–408. ⟨https://www.ncbi.nlm.nih.gov/pubmed/12172880⟩.

[35] H. Rice, O. Seynnes, A. Werkhausen, Effect of increased running speed and weight carriage on peak and cumulative tibial loading, Scand. J. Med Sci. Sports 33 (12) (2023) 2516–2523. ⟨https://www.ncbi.nlm.nih.gov/pubmed/37642291⟩.

[36] P.C. Raffalt, M.K. Guul, A.N. Nielsen, S. Puthusserypady, T. Alkjaer, Economy, Movement Dynamics, and Muscle Activity of Human Walking at Different Speeds, Sci. Rep. 7 (2017) 43986. ⟨https://www.ncbi.nlm.nih.gov/pubmed/28272484⟩.

[37] T.N. Brown, A.C. Fain, K.D. Seymore, N.J. Lobb, Sex and stride impact joint stiffness during loaded running, J. Appl. Biomech. 37 (2) (2021) 95–101. ⟨https://www.ncbi.nlm.nih.gov/pubmed/33361491⟩.

[38] G. Vernillo, M. Giandolini, W.B. Edwards, J.B. Morin, P. Samozino, N. Horvais, et al., Biomechanics and physiology of uphill and downhill running, Sports Med 47 (4) (2017) 615–629. ⟨https://www.ncbi.nlm.nih.gov/pubmed/27501719⟩.

[39] R. Di Michele, S. Fantozzi, S. Bartolomei, G. Semprini, F. Stella, S. Ciacci, Effects of fatigue on kinematics and shock attenuation during downhill trail running, ISBS Proc. Arch. 38 (1) (2020) 780.

[40] C. Hamill, Effect of grade running on kinematics and impact force, Med. Sci. Sports Exerc. 16 (1984) 185.

[41] S. Kimel-Naor, A. Gottlieb, M. Plotnik, The effect of uphill and downhill walking on gait parameters: a self-paced treadmill study, J. Biomech. 60 (2017) 142–149. ⟨https://www.ncbi.nlm.nih.gov/pubmed/28757238⟩.