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Ankle and knee moment and power adaptations are elicited through load carriage conditioning in males

Jodie A. Wills^a, David J. Saxby^b, Gavin K. Lenton^b, Timothy L.A. Doyle^{a,*}

^a Department of Health Professions, Faculty of Medicine and Health Sciences, Macquarie University, Australia

^b Griffith Centre for Biomedical and Rehabilitation Engineering, Menzies Health Institute Queensland, School of Allied Health Sciences, Griffith University, Australia

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ABSTRACT

Soldiers routinely conduct load carriage and physical training to meet occupational requirements. These tasks are physically arduous and are believed to be the primary cause of musculoskeletal injury. Physical training can help mitigate injury risk when specifically designed to address injury mechanisms and meet task demands. This study aimed to assess lower-limb biomechanics and neuromuscular adaptations during load carriage walking in response to a 10-week evidence-based physical training program. Thirteen male civilian participants donned 23 kg and completed 5 km of load carriage treadmill walking, at 5.5 km h⁻¹ before and after a 10-week physical training program. Three-dimensional motion capture and force plate data were acquired in over-ground walking trials before and after treadmill walking. These data were inputs to a musculoskeletal model which estimated lower-limb joint kinematics and kinetics (i.e., moments and powers) using inverse kinematics and dynamics, respectively. A two-way analysis of variance revealed significant main effect of training for kinematic and kinetics parameters at the knee and ankle joints ($p < 0.05$). Post-Hoc comparisons demonstrated a significant decrease (4.2%) in total negative knee power between pre- and post-March 5 km measures after training ($p < 0.05$). Positive power contribution shifted distally after training, increasing at the post-march measure from 39.9% to 43.6% at the ankle joint ($p < 0.05$). These findings demonstrate that a periodised training program may reduce injury risk through favourable ankle and knee joint adaptations.

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1. Introduction

Military personnel routinely conduct occupational load carriage tasks and associated physical training. Exposure to load carriage is known to increase physical demands and is believed to increase injury risk of musculoskeletal injury (Brushøj et al., 2008). Despite this, exposure to load carriage cannot be reduced as it remains essential to supporting and sustaining military operations. As an alternative, physical training can help mitigate musculoskeletal injury risk if designed and implemented to target specific injury mechanisms using an evidence-based approach (Finch et al., 2016). Therefore, developing and implementing a physical training program that decreases the gap between specific task demands and soldier's physical capacity may be a simple and effective method to optimise load carriage performance and minimise injury risk.

Emerging evidence suggests that interventions focussing on specific injuries and associated mechanisms using an evidence-based approach are effective in reducing injury incidence during initial physical training in the military (Coppack et al., 2011; Friedl et al., 2015; Sharma et al., 2014). Many studies have examined the effects of physical training on load carriage performance, and generally it has been established that occupational load carriage demands high muscular strength and aerobic capacity (Friedl et al., 2015). Progressive resistance training and repeated task exposure (e.g., simulated loaded walking tasks) are known to result in improved occupational task performance (Kraemer et al., 2001; Szivak and Kraemer, 2015). Combining these modalities into a periodised training program could assist in the reduction of cumulative demands and detrimental effects that are often experienced by soldiers during physical training and load carriage tasks (Kraemer et al., 2001; Szivak and Kraemer, 2015; Williams et al., 2002).

To successfully develop a physical training program for load carriage, the primary joints and muscle groups responsible for movement during task specific load carriage need to be identified. Mechanical work describes the amount of, or change in energy

* Corresponding author.

E-mail addresses: jodie.wills@hdr.mq.edu.au (J.A. Wills), d.saxby@griffith.edu.au (D.J. Saxby), g.lenton@griffith.edu.au (G.K. Lenton), tim.doyle@mq.edu.au (T.L.A. Doyle).

