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Primarily hip-borne load carriage does not alter biomechanical risk factors for overuse injuries in soldiers

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ABSTRACT

Objectives: To determine the effects of different body armour types, carried loads, and walking speeds on trunk and lower-limb joint biomechanics.
Design: Within-subjects repeated measures to determine the effects of different body armour types, carried loads, and walking speeds on trunk and lower-limb joint biomechanics.
Methods: Twenty soldiers (29.5 ± 7.1yrs) completed a treadmill walking protocol in an unloaded (baseline) condition and wearing a control, Tiered Body Armour System (TBAS) and five different armour types (cARM1-2, pARM1) with two load configurations (15 and 30 kg) for a total of eight armour × load ensembles. In each ensemble, participants walked for 10 min at 1.53 m s^{-1} and 1.81 m s^{-1} speeds. Whole-body marker kinematics and ground reaction forces were used, along with a scaled anatomic model, to determine peak lower-limb joint angles, net joint moments, and negative knee work. Peak parameters were compared between armour types, walking speeds, and carried loads using repeated measures ANOVAs.
Results: Peak plantarflexion and hip abduction moments were reduced when wearing cARM1 ($p = 0.040$, $p = 0.045$) and cARM2 ($p = 0.045$, $p = 0.003$) compared to TBAS, while carrying 30 kg and/or walking fast. This suggests positive benefits of load distribution at higher task demands. Joint moments increased when participants carried greater load and/or walked faster, and the combined effects of carried load and walking speed were mostly additive.
Conclusions: Primarily hip-borne load carriage does not negatively alter joint kinetics, and some positive adaptations occurred during tasks with higher demands. These results can inform equipment design and physical training programs for load carriage.

1. Introduction

The physical demands of military service place soldiers at risk of musculoskeletal injuries (MSI).[1] Over the past decade in the military, lower extremity MSI rates have more than doubled, with U.S soldiers sustaining one or more MSI during a typical service career.[2] Detrimental to military capability, MSI impose substantial medical and rehabilitation costs, as well as indirect costs due to reduced work capacity. Numerous studies have identified training mileage (e.g., running and marching), structural bone abnormalities, and exposure to heavy carried loads [3–5] as risk factors for MSI. Indeed, soldiers carrying equipment to enhance their capability in the field have a greater risk of sustaining MSI.[3] However, the MSI mechanisms remain unclear. Knowledge of the biomechanics behind risk factors for lower-limb MSI (e.g., load carriage) could inform injury prevention programs or equipment design modifications. However, this requires understanding the mechanisms of MSI, and relies upon characterising the biomechanical surrogates (e.g., joint kinematics, moments, and powers) of injury during specific tasks.[6] These measures may provide better prediction of injury risk than, for instance, previous physical activity history, because they more closely relate to the joint and tissue mechanical environment.[6]

Previous studies have examined the biomechanics of walking and running with carried load. Trunk and knee flexion

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increase to mediate compromised stability caused by carried load.[7] Consequently, knee extension moments and negative knee work during weight acceptance increase[8–10] to prevent the body collapsing under the additional mass, and to regulate torso angular momentum.[11,12] These biomechanical changes are exacerbated when carried load, gait speed, or a combination of both, increase.[8] Consequently, hip and knee reaction forces likely increase as peak knee extension moment is positively correlated with patellofemoral ($R^2 = 0.68$) and tibiofemoral ($R^2 = 0.75$) reaction forces.[13,14] Moreover, joint reaction forces may influence compressional and tensional strains to lower-limb bones (e.g., tibia),[15] thereby increasing stress fracture risk. Reducing the peak knee extension moment and negative knee work during load carriage may therefore be important targets for reducing MSI risk.

