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Enforcing walking speed and step-length affects joint kinematics and kinetics in male and female healthy adults

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ABSTRACT

Background: Individuals increase walking speed by increasing their step-length, increasing their step-frequency, or both. During basic training military recruits are introduced to marching "in-step", and thus the requirement to walk at fixed speeds and step-lengths. The extent to which individuals are required to *under-* or *over-stride* will vary depending on their stature, and the stature of others in their section. The incidence of stress fractures in female recruits undergoing basic training is higher than that for their male counterparts.

Research question:: Therefore, the purpose of this study was to determine how joint kinematics and kinetics are affected by walking speed, step-length, and sex.

Methods: Thirty-seven (19 female) aerobically active non-injured individuals volunteered for this study. Synchronised three-dimensional kinematic and kinetic data were collected while participants walked overground at prescribed speeds. Audio and visual cues were used to control step-lengths. Linear mixed models were run to analyse the effects of speed, step-length condition, and sex on peak joint moments.

Results and Significance: The findings of this study showed that, in general, walking faster and *over-striding* predominantly increased peak joint moments, suggesting that *over-striding* is more likely to negatively affect injury risk than *under-striding*. This is especially important for individuals unaccustomed to *over-striding* as the cumulative effect of increased joint moments may affect a muscles capability to withstand the increased external forces associated with walking faster and with longer step-lengths, which could then lead to an increased risk of developing an injury.

1. Introduction

To increase walking speed individuals either increase their step-length, step-frequency, or both. Military recruits undergoing basic training are introduced to marching "in-step", and thus the requirement to walk at fixed speeds and step-lengths. In the British Military, female personnel (1.65(0.06) m) are approximately 0.12 m shorter than their male counterparts (1.77(0.07) m) [1]. This suggests female personnel/recruits are more likely to have to *over-stride* when marching; for the shortest female soldier to match the *preferred* step-length of the tallest male soldier, she would theoretically need to *over-stride* by 40.8 %. *Over-striding* has already been identified as a risk factor for pelvis stress fractures [2–4] and the incidence of stress fractures for female recruits undergoing basic training is 1.5–6.7 times higher than that for male recruits [5]. Therefore, understanding how walking with fixed step-lengths, and the extent to which an individual *over-strides* or *under-strides*, affects joint moments is important for mitigating injury risk in those who regularly alter their *preferred* walking gait, i.e., military recruits.

Increased joint moments can indicate an increased injury risk as either fatigued or weak muscles have insufficient capacity to resist these external loads. Participants showed significant increases in hip, knee,

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and ankle moments when they increased their step-length, or both step-length and step-frequency, to walk at fast (self-selected) speeds, compared with their preferred (self-selected) speed [6]. Furthermore, increasing either step-length [7–11] and/or walking speed [8,10,12] increases lower limb joint moments. However, these previous studies considered relatively slow walking speeds ($\leq$1.5 m·s^{-1}) [7,8], did not control walking speed [6,12], or did not consider changes in *preferred* step-lengths [9]; absolute changes to step-length, or changes to "nominal" step-lengths, may result in relatively different effects for different individuals, as *preferred* step-length is related to stature [13]. Therefore, there is a need to understand how altering *preferred* step-lengths affects joint moments at faster walking speeds.

Existing literature report some conflicting results about the effects of changing step-length on joint moments. These differences in results could be due to methodological differences, for example, Lim, et al. [9] imposed changes to "nominal" step-lengths, which may have already required participants to adjust their *preferred* walking biomechanics, whereas other studies investigated relatively small changes to step-length (± 10 %) [8], which may not have elicited meaningful changes in biomechanics. Walking speed also varied between studies (Lim, et al. [9]: 0.89–2.02 m·s^{-1}, Buddhadev, et al. [8]: 1.1–1.5 m·s^{-1}, or Allet, et al. [7]: 1.3 m·s^{-1}), which could have affected the results; as walking speed also affects joint moments [12].

There are relatively limited data on the effects of changing step-length on frontal plane kinetics, with most studies focusing solely on the sagittal plane [7–9]. Also, despite known sex differences in anthropometrics, most studies that included both male and female participants did not compare results between sexes [6–8]. One study found sex significantly affected frontal plane kinematics at the hip and knee, during walking [12], however, corresponding sex differences in joint kinetics were not found [12]. Given the increase in participation rates of walking for leisure [14], as well as the increasing number of female recruits applying to join the British Army, there is a need to understand if large changes in step-lengths elicit sex differences at the joint level.