transferred by force, internal (i.e., muscle) or external (i.e. gravity) over time (Robertson et al., 2013). Joint moments and powers are representative of net muscular contributions effectively controlling and counteracting external loads being carried (Seay et al., 2014), and reportedly increase in response to increased loads carried during walking (Huang and Kuo, 2014; Wang et al., 2013). Seay et al. (2014) concluded that carrying 15 kg of external load substantially increased knee joint moments during walking, suggesting the knee is the primary joint contributing towards adaptive responses. Similarly, Wang et al. (2013) reported increases in knee and hip joint moments, suggesting an increased demand on joints up the kinetic chain during such tasks. These findings are further supported through observations by Lenton et al. (2019) who identified approximately 65% of positive power is generated by the hip during load carriage walking, followed by the ankle (25%), and knee (10%). Shifting power production to proximal joints would increase the work performed by the hip musculature and increase active work by hip-spanning muscles due to the muscle-tendon architecture (Neptune et al., 2009). The shift towards a hip-dominated strategy during load carriage reduces reliance on knee musculature to produce positive work/power during physically demanding load carriage tasks (Blacker et al., 2013; Teng and Powers, 2014, 2016). Furthermore, this facilitates reductions in stress and loading at the knee, which is the most commonly injured site for army personnel (Defence, 2000). As a result, a training program targeting the hip musculature has the potential to enhance load carriage performance whilst mitigating injury risks in military personnel.

The purpose of this study was to investigate lower-limb biomechanical changes during a load carriage task, in response to a 10-week evidence-based physical training program. We hypothesised that before training, knee joint moments will increase more pre-march to post-march the load carriage task than after training. Additionally, we hypothesised that after training, lower-limb net joint powers will be maintained from pre-march to post-march compared to before training.

2. Methods

2.1. Participants

Thirteen male civilians (age: 22.4 ± 1.7 years, height: 1.82 ± 0.06 m, mass 83.91 ± 6.5 kg) participated in this study. Participants had no recent (<6 months) acute or chronic injuries at the time of testing. Previous experience with load carriage was not required. Participants provided their written informed consent to the protocol, which was approved by Macquarie University Human Research Ethics Committee (Protocol 5201700406).

3. Inclusion criteria

Study eligibility required participants to meet or exceed the Australian Army Basic Fitness Assessment (BFA) standards for male soldiers ≤ 25 years including: (i) achieve a minimum of 40 push-ups and 70 sit-ups in 2 min each, and (ii) achieve a minimum of level 7.5 on the beep test (to calculate estimated maximal aerobic power (VO_{2max} , $ml \cdot kg^{-1} \cdot min^{-1}$) (Australian Defence Force). Participants were also required to have a body mass ≥ 73 kg, to ensure the 23 kg load carried was <30% body mass, (Mullins et al., 2015).

4. Physical training intervention

A 10-week physical training program was completed by all participants and consisted of up to three resistance training sessions and two weighted walking sessions per week (supplementary

Table 1 outlines resistance training and supplementary Table 2 outlines weighted walking training included in the 10-week training program). An accredited strength and conditioning coach delivered all resistance training sessions, with resistance and weekly progressions tailored to individual abilities. Exercise resistance incrementally increased weekly if participants successfully completed the required number of repetitions and sets for individual exercises. If participants were unable to perform the required repetitions, the number of repetitions performed was recorded and the resistance was adjusted accordingly. Weighted walking sessions were self-directed on a separate day to the resistance training sessions, with load incrementally increasing over the 10-week training program from 0 kg to 25 kg.

5. Procedures

Participants completed a standardised treadmill walking task of 5 km at 5.5 km h^{-1} , wearing a 23 kg torso-borne vest before and after the 10-week training program, which is the equivalent of the Australian Army All Corps minimum employment standard (Australian Defence Force). For testing, participants wore their own footwear, which were standard athletics trainers. Ten successful over-ground walking trials were completed immediately before and immediately after the 5 km walk (<3 min lapse between treadmill to over-ground transition). Prior to walking trials, participants were randomly assigned either their left or right limb to strike the in-ground force plate. To ensure this did not influence foot strike mechanics (e.g., targeting), participants were informed to take their initial step with the randomly allocated limb. During the trials, ground reaction forces (GRF) were collected using an in-ground force plate (Type 9281E, Kistler, Germany), sampling at 1000 Hz, synchronously with three-dimensional (3D) motion data using an eight-camera motion capture system (T40, Vicon, Oxford, UK), sampling at 100 Hz. Trials were successful if the participant: (i) struck the force plate cleanly, (ii) struck the force plate with the randomly allocated left or right limb, and (iii) walked at a speed of $5.5 \text{ km} \cdot \text{h}^{-1} \pm 0.1\%$. Walking speed was monitored using a portable timing gate system (OptoSmart Sensor Porta Kit, Fitness Technology, Adelaide, SA, Australia). Spherical, retro-reflective 14 mm diameter markers, and marker clusters were placed on the participant's torso and bilaterally on landmarks including the head, arms, and legs consistent with a previously validated marker set (Lenton et al., 2017). 3D positions of 12 markers were acquired via static standing calibration and pointer trials (Cappozzo et al., 1995), which were used to define joint centres and track body segments during over-ground walking trials.