Reducing MSI risk in soldiers performing load carriage tasks may be achieved by redistributing carried load. Compared to standard load placement (i.e., mid-back), load placed closer to the body centre of mass (COM) and higher on the carrier has been shown to reduce metabolic cost.[16,17] However, it also increases trapezius and erector spinae muscle activity,[18] and is sub-optimal for soldiers who require quick access to mission-vital equipment. Backpack designs incorporating a hip belt that brings the load closer to the body COM have been shown to reduce forces experienced on the upper back[19] and peak trunk rotation.[20] Moreover, a front-back load configuration produces walking patterns (e.g., trunk and hip flexion) similar to unloaded walking,[21] which could be interpreted as beneficial. There is little research focused on the lower-limb joint loading produced while wearing the above mentioned equipment configurations at different load magnitudes and walking speeds. Furthermore, there has been no investigation as to whether incorporating a hip belt with body armour load reduces specific biomechanical risk factors for MSI.

This study aimed to determine the effects of changing where torso load is borne, the magnitude of load, and walking speed on joint kinematics, lower-limb joint moments, and energy absorption at the knee during walking. We hypothesised that bearing load closer to the hips will result in a more upright posture compared to wearing a standard design, and decrease the magnitude of knee extension moment and negative knee work during the early stance phase of walking. Additionally, we hypothesised that in response to increasing the carried load and walking faster, trunk flexion, peak knee flexion, negative knee work, and joint moments will increase.

2. Methods

Twenty Australian Army Reserve male soldiers (age: 29.5 ± 7.1 years, height: 1.77 ± 0.08 m, mass 82.8 ± 12.1 kg) participated in this study. All were in good health, reporting no injuries, and no physician recommendation to avoid strenuous exercise. Participants gave written informed consent approved by the Australian Defence Human Research Ethics Committee (Protocol 756-14). Defence Science and Technology Group contributed funding to support this research, assisted with data collection, and provided a Defence perspective on data interpretation.

Six different body armour systems were tested. These were the current Australian Army standard-issue Tiered Body Armour System (TBAS), with predominantly shoulder-borne loading, and five load sharing systems designed to transfer some load to the hips. Two commercially available load sharing systems (cARM1 and cARM2) and three prototypes (pARM1-3) were tested. While data from pARM2 and pARM3 armour systems were collected, these systems were not included in subsequent analyses because these designs were almost identical to pARM1, but were very new prototypes whereas pARM1 was a mature prototype. A description and illustration of armour types and load configurations is available in Supplemental Content 1.

Participants attended four separate laboratory-based testing sessions to complete a standardised treadmill walking protocol in a baseline condition and 12 armour conditions (six armour types and two carried loads). An armour condition refers to a combination of armour type and carried load (e.g., TBAS with 15 kg or cARM1 with 30 kg). In the first session, participants completed the walking protocol in the baseline condition (i.e., athletic shirt, shorts, and military boots) first followed by three armour conditions. In each of the three remaining sessions, participants completed the walking protocol in three armour conditions without the baseline condition being completed again. The order in which participants completed the walking protocol in each armour condition was randomised and counterbalanced to prevent order effects. Each testing session was separated from the previous by at least 3 days to minimise fatigue effects.

During treadmill walking, full-body three-dimensional motion and ground reaction forces (GRF) were acquired. Spherical 14 mm diameter retro-reflective markers and marker clusters were placed on the head, torso, arms, and legs of the participant consistent with a marker set specifically developed and validated for military load carriage applications.[22] Marker and GRF data were measured using an 11-camera (100 Hz) three-dimensional motion capture system (Vicon, Oxford, UK) and a fore-aft split-belt force-measuring (1000 Hz) treadmill (AMTI Compact Tandem, Watertown, MA, USA), respectively. Prior to treadmill walking, a static calibration trial was recorded of the participant standing for subsequent musculoskeletal model scaling.

In each armour/load ensemble, participants completed a treadmill walking protocol consisting of a 5-min warm-up at 1.25 m s^{-1}, 10 min "moderate" walking at 1.53 m s^{-1}, 10 min "fast" walking at 1.81 m s^{-1}, and a 5-min cool-down at 1.25 m s^{-1}. These speeds represent administrative (i.e., moderate) and approach (i.e., fast) marching speeds in the Australian Army. Marker and GRF data were acquired for 30 s at the beginning of the final minute of each walking speed. Participants were given a minimum of 25 min rest after each treadmill protocol and then repeated the protocol for the remaining armour conditions assigned for that session.