This study aimed to identify the effects of speed, step-length, and sex on lower limb biomechanics during walking. We hypothesised that 1) joint moments will increase with walking speed and step-length; 2) sex differences will exist in joint moments.

2. Methods

Nineteen female and eighteen male participants volunteered for this study. Participants were recruited from the University of Salford staff, student, and visitor population. Participants were aerobically active healthy non-injured individuals, with no history of lower limb surgery, and aged 18–40 years. The sex-specific mean and standard deviations (SD) of participant demographics are given in Table 1. This study was approved by the Ministry of Defence Research Ethics Committee (XXX) and the University of Salford Ethics Committee (XXX). All participants provided written informed consent prior to participation.

Twenty-eight markers were placed on the lower limbs. Markers were attached to the skin on the iliac crests, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanters, lateral and medial epicondyles, and lateral and medial malleoli. Markers were also placed above the 1st, 2nd, and 5th metatarsals and the posterior, medial and lateral heel calcanei, on the standardised footwear (MAGNUM, Hi-Tec Sports International Holdings BV, The Netherlands) participants wore. In addition, four cluster plates, each with four markers, were attached to the thighs and shanks.

Synchronous, kinematic (100 Hz) and kinetic (2000 Hz) data were collected using 10 VICON T20 optoelectronic motion analysis cameras (VICON, Oxford, UK) and four Kistler force plates (Kistler, Alton, UK) embedded in the laboratory floor. A static calibration trial was collected before the dynamic walking trials, where participants were asked to across a 10 m walkway at three speeds (1.4, 1.6, and 1.8 m•s^{-1} - assured using timing gates (TCi System, Brower Timing Systems LLC, UT, USA)). At each speed, acoustic (metronome) and visual (light projection) cues (projected using an Epson EB-585 W and the Interactive Walkway IWW-v2.05 software [15]), were used to control the participant's walking gait, targeted at ± 0 %, ± 10 %, ± 20 %, and ± 30 % of their *preferred* step-length. The *preferred* step-length was determined using the C-Mill (Motek Medical, The Netherlands) instrumented treadmill in a previous session [16]. Participants were given verbal feedback after each trial; a trial was deemed acceptable if the speed was within 5 % of the target speed and the step-length was within 2.5% of the target step-length. Participants were asked to complete five acceptable trials, up to a maximum of ten trials, per condition. Participants were given as much rest as they wanted between trials.

All overground data were processed with Vicon Nexus (2.8.1, VICON, Oxford, UK) and exported to Visual3D (C-Motion Inc., Maryland, USA) to create the six-degree of freedom model. Kinematic data were gap-filled using spline interpolation (maximum 10 frames), and kinematic and kinetic data were filtered using a low pass 4th order Butterworth filter with a cut off frequency of 6 Hz and 25 Hz, respectively [17,18]. A seven-segment model consisting of the pelvis, thighs, shanks, and feet was created. The pelvis was estimated as a cylinder, and the thighs, shanks, and feet as cones. A virtual foot segment was also created by projecting foot markers onto the floor. Segment masses were estimated based on Dempster, et al. [19], while segment moments of inertia and centre of mass locations were estimated using Hanavan [20]. A CODA pelvis was defined, using ASIS and PSIS markers, and a regression model was used to calculate the hip joint centres [21]. Knee and ankle joint centres were calculated as the midpoint between the lateral and medial epicondyles and malleoli, respectively. Newton-Euler equations and an XYZ Cardan sequence were used to calculate joint kinematics and three-dimensional inverse dynamics were used to calculate net internal joint moments, in the coordinate system of the proximal segment. Joint moments were normalised to body mass with extensor, abductor, external rotator, plantar flexor and evertor moments defined as positive. Ensemble averaging and identification of the peak values of the primary outcome measures were undertaken using custom MATLAB (The MathWorks Inc., Massachusetts, USA) scripts.