6. Data processing

Static and dynamic trial raw marker trajectories data were reconstructed and marker gaps (<10 frames) were interpolated using cubic splines in Vicon Nexus (version 2.7.0). Cleaned experimental data were then exported into Matlab (R2017b, The Mathworks) and processed using a modified version of MOtoNMS (Mantoan et al., 2015). Lower-limb joint centers were defined from static calibration trials using Harrington regression equations (Harrington et al., 2007) at the hip, and the midpoint of the medial and lateral femoral condyles and malleoli at the knee and ankle, respectively. A single gait cycle per successful over-ground trial was determined using the vertical ground reaction force data of the foot in contact with the plate, with the detection threshold set to 20 N for both heel-strike and toe-off. Spatio-temporal and angular variables were determined using a velocity-based algorithm (Zeni et al., 2008). Gait events of heel strike and toe off were automatically detected using changes in the direction of velocity of heel and toe markers. Marker trajectories and GRF's were filtered

using a 4th order zero-lag (Robertson and Dowling, 2003) Butterworth low-pass filter, with a 10 Hz cutoff. Subsequently, marker position data were transformed from the laboratory coordinate system to the global coordinate system used within OpenSim (Delp et al., 2007).

7. Biomechanical modelling

A generic, full-body OpenSim musculoskeletal model was created for each participant (Rajagopal et al., 2016), comprising of three rotational degrees of freedom (DOF) for the hip, one DOF for the knee, and one DOF for the ankle. At the knee joint, abduction/adduction and internal/external rotations were prescribed as a function of knee flexion angle. Generic models were scaled using marker pairs on each body within the model to match the gross anatomy, mass, and inertia of each participant. Scaled models were then used to determine model kinematics and kinetics using inverse kinematics (IK) (Reinbolt et al., 2005) and inverse dynamics (ID). From the IK and ID analyses, joint angular velocities and moments were used to determine sagittal plane hip, knee, and ankle joint powers, which were normalized to each participant's body mass ($W \cdot kg^{-1}$). Hip, knee, and ankle powers were calculated and represented by instantaneous joint power curves which were split into positive (energy generation) and negative (energy absorption) phases throughout the gait cycle (Winter, 1983). From these defined phases, positive and negative joint work ($J \cdot kg^{-1}$) were calculated through numerical integration of the instantaneous joint power curves. The sum of positive and negative hip, knee, and ankle joint work determined total positive (W_j^+) and negative (W_j^-) limb work. Individual joint contributions towards total positive work (W_{tot}^+) and total negative work (W_{tot}^-), throughout the gait cycle were identified through expressing W_j^+ and W_j^- as a percentage of W_{tot}^+ and W_{tot}^- , respectively.

7.1. Statistical analysis

Statistical analysis was performed using IBM SPSS statistics version 25 software for Windows (IBM Corp Armonk, NY, USA). A two-way analysis of variance tested for significant interactions between, and main effects of training and march distance. Normal distribution of data was confirmed using the Shapiro-Wilk test.

Post-hoc Tukey tests with Bonferroni corrections were performed on significant main and interaction effects to identify specific differences between training and march distance measures. Significance was set at $p < 0.05$. Effect sizes were calculated using the partial Eta squared (η_p^2), with small, medium, and large effects defined as η_p^2 between 0.01 and 0.06, 0.06 and 0.14, and greater than 0.14, respectively (Richardson, 2011).

8. Results

Data are presented as mean \pm standard deviation for $n = 13$. Overall, participant adherence to the 10-week training program was 97%, with participants completing 99% of total resistance training sessions and 94% of loaded walking sessions.