Using Vicon Nexus version 2.3, marker positions were reconstructed and cleaned. Marker and GRF data were exported from Vicon Nexus via MotoNMS,[23] a Matlab toolbox (R2014b, The Mathworks) for preparing motion capture data for modelling in OpenSim.[24] From the 30 s walking trials, heel strike gait events were detected by calculating the horizontal distance between the calcaneus marker and sacrum marker, and finding the times corresponding to the maximum distances.[25] Consecutive heel strikes were used to demarcate individual gait cycles. To ensure erroneous peaks were not detected, peaks were only obtained if they occurred when the heel marker was lower than 30% of the range of marker height during the trial (i.e., occurred when the foot was close to the ground). Marker and GRFs data were then filtered using a 4th order zero-lag low-pass Butterworth design[26] (10 Hz cut-off), and transformed from the laboratory to OpenSim coordinate system.

A generic full-body anatomic model[27] was used within OpenSim version 3.3. For each participant, the model bodies were linearly scaled, preserving relative mass distribution, but adjusting the overall model mass and dimensions to match the participant. This scaled anatomic model was used to determine whole-body gait kinematics using an OpenSim inverse kinematics (IK) algorithm. This algorithm solved knee flexion, with knee rotations and translations prescribed from a spline within the model. However, to enable computation of all knee moments, knee adduction/abduction and internal/external rotation degrees of freedom were added. The OpenSim inverse dynamics tool was then used to calculate net joint moments, from which total negative knee joint work per stride was calculated by integrating the area under the instantaneous knee joint power curve from heel strike until the first peak of knee flexion angle in early stance phase.[12]

Each 30 s epoch of a participant's gait data contained at least 25 gait cycles, from which at least 10 gait cycles (mean 21 ± 5) were used to create ensemble averages of the joint angles and moments. The ensemble average joint angles, moments, and work were averaged across all participants for each walking speed and armour × load configuration to generate mean curves. From the mean curves, peak joint angles and moments extracted from the entire gait cycle (except for peak knee flexion which extracted from early stance), and negative work were used in statistical analysis.

Statistical analysis was performed with R version 3.3.1 in R-Studio (RStudio, Inc, Boston, MA) v0.99.903. The Shapiro–Wilk test was used to confirm the data were normally distributed. Three-way repeated measures ANOVA tested for significant main effects of armour type (TBAS, cARM1-2, and pARM1), carried load (15 kg and 30 kg), and walking speed (moderate and fast), and their interactions for each dependent variable. Post-hoc analyses (i.e., paired t-tests) were performed if significant main effects or interactions were revealed to identify specific paired differences. All p-values for the post-hoc tests were adjusted using the Benjamini–Hochberg method.[28] Effect sizes (i.e., partial η^2) were calculated for all the significant main and interaction effects, with small, medium, and large effects defined as η^2 between 0.01 and 0.06, 0.06 and 0.14, and greater than 0.14 respectively.[29] Significance was set at $p < 0.05$ for all post-hoc comparisons.

3. Results

An interaction was shown between armour type, load, and walking speed for peak ankle plantarflexion moment ($p = 0.015$, $\eta^2 p = 0.01$). Post-hoc analysis revealed peak ankle plantarflexion moment was higher for TBAS during fast walking with 30 kg (2.16 ± 0.21 Nm kg^{-1}) compared to values obtained with cARM1 (2.04 ± 0.27 Nm kg^{-1}, $p = 0.040$) and cARM2 (2.00 ± 0.26 Nm kg^{-1}, $p = 0.045$). No other three-way interactions existed for the remaining variables. There was a significant interaction between armour type and load magnitude for peak hip abduction moment ($p = 0.035$, $\eta^2 p = 0.03$). When carrying 30 kg load, peak hip abduction moment was higher for TBAS (1.14 ± 0.27 Nm kg^{-1}) compared to cARM1 (1.00 ± 0.27 Nm kg^{-1}, $p = 0.045$), cARM2 (0.94 ± 0.26 Nm kg^{-1}, $p = 0.003$), and pARM1 (0.98 ± 0.26 Nm kg^{-1}, $p = 0.003$). Comparisons of peak joint angles revealed significant interactions between armour type and carried load for mean trunk flexion ($p = 0.021$), trunk bending ($p < 0.001$), and trunk rotation ROM ($p < 0.001$) (Fig. 1). When 15 kg was carried, mean trunk flexion and trunk bending were lower for all load sharing systems compared to TBAS ($p < 0.001$). Conversely, when 30 kg was carried, there were no differences in trunk kinematics between armour types. Significant main effects of armour type were shown for peak hip flexion and trunk flexion (Supplemental Content 2), while joint moments were not different.