Linear mixed models were run in SPSS (IBM SPSS Statistics for Windows, Version 25.0. Armonk, NY: IBM Corp.) to analyse the effect of speed, step-length condition, and sex on peak joint moments during the stance phase of the gait cycle. Speed, step-length condition, sex, stature, and mass were defined as fixed effects; speed, step-length condition, and sex were defined as factors, and stature and mass were defined as covariates. Two models, one where random intercepts (for participants) were assumed and one where random slopes (for speed and step-length condition) were also added, were compared using a Log-Likelihood (LL) test in RStudio (RStudio: Integrated Development for R. Studio, Inc., Version 1.1456, Boston, MA). Random intercepts, and the multi-level nature of the model, allowed us to adjust for systematic differences at the participant level and resolve the non-independence problem of multiple measures from the same participant. Random slopes allow the effect of the speed/step-length condition, on peak joint moments during stance, to differ for each participant. If the inclusion of random slopes did not have a significant effect on the model fit, or if the increased complexity resulted in the model failing to converge, then the random

Table 1
Mean (standard deviation) participant demographics.

	All participants (37)	Female participants (19)	Male participants (18)	Sig. (2-tail)
AGE [yrs.]	27 (6)	27 (6)	28 (7)	0.527
MASS [kg]	70.4 (12.4)	65.4 (13.8)	75.6 (8.1)	**0.011**
STATURE [m]	1.72 (0.08)	1.67 (0.07)	1.78 (0.04)	**< 0.001**

intercept model was used for analysis. A Holm-Bonferroni correction of alpha was performed to guard against type 1 errors.

3. Results

Most participants stated that they found certain step-length/speed combinations difficult, in particular *over-striding* and/or *under-striding* by 30% at 1.8 m·s^{-1}. However, post-processing revealed that a maximum of seven participants were unable to complete any given condition.

The effects of speed, step-length, and sex on peak round reaction forces (GRFs), vertical loading rate, and joint moments are shown in Tables 2–6. A linear mix model with random slopes (LL: $p \leq 0.004$) was used for peak medial GRFs, peak hip flexor and internal rotator moments, and knee extensor and adductor moments, whereas a linear mix model with random intercepts was used for all other variables.

3.1. Speed (after accounting for sex and changes in step-length)

There were significant increases in the vertical loading rate (Table 3) and all peak GRFs (Table 2), except the peak medial GRF when walking at both 1.6 m·s^{-1} and 1.8 m·s^{-1} than when walking at 1.4 m·s^{-1}. All peak joint (except ankle invertor) moments (Tables 4–6), were significantly increased when walking at 1.6 m·s^{-1} and 1.8 m·s^{-1} than when walking at 1.4 m·s^{-1}. Effects were greatest at the faster speed.

Table 3
Linear mixed model outputs for the vertical loading rate.

	Vertical loading rate [BW·s^{-1}]		
	Difference [95%CI]	Sig.	Corrected sig.
speed			
1.6 m·s^{-1}	2.12 [1.30 2.94]	< 0.001	< 0.001
1.8 m·s^{-1}	5.50 [4.66 6.33]	< 0.001	< 0.001
condition			
-30%	7.77 [6.45 9.10]	< 0.001	< 0.001
-20%	5.10 [3.80 6.39]	< 0.001	< 0.001
-10%	1.46 [0.23 2.68]	0.020	1.000 *
+ 10%	-0.25 [− 1.44 0.94]	0.683	1.000
+ 20%	0.17 [− 1.04 1.37]	0.787	1.000
+ 30%	1.60 [0.41 2.79]	0.009	0.657 *
sex			
M-F	0.66 [− 2.10 3.41]	0.632	1.000

The difference was calculated between each condition and the reference condition (walking with preferred step lengths at 1.4 m-s-1). 95%CI indicates the 95% confidence interval. * indicates a change in significance after applying the Holm-Bonferroni correction.

3.2. Step-length (after accounting for sex and changes in speed)

The 1st peak in the vertical GRF increased in proportion to the deviation from the preferred step-length when both *under-striding* and *over-striding* (Table 2), however, only *over-striding* by 30 % was significantly

Table 2
Linear mixed model outputs for peak ground reaction forces.