8.1. Kinematics

Significant main effects due to the march were observed in sagittal plane kinematics (Fig. 1) for peak hip extension ($p < 0.05$, $\eta_p^2 = 0.73$), hip flexion angle ($p < 0.05$, $\eta_p^2 = 0.41$), knee flexion ($p < 0.05$, $\eta_p^2 = 0.31$), and mean trunk flexion ($p < 0.05$, $\eta_p^2 = 0.85$) angles, in addition to knee pose ($p < 0.05$, $\eta_p^2 = 0.29$) and hip pose ($p < 0.05$, $\eta_p^2 = 0.35$) at heel strike. Specifically, peak hip flexion angle and hip pose at heel strike values decreased from pre-to-post march measures, before and after training, whereas all other variable values increased. A significant main effect of training was observed at the ankle ($p < 0.05$, $\eta_p^2 = 0.52$), with no significant differences found at the hip or knee joint (Table 1). Additionally, a significant interaction effect was found for step width ($p < 0.05$, $\eta_p^2 = 0.40$); compared to before training, values decreased from the pre-to post-march measure, whereas values increased from pre-to-post-march after training. Non-sagittal plane joint angle variables demonstrated no main effects or interactions of distance or training (supplementary Table 3).

8.2. Joint moments, powers, and work

A significant main effect of distance was observed for peak hip extension ($p < 0.05$, $\eta_p^2 = 0.75$) and second peak moment knee extension ($p < 0.05$, $\eta_p^2 = 0.48$), with both values significantly

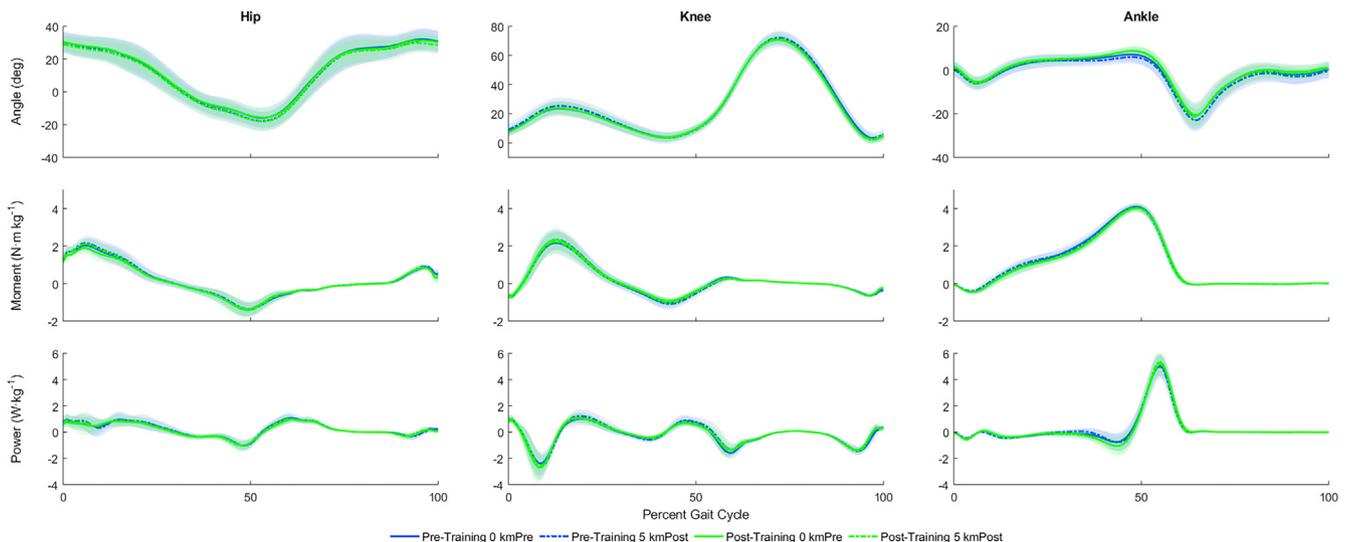


Fig. 1. Mean (lines) and standard deviation (shaded regions) for joint angles, moments, and powers for the hip, knee, and ankle during the 5 km load carriage walking task before and after the 10-week physical training intervention. Asterisks (*) indicate significant differences in variable values.

Table 1

Mean \pm standard deviation magnitudes for spatial-temporal and kinematic variables. *Indicates a significant main effect of distance, #indicates a significant main effect of training, †indicates a significant interaction effect ($p < 0.05$).