Repeated measures ANOVAs revealed significant interactions between carried load and walking speed for peak hip extension moment (Table 1). Specifically, peak hip extension moment increased 46% when walking fast and carrying 30 kg of load ($p < 0.001$), and 37.7% when walking fast and carrying 15 kg of load ($p < 0.001$) compared to values obtained when walking at the moderate speed. Significant main effects of load magnitude were shown for peak knee extension moment, negative knee work, peak knee flexion, peak hip flexion, and trunk flexion ROM. Increasing load magnitude from 15 kg to 30 kg resulted in greater peak knee extension moments ($p < 0.001$), 0.13 J kg^{-1} greater negative work done ($p = 0.003$), and increased peak knee flexion, hip flexion, and trunk flexion (Table 1).

4. Discussion

This study aimed to determine the effects of different body armour designs, load magnitudes, and walking speeds on lower-limb joint biomechanics during load carriage. Motion capture data and GRFs were collected while soldiers walked on a treadmill at moderate and fast speeds wearing different armour/load configurations. Joint angles, moments, and negative knee work were computed, and the results revealed that redistribution of upper-body loads through hip-belt integrated armour designs improved walking posture and reduced both ankle plantarflexion and hip abduction moments during fast walking with 30 kg of load. Importantly, lower-limb joint loading was not increased with the load sharing armour designs compared with the standard issue TBAS.

Supporting the first hypothesis, hip-belt integrated armour designs resulted in a more upright walking posture and reduced trunk ROM compared to TBAS. However, these postural adjustments at the trunk were not observed when carrying 30 kg, suggesting heavy loads negate the effects during less demanding tasks. Nevertheless, the changes in joint angles between armour designs were similar to the changes observed when carried load was distributed close to, and low on, the body.[8] The hip belt forms an external physical link between trunk and pelvis, thus providing a mechanism through which mass worn on the trunk are, in part, borne by the pelvis. However, due to the external load being added to the same position on each armour type, the COM position was likely similar, which may explain why no differences in hip and knee joint moments and negative knee work were observed between armour designs. Based on these results, military organisations could confidently prescribe hip-belt integrated armour designs without risking an increase in lower-limb joint demands (i.e., the magnitude of joint moments) compared to current-issue body armour. Additionally, under high mass and inertia conditions (i.e., when walking fast and carrying 30 kg) our results showed 5.9% and 7.5% reductions in ankle plantarflexion moment over the late stance phase for cARM1 and cARM2 compared to TBAS, respectively. With these systems, there was less trunk flexion and more hip flexion, which may act to place the COM of the head, arms, trunk and armour closer to the ankle joint centre (Fig. 2), thereby decreasing the peak ankle plantarflexion moment generated by gastrocnemius and soleus muscles and tendons. Possibly, lower co-activation of gastrocnemii driven by lower plantar flexion moments may lower knee joint contact forces, since the gastrocnemii muscles cross the knee and are major contributors to knee joint contact forces.

The observed kinematic and kinetic differences between hip-belt integrated armour design and TBAS may have implications for MSI risk. First, reduced peak plantarflexion moment suggests reduced forces in gastrocnemius and soleus muscles, and reduced peak stress within the Achilles tendon.[30] However, further studies using neuromusculoskeletal models informed from experimental muscle excitations are required to elucidate potential benefits. Second, in providing stability through reduced hip frontal plane kinetic demands (i.e., hip abduction moment), hip-belt integrated armour designs possibly reduce the need for peak hip internal rotation compared to TBAS. This may be relevant to tibial torsion injuries as improper alignment of proximal joints may result in inappropriate tibial loading. Previous epidemiologic studies have shown 7–8° reductions in hip internal rotation during physical examinations is associated with increased risk of tibial stress fractures.[31]