	Ground reaction forces [BW]					
	Difference [95%CI]	Sig.	Corrected sig.	Difference [95%CI]	Sig.	Corrected sig.
	Posterior (-ve)			**Anterior (+ve)**		
speed						
1.6 m·s^{-1}	-0.03 [− 0.04 − 0.03]	< 0.001	< 0.001	0.03 [0.03 0.04]	< 0.001	< 0.001
1.8 m·s^{-1}	-0.05 [− 0.06 − 0.05]	< 0.001	< 0.001	0.06 [0.06 0.07]	< 0.001	< 0.001
step length						
-30%	0.11 [0.10 0.12]	< 0.001	< 0.001	-0.12 [− 0.12 − 0.11]	< 0.001	< 0.001
-20%	0.06 [0.06 0.07]	< 0.001	< 0.001	-0.07 [− 0.08 − 0.06]	< 0.001	< 0.001
-10%	0.03 [0.02 0.03]	< 0.001	< 0.001	-0.03 [− 0.04 − 0.02]	< 0.001	< 0.001
10%	-0.03 [− 0.03 − 0.02]	< 0.001	< 0.001	0.03 [0.02 0.04]	< 0.001	< 0.001
20%	-0.05 [− 0.06 − 0.04]	< 0.001	< 0.001	0.05 [0.05 0.06]	< 0.001	< 0.001
30%	-0.08 [− 0.09 − 0.08]	< 0.001	< 0.001	0.09 [0.08 0.09]	< 0.001	< 0.001
sex						
M-F	0.02 [0.00 0.05]	0.096	1.000	-0.01 [− 0.04 0.01]	0.333	1.000
	Lateral (-ve)			**Medial (+ve)**		
speed						
1.6 m·s^{-1}	-0.008 [− 0.01 − 0.005]	< 0.001	< 0.001	-0.0003 [− 0.01 0.01]	0.917	1.000
1.8 m·s^{-1}	-0.015 [− 0.02 − 0.01]	< 0.001	< 0.001	0.005 [− 0.002 0.01]	0.130	1.000
step length						
-30%	0.004 [0.0001 0.01]	0.043	1.000 *	0.014 [0.01 0.02]	< 0.001	0.037
-20%	0.005 [0.001 0.01]	0.018	1.000 *	0.011 [0.005 0.02]	< 0.001	0.047
-10%	-0.001 [− 0.01 0.003]	0.564	1.000	0.005 [0.0002 0.01]	0.043	1.000 *
10%	0.001 [− 0.003 0.005]	0.715	1.000	< −0.0001 [− 0.01 0.01]	0.993	1.000
20%	0.006 [0.002 0.01]	0.004	0.308 *	0.002 [− 0.004 0.01]	0.488	1.000
30%	0.007 [0.003 0.01]	< 0.001	0.020	0.001 [− 0.004 0.01]	0.651	1.000
sex						
M-F	-0.015 [− 0.03 0.003]	0.094	1.000	-0.018 [− 0.03 − 0.002]	0.033	1.000 *
	Vertical (First Peak)			**Vertical (Second Peak)**		
speed						
1.6 m·s^{-1}	0.09 [0.07 0.10]	< 0.001	< 0.001	0.04 [0.02 0.5]	< 0.001	< 0.001
1.8 m·s^{-1}	0.15 [0.13 0.16]	< 0.001	< 0.001	0.08 [0.06 0.09]	< 0.001	< 0.001
step length						
-30%	0.04 [0.02 0.06]	0.001	0.093 *	-0.17 [− 0.20 − 0.15]	< 0.001	< 0.001
-20%	0.01 [− 0.02 0.03]	0.591	1.000	-0.05 [− 0.08 − 0.03]	< 0.001	0.006
-10%	-0.001 [− 0.02 0.02]	0.898	1.000	-0.02 [− 0.05 0.00]	0.071	1.000
10%	0.02 [0.00 0.04]	0.081	1.000	0.01 [− 0.01 0.03]	0.309	1.000
20%	0.03 [0.01 0.05]	0.004	0.308 *	0.02 [− 0.01 0.04]	0.201	1.000
30%	0.05 [0.03 0.07]	< 0.001	0.002	0.03 [0.0 1 0.05]	0.012	0.852 *
sex						
M-F	-0.04 [− 0.13 0.05]	0.409	1.000	0.02 [− 0.05 0.09]	0.534	1.000

The difference was calculated between each condition and the reference condition (walking with preferred step lengths at 1.4 m-s-1). 95%CI indicates the 95% confidence interval. As the peak posterior and lateral ground reaction forces are negative, negative changes indicate increases in the peaks. * indicates a change in significance after applying the Holm-Bonferroni correction.

Table 4
Linear mixed model outputs for peak hip joint moments during stance.

	Hip moments [Nm/kg]					
	Difference [95%CI]	Sig.	Corrected sig.	Difference [95%CI]	Sig.	Corrected sig.
	Extensor (+ve)			**Flexor (-ve)**		
speed						
1.6 m·s^{-1}	0.25 [0.22 0.28]	< 0.001	< 0.001	-0.15 [− 0.19 − 0.11]	< 0.001	< 0.001
1.8 m·s^{-1}	0.49 [0.46 0.53]	< 0.001	< 0.001	-0.29 [− 0.33 − 0.26]	< 0.001	< 0.001
step length						
-30%	0.14 [0.09 0.19]	< 0.001	< 0.001	0.13 [0.07 0.19]	< 0.001	0.020
-20%	0.14 [0.09 0.19]	< 0.001	< 0.001	0.02 [− 0.02 0.07]	0.331	1.000
-10%	0.08 [0.03 0.12]	0.002	0.170 *	-0.04 [− 0.08 0.00]	0.068	1.000
10%	-0.04 [− 0.09 0.00]	0.059	1.000	0.04 [0.00 0.09]	0.041	1.000 *
20%	-0.05 [− 0.10 − 0.01]	0.03	1.000 *	0.10 [0.06 0.14]	< 0.001	< 0.001
30%	0.07 [0.03 0.12]	0.002	0.170 *	0.10 [0.05 0.15]	0.002	0.170 *
sex						
M-F	0.003 [− 0.11 0.12]	0.959	1.000	-0.05 [− 0.17 0.06]	0.355	1.000
	Abductor (+ve)			**Adductor (-ve)**		
speed						
1.6 m·s^{-1}	0.07 [0.06 0.09]	< 0.001	< 0.001	-0.04 [− 0.05 − 0.03]	< 0.001	< 0.001
1.8 m·s^{-1}	0.12 [0.11 0.14]	< 0.001	< 0.001	-0.10 [− 0.11 − 0.09]	< 0.001	< 0.001
step length						
-30%	0.03 [0.00 0.05]	0.036	1.000 *	0.04 [0.02 0.05]	< 0.001	< 0.001
-20%	0.02 [0.00 0.05]	0.071	1.000	0.03 [0.02 0.05]	< 0.001	0.011
-10%	0.01 [− 0.01 0.03]	0.345	1.000	0.01 [0.00 0.03]	0.058	1.000
10%	0.02 [0.00 0.04]	0.058	1.000	-0.02 [− 0.03 0.00]	0.011	0.792 *
20%	-0.01 [− 0.04 0.01]	0.254	1.000	-0.06 [− 0.08 − 0.05]	< 0.001	< 0.001
30%	-0.04 [− 0.07 − 0.02]	< 0.001	0.010	-0.10 [− 0.12 − 0.09]	< 0.001	< 0.001
sex						
M-F	-0.15 [− 0.27 − 0.03]	0.018	1.000 *	0.04 [0.00 0.09]	0.042	1.000 *
	External Rotator (+ve)			**Internal Rotator (-ve)**		
speed						
1.6 m·s^{-1}	0.05 [0.04 0.06]	< 0.001	< 0.001	-0.02 [− 0.03 − 0.01]	< 0.001	0.003
1.8 m·s^{-1}	0.08 [0.07 0.09]	< 0.001	< 0.001	-0.04 [− 0.06 − 0.03]	< 0.001	< 0.001
step length						
-30%	-0.07 [− 0.08 − 0.06]	< 0.001	< 0.001	0.03 [0.01 0.05]	0.002	0.170 *
-20%	-0.05 [− 0.07 − 0.04]	< 0.001	< 0.001	0.02 [0.01 0.04]	0.003	0.234 *
-10%	-0.03 [− 0.04 − 0.02]	< 0.001	< 0.001	0.01 [0.00 0.02]	0.075	1.000
10%	0.04 [0.02 0.05]	< 0.001	< 0.001	-0.01 [− 0.02 0.00]	0.132	1.000
20%	0.05 [0.04 0.07]	< 0.001	< 0.001	-0.02 [− 0.04 − 0.01]	0.001	0.093 *
30%	0.09 [0.08 0.10]	< 0.001	< 0.001	-0.04 [− 0.05 − 0.02]	< 0.001	< 0.001
sex						
M-F	-0.12 [− 0.23 − 0.01]	0.031	1.000 *	0.02 [− 0.05 0.09]	0.289	1.000