Consistent with our second hypothesis, joint moment magnitudes increased in response to increases in carried load and/or

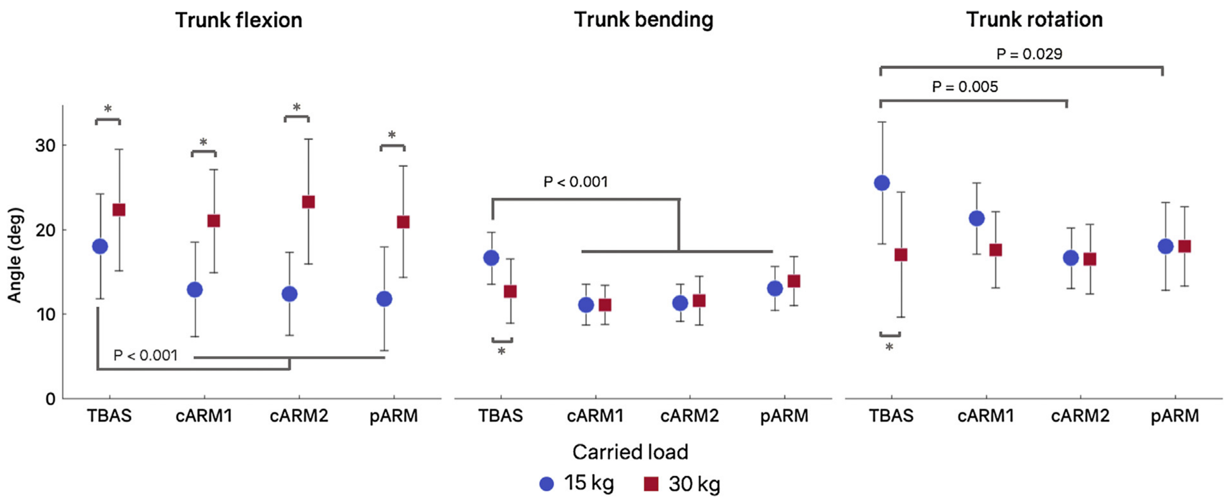

Fig. 1. The interaction between carried load and armour type for mean trunk flexion, and trunk bending, and rotation ROM.
*Indicates values are significantly different between loads within the same armour type. All other significant differences are between armour types with 15 kg load.

Table 1
Mean ± SD magnitudes for the joint angle and kinetic variables for different walking speeds and loads carried aggregated across armour types. All kinematic variables are reported in degrees, while joint moments are reported in $Nm\,kg^{-1}$ and work in $J\,kg^{-1}$.