The difference was calculated between each condition and the reference condition (walking with preferred step lengths at 1.4 m·s-1). 95%CI indicates the 95% confidence interval. As the peak hip flexor, adductor, and internal rotator moments are negative, negative changes indicate increases in the peaks. * indicates a change in significance after applying the Holm-Bonferroni correction.

different to walking with *preferred* step-lengths. *Under-striding* had a greater effect on the 2nd peak in the vertical GRF, with both under-striding by 10 % and 20 % being significantly different to walking with *preferred* step-lengths, than *over-striding*; with the forces reducing as step-length reduced (Table 2). Peak anterior-posterior GRFs were significantly reduced as *under-striding* increased and significantly increased as *over-striding* increased (Table 2). Peak medial GRFs were significantly increased when *under-striding* by 20 % and 30 %, whereas peak lateral GRFs were significantly reduced when over-*striding* by 30 % only (Table 2). Vertical loading rates significantly increased in proportion to the level of *under-striding,* although non-significant at 10% (Table 3).

There were inconsistencies in the way peak hip and knee moments changed with changes to step length (Table 4 and Table 5, respectively). Peak hip extensor moments were significantly increased when *over-striding* by 20 % and 30 %, whereas peak hip flexor moments were significantly reduced when *under-striding* by 30 % and *over-striding* by 20 %. Peak hip abductor moments were significantly reduced when *over-striding* by 30 % only, whereas peak hip adductor moments were significantly reduced when *under-striding* and increased when *over-striding* by 20 % and 30 %. Peak hip external rotator moments were significantly reduced when *under-striding* and increased when *over-striding*, but peak hip internal rotator moments were only significantly increased when *over-striding* by 30%. Knee extensor moments were significantly increased when *over-striding* by 20 % and 30 %, whereas knee flexor moments were significantly reduced when *under- and over-striding* by 30%. Knee abductor moments were significantly reduced when *under-striding* by 20 % and 30 %.

There was more consistency in how ankle moments responded to changes in step length (Table 6). All ankle moments, except peak ankle inverter moments, were significantly reduced when *under-striding* and increased when *over-striding*. Differences were non-significant for peak ankle plantar flexor moments at + 10% and peak ankle evertor moments at − 10 % of *preferred* step-lengths. There were no significant differences in peak ankle invertor moments.

3.3. Sex (after accounting for changes in speed and step-length)

There were no significant differences between male and female civilians.

4. Discussion

This study aimed to identify the effects of speed, step-length, and sex on lower limb joint moments during walking. The data support our first hypothesis that increasing walking speed increased peak joint moments, however, the response to changes in step-length were less obvious. Our second hypothesis must be rejected as no sex differences were found for hip, knee, or ankle joint moments.

We found all peak joint moments normalised to body mass, except

Table 5
Linear mixed model outputs for peak knee joint moments during stance.

	Knee moments [Nm/kg]					
	Difference [95%CI]	Sig.	Corrected sig.	Difference [95%CI]	Sig.	Corrected sig.
	Extensor (+ve)			**Flexor (-ve)**		
speed						
1.6 m·s^{-1}	0.24 [0.19 0.29]	< 0.001	< 0.001	-0.05 [− 0.07 − 0.03]	< 0.001	< 0.001
1.8 m·s^{-1}	0.41 [0.36 0.46]	< 0.001	< 0.001	-0.10 [− 0.12 − 0.09]	< 0.001	< 0.001
step length						
-30%	-0.06 [− 0.14 0.02]	0.135	1.000	0.06 [0.03 0.09]	< 0.001	0.004
-20%	-0.07 [− 0.15 0.00]	0.04	1.000 *	0.03 [0.00 0.05]	0.048	1.000 *
-10%	-0.08 [− 0.14 − 0.02]	0.008	0.592 *	-0.01 [− 0.04 0.01]	0.255	1.000
10%	0.10 [0.04 0.16]	0.001	0.093 *	0.01 [− 0.01 0.03]	0.441	1.000
20%	0.20 [0.14 0.27]	< 0.001	< 0.001	0.04 [0.02 0.07]	0.001	0.093 *
30%	0.31 [0.24 0.38]	< 0.001	< 0.001	0.07 [0.05 0.10]	< 0.001	< 0.001
sex						
M-F	-0.07 [− 0.22 0.08]	0.369	1.000	0.003 [− 0.06 0.06]	0.923	1.000
	Abductor (+ve)			**Adductor (-ve)**		
speed						
1.6 m·s^{-1}	0.02 [0.01 0.04]	< 0.001	0.002	-0.02 [− 0.04 − 0.01]	< 0.001	0.044
1.8 m·s^{-1}	0.04 [0.03 0.05]	< 0.001	< 0.001	-0.05 [− 0.07 − 0.04]	< 0.001	< 0.001
step length						
-30%	-0.04 [− 0.05 − 0.02]	< 0.001	0.011	0.02 [0.00 0.04]	0.049	1.000 *
-20%	-0.03 [− 0.05 − 0.01]	< 0.001	0.036	0.01 [− 0.01 0.02]	0.472	1.000
-10%	-0.02 [− 0.03 0.00]	0.03	1.000 *	0.001 [− 0.02 0.02]	0.937	1.000
10%	0.03 [0.01 0.04]	0.001	0.093 *	0.01 [− 0.01 0.02]	0.239	1.000
20%	0.03 [0.01 0.04]	0.001	0.093 *	0.01 [− 0.01 0.03]	0.293	1.000
30%	0.03 [0.01 0.04]	0.001	0.093 *	-0.01 [− 0.03 0.02]	0.593	1.000
sex						
M-F	0.07 [− 0.06 0.21]	0.269	1.000	0.08 [0.04 0.13]	0.002	0.170 *