	Walking speed		Effect size	Load		Effect size	Interaction	Effect size
	Moderate	Fast		15 kg	30 kg		p-Value	
Kinematics								
Ankle plantarflexion[a,b]	18.4 ± 4.8	19.0 ± 5.3	0.004	19.3 ± 4.9	18.1 ± 5.1	0.011	0.151	0.001
Ankle dorsiflexion[a,b]	12.5 ± 4.1	10.9 ± 3.8	0.043	11.0 ± 4.0	12.4 ± 4.0	0.036	0.261	0.001
Knee flexion[a,b,c]	24.6 ± 4.5	27.8 ± 4.8	0.140	24.1 ± 4.1	28.4 ± 4.7	0.231	<0.001	0.010
Hip flexion[a,b,c]	35.5 ± 5.9	40.1 ± 6.2	0.149	36.1 ± 5.9	39.5 ± 6.6	0.088	0.003	0.007
Hip extension[c]	16.1 ± 5.8	16.2 ± 5.7	0.000	15.8 ± 5.7	16.5 ± 6.2	0.005	0.035	0.010
Hip internal rotation[b]	0.7 ± 5.2	2.3 ± 4.9	0.029	1.3 ± 4.7	1.7 ± 5.1	0.001	0.282	0.000
Trunk flexion[a,b]	20.3 ± 8.6	21.2 ± 8.8	0.004	15.4 ± 6.6	26.1 ± 7.1	0.396	0.333	0.000
Trunk bending ROM[c]	12.6 ± 3.1	13.0 ± 3.3	0.005	12.9 ± 3.2	12.7 ± 3.2	0.000	0.024	0.004
Trunk rotation ROM[a,b]	16.0 ± 4.6	20.6 ± 5.4	0.056	19.3 ± 5.6	17.2 ± 5.2	0.019	0.822	0.000
Kinetics								
Ankle plantarflexion[a,b]	1.88 ± 0.25	1.97 ± 0.24	0.043	1.80 ± 0.20	2.04 ± 0.24	0.251	0.134	0.008
Ankle dorsiflexion[b]	−0.24 ± 0.19	−0.25 ± 0.19	0.001	−0.27 ± 0.20	−0.21 ± 0.18	0.022	0.068	0.006
Knee flexion[a]	−0.59 ± 0.22	−0.80 ± 0.23	0.194	−0.67 ± 0.24	−0.72 ± 0.25	0.016	0.262	0.002
Knee extension[b]	0.48 ± 0.25	0.49 ± 0.24	0.000	0.44 ± 0.23	0.53 ± 0.25	0.029	0.091	0.004
Knee abduction[a,b]	0.70 ± 0.23	0.84 ± 0.24	0.086	0.73 ± 0.23	0.81 ± 0.25	0.027	0.831	0.000
Hip flexion[a,b]	−1.14 ± 0.33	−1.18 ± 0.34	0.004	−1.10 ± 0.30	−1.21 ± 0.35	0.030	0.602	0.000
Hip extension[a,b,c]	1.86 ± 0.59	2.62 ± 0.62	0.324	2.01 ± 0.62	2.47 ± 0.72	0.149	0.001	0.020
Hip abduction[b]	0.93 ± 0.26	0.96 ± 0.27	0.002	0.89 ± 0.25	1.00 ± 0.27	0.047	0.938	0.000
Hip adduction[a]	−0.53 ± 0.17	−0.67 ± 0.20	0.115	−0.58 ± 0.20	−0.62 ± 0.19	0.008	0.172	0.001
Negative knee work[a,b]	−0.52 ± 0.36	−0.63 ± 0.39	0.022	−0.54 ± 0.36	−0.67 ± 0.41	0.027	0.392	0.001

ROM = range of motion. $p < 0.05$ for main effects.
[a] Indicates significant differences in absolute magnitude of the variable between walking speeds.
[b] Indicates significant differences in absolute magnitude of the variable between loads.
[c] Indicates significant interaction from repeated-measures ANOVA between walking speed and carried load.

walking speed. Unique to this study, hip joint moments experienced the largest magnitude change as task demands increased. Previous load carriage studies showed knee, not hip, extension moments were more sensitive to carried load,[8,32,33] increasing ~200% when loading increased from no external load to 32 kg and 55 kg.[8,10] This substantial increase in knee extensor moment suggests the quadriceps absorb more energy with increasing load, which may precipitate fatigue and, ultimately, increase knee injury risk.[8,32] Our results revealed increased negative knee work with increasing load, however, the magnitudes of peak hip extensor moments were more than two times those of knee extensor moments. Therefore, while quadriceps muscles probably absorb more energy with increasing carried load, the increased biomechanical demands of heavy load carriage at faster walking speeds are met predominantly by the hip joint. A possible explanation for the disagreement between this study and previous research is that our participants walked up to $1.81\,m\,s^{-1}$ and carried up to 30 kg of load, whereas others had either civilian participants carrying less external load (e.g., 10–20 kg) or military participants carrying increased external load (e.g., 30–55 kg) but walking slower (e.g., $1.25–1.67\,m\,s^{-1}$) compared the current study. Based on these results, load-carriage equipment design should focus on assisting or not impeding hip extensors, and physical conditioning programs for load-carriage should include specific training of hip musculature.