The difference was calculated between each condition and the reference condition (walking with preferred step lengths at 1.4 m·s-1). 95%CI indicates the 95% confidence interval. As the peak knee flexor and adductor moments are negative, negative changes indicate increases in the peaks. * indicates a change in significance after applying the Holm-Bonferroni correction.

Table 6
Linear mixed model outputs for peak ankle joint moments during stance.

	A nkle moments [Nm/kg]					
	Difference [95%CI]	Sig.	Corrected sig.	Difference [95%CI]	Sig.	Corrected sig.
	Plantar Flexor (+ve)			**Dorsiflexor (-ve)**		
speed						
1.6 m·s^{-1}	0.08 [0.06 0.10]	< 0.001	< 0.001	-0.04 [− 0.05 − 0.03]	< 0.001	< 0.001
1.8 m·s^{-1}	0.14 [0.12 0.16]	< 0.001	< 0.001	-0.07 [− 0.08 − 0.06]	< 0.001	< 0.001
step length						
-30%	-0.36 [− 0.39 − 0.33]	< 0.001	< 0.001	0.11 [0.10 0.13]	< 0.001	< 0.001
-20%	-0.20 [− 0.23 − 0.17]	< 0.001	< 0.001	0.08 [0.07 0.10]	< 0.001	< 0.001
-10%	-0.09 [− 0.12 − 0.06]	< 0.001	< 0.001	0.04 [0.03 0.05]	< 0.001	< 0.001
10%	0.05 [0.02 0.07]	0.002	0.170 *	-0.05 [− 0.06 − 0.03]	< 0.001	< 0.001
20%	0.08 [0.05 0.11]	< 0.001	< 0.001	-0.08 [− 0.09 − 0.07]	< 0.001	< 0.001
30%	0.09 [0.07 0.12]	< 0.001	< 0.001	-0.11 [− 0.12 − 0.09]	< 0.001	< 0.001
sex						
M-F	0.07 [− 0.03 0.17]	0.146	1.000	-0.08 [− 0.12 − 0.03]	0.002	0.170 *
	Evertor (+ve)			**Invertor (-ve)**		
speed						
1.6 m·s^{-1}	0.03 [0.02 0.04]	< 0.001	< 0.001	-0.004 [− 0.01 0.002]	0.157	1.000
1.8 m·s^{-1}	0.05 [0.04 0.06]	< 0.001	< 0.001	0.002 [− 0.003 0.01]	0.411	1.000
step length						
-30%	-0.07 [− 0.08 − 0.06]	< 0.001	< 0.001	-0.01 [− 0.01 0.004]	0.255	1.000
-20%	-0.05 [− 0.06 − 0.04]	< 0.001	< 0.001	-0.002 [− 0.01 0.01]	0.686	1.000
-10%	-0.02 [− 0.03 − 0.01]	0.001	0.093 *	-0.01 [− 0.01 0.003]	0.227	1.000
10%	0.02 [0.01 0.04]	< 0.001	0.003	-0.01 [− 0.01 0.002]	0.12	1.000
20%	0.04 [0.03 0.05]	< 0.001	< 0.001	-0.01 [− 0.02 − 0.002]	0.016	1.000 *
30%	0.07 [0.06 0.08]	< 0.001	< 0.001	-0.01 [− 0.02 − 0.003]	0.006	0.450 *
sex						
M-F	0.03 [− 0.05 0.11]	0.438	1.000	-0.001 [− 0.06 0.06]	0.962	1.000