The present study had limitations that should be acknowledged. First, net moments computed from inverse dynamics do not account for co-contraction of antagonist muscles. Future studies of load carriage should incorporate experimental measures of muscle activity to inform estimates of muscle joint loading and work. Second, participants walked in ideal laboratory conditions on a treadmill to facilitate data collection and control

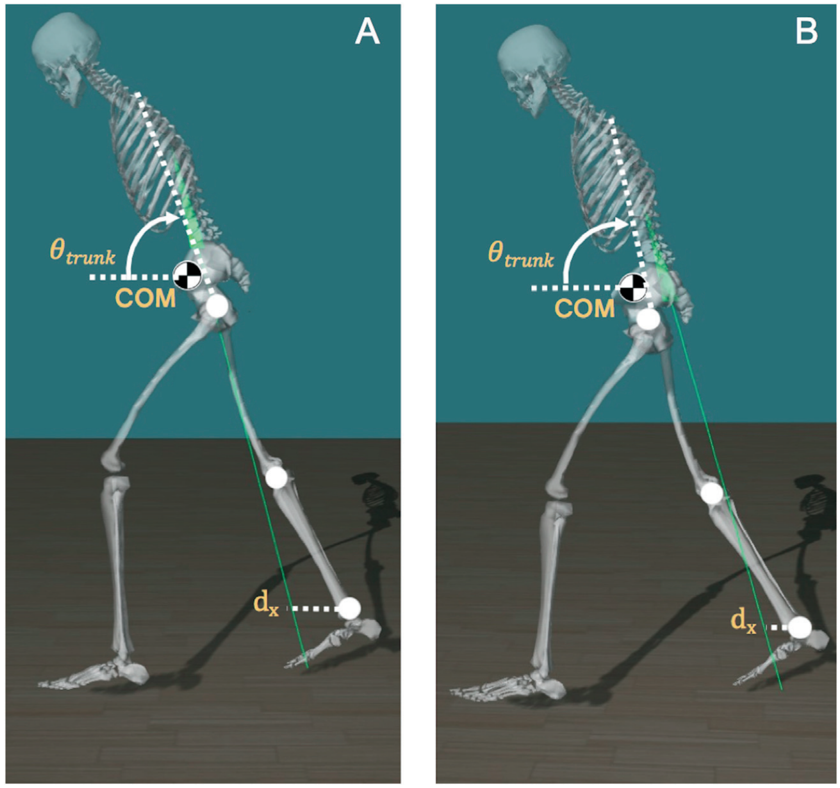

Fig. 2. Comparison of how trunk flexion angle and centre of mass (COM) position are altered when walking with (A) TBAS armour type and (B) cARM1 armour type during fast walking with 30 kg of carried load. Conceivably, the distance (d_x) from the ground reaction force line of action to the ankle joint centre is reduced with cARM1, which may have caused the 7.5% reduction in ankle plantarflexion moment compared to TBAS.

confounding variables. Although a recent study has demonstrated overground and treadmill running are similar,[34] laboratory conditions do not replicate external environmental factors (e.g., heat, humidity, undulating surfaces) which would make tasks more representative of in-field conditions. Third, we prescribed short bouts of walking and adequate rest to minimise fatigue effects, but we did not measure physiologic fatigue explicitly. Previous studies have demonstrated >40 min of walking are required to induce fatigue in gait,[33] which suggests fatigue was not a factor influencing results in this study. The effects of carried load on gait biomechanics when fatigued may be different to what has been reported in this study and would be interesting to examine.

5. Conclusions

This study indicated that soldiers carrying heavy loads, even for short periods, experience kinematic and kinetic changes that may place them at risk for developing overuse MSI. This study was the first to show that body armour load borne closer to the hips does not adversely affect hip, knee, and ankle moments. Results from this study can be used to inform physical training programs, i.e., targeting hip extension/flexion strength during training to enable soldiers' to tolerate carrying heavy loads. Additionally, assistive devices for load carriage should target the hip joint to provide the most benefits.

Practical implications

- Soldiers use different movement patterns to walk in body armour that contains a hip belt compared to walking in body armour without a hip belt. As such, physical training programs for load carriage should be designed specifically for different equipment and carried load.
- Because peak hip extension moments were more than double peak knee extension moments, physical training programs for load carriage should focus on strengthening muscles spanning the hip.
- Assistive devices for load carriage should provide assistance at the hip joint during the weight acceptance phase of gait to ease load carriage burden.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.jsams.2018.06.013.

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