The difference was calculated between each condition and the reference condition (walking with preferred step lengths at 1.4 m·s-1). 95%CI indicates the 95% confidence interval. As the peak ankle dorsiflexor and invertor moments are negative, negative changes indicate increases in the peaks. * indicates a change in significance after applying the Holm-Bonferroni correction.

the peak ankle invertor moments, were significantly larger when walking at 1.6 m·s^{-1} and 1.8 m·s^{-1}, compared to walking at 1.4 m·s^{-1}, similar to existing literature. Increases in the average hip, knee, and ankle sagittal plane moments, normalised to body weight and leg length, have been reported as walking speed increased from 1.1 m to 1.5 m·s^{-1} [8]. Non-normalised peak sagittal, frontal, and transverse lower limb joint moments are also significantly correlated with deviations from preferred walking speeds [12].

Any increases in peak joint moments were seen in response to *over-striding*, except for hip extensor moments which were larger when *under-striding*. Existing literature has similarly shown knee extensor moments to be larger when *over-striding* [7–9]. Allet, et al. [7] reported similar results to this study, with peak hip flexor moments being larger when *under-striding*, though only significant in the most extreme condition, and peak ankle plantar flexor moments being significantly smaller when *under-striding*. In contrast, other studies reported increases in the peak [9] and average [8] hip extensor moments, however, differences in methodologies might explain differences in these results. Allet, et al. [7] reported non-normalised joint moments, which wouldn't account for differences in body mass, Buddhadev, et al. [8] used the average net extensor moments for each joint, defined "*by dividing each joint's extensor angular impulse by the time (in seconds) of its extensor moment*", whereas Lim, et al. [9] investigated changes to "nominal" step-lengths, which may have already required participants to deviate from their *preferred* walking gait.

No sex differences were found for the peak GRFs, loading rates, or joint moments. These findings are different to those reported by Chehab, et al. [12] who found that female participants had larger hip internal rotator moments and smaller knee adductor moments than male participants, however, it's worth noting they compared non-normalised data. Other studies either did not compare between sexes [7,8] or did not mention the sex of participants [9].

Repetitive loading related to activity/profession have been associated with stress fractures and soft-tissue injuries [22–25], as well as the onset and progression of osteoarthritis [26]. Interventions to military training, such as removing the requirement to march "in-step" [2], reducing the standard step-length during marches [3], and grouping recruits based on physical ability and aerobic fitness [27], has resulted in decreases in the incidence of stress fractures in military recruits. As female recruits/personnel are generally shorter than their male counterparts, they are more likely to have to *over-stride* where groups are required to move together or, within the military, march "in-step". We, therefore, believe that this contributes to the greater risk of stress fracture seen in female recruits [5,28]. To mitigate this increased risk, our findings suggest that individuals should be encouraged to use *preferred* gait patterns where possible. Alongside reports of other sex-specific responses to physical training [29], training should be tailored to meet the requirements of each sex.

There are some limitations to consider in the interpretation of our results. First, there is a possibility that missing data and a low sample size may have influenced the statistical power of the analysis. However, linear mixed models are robust to missing data and although they may be underpowered due to the number of participants, the results presented are still noteworthy and provide an intriguing perspective. The authors believe these findings are still worth considering and may serve as a catalyst for further investigation in this field. Second, markers were attached to the shoe and not to the foot, so all ankle variables are of the shoe relative to the shank and so may not reflect the true movement of the foot. Third, the participants included were active civilians and although they are likely representative of military recruits commencing basic training, the findings of this study may not extrapolate to trained military personnel.

In conclusion, this study has shown that walking faster and *over-striding* predominantly increases peak joint moments; suggesting *over-striding* is more likely to negatively affect injury risk than *under-striding*. Increased joint moments are likely to have a cumulative effect on a muscles capability to withstand the increased external forces associated with walking faster and with longer step-lengths and thus could lead to an increased risk of developing an injury.

Declaration of Competing Interest

There are no conflicts of interest for any of the authors.

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