Variable	Pre-Training				Post-Training				Effect Size (η_p^2)	
	Pre-march		Post-march		Pre-march		Post-march			
	Mean \pm SD	95% CI (lower, upper)	Mean \pm SD	95% CI (lower, upper)	Mean \pm SD	95% CI (lower, upper)	Mean \pm SD	95% CI (lower, upper)	Training	Distance
<i>Spatial-temporal</i>										
Stride length (m)	1.61 \pm 0.05	1.58, 1.64	1.61 \pm 0.06	1.58, 1.64	1.59 \pm 0.06	1.55, 1.62	1.61 \pm 0.07	1.57, 1.65	0.10	0.15
Stride time (s)	1.06 \pm 0.05	1.03, 1.09	1.07 \pm 0.06	1.03, 1.11	1.07 \pm 0.05	1.04, 1.10	1.07 \pm 0.06	1.04, 1.10	0.06	0.09
Step width (m)	0.06 \pm 0.03	0.05, 0.08	0.05 \pm 0.03	0.04, 0.07	0.07 \pm 0.04	0.04, 0.09	0.07 \pm 0.04 [†]	0.04, 0.09	0.19	0.19
Walk speed (Km/h)	5.45 \pm 0.17	5.35, 5.55	5.42 \pm 0.21	5.29, 5.55	5.35 \pm 0.16	5.25, 5.44	5.39 \pm 0.13	5.33, 5.47	0.18	0.01
<i>Hip</i>										
Extension peak angle (°)	-15.59 \pm 6.89	-19.76, -11.43	-17.71 \pm 6.25*	-21.50, -13.94	-16.32 \pm 6.04	-20.03, -12.62	-18.63 \pm 6.04	-22.29, -14.98	0.03	0.73
Flexion peak angle (°)	34.00 \pm 6.69	29.96, 38.05	33.00 \pm 6.57	29.04, 36.98	31.86 \pm 5.92	28.28, 35.43	30.90 \pm 5.86	27.59, 34.22	0.16	0.41
Pose at heel strike (°)	31.80 \pm 6.00	28.18, 35.42	30.98 \pm 6.01	27.35, 34.61	29.84 \pm 5.63	26.44, 33.24	29.08 \pm 5.89	25.52, 32.64	0.014	0.35
<i>Knee</i>										
Extension peak angle (°)	1.96 \pm 2.94	-0.18, 3.74	1.55 \pm 3.24	-0.41, 3.50	0.90 \pm 2.82	-0.81, 2.60	0.53 \pm 2.7	-1.13, 2.18	0.20	0.07
Flexion peak angle (°)	71.66 \pm 5.50	68.34, 74.98	72.65 \pm 4.70*	69.80, 75.50	71.08 \pm 3.68	68.86, 73.30	71.67 \pm 3.73	69.41, 73.92	0.07	0.04
Pose at heel strike (°)	7.82 \pm 3.41	5.77, 9.88	9.05 \pm 4.35*	6.42, 11.67	7.06 \pm 2.64	5.47, 8.66	7.40 \pm 2.90	5.65, 9.16	0.16	0.05
<i>Ankle</i>										
Dorsiflexion peak angle (°)	8.34 \pm 2.93	6.56, 10.11	7.19 \pm 2.45	5.71, 8.67	9.15 \pm 2.23	7.81, 10.50	9.13 \pm 1.98 [#]	7.94, 10.33	0.52	0.36
Plantarflexion peak angle (°)	-21.74 \pm -7.41	-26.22, -17.26	-23.38 \pm 5.42	-26.65, -20.10	-21.65 \pm 5.87	-25.19, -18.10	-22.21 \pm 5.88	-25.77, -18.61	0.10	0.13
Pose at heel strike (°)	1.17 \pm 3.68	-1.06, 3.40	0.07 \pm 3.45	-2.01, 2.16	1.42 \pm 3.02	-0.40, 3.24	1.46 \pm 3.00	-0.36, 3.27	0.18	0.34
<i>Torso</i>										
Extension peak angle (°)	3.66 \pm 8.89	9.04, 1.70	6.29 \pm 8.50	11.42, -1.16	5.26 \pm 8.03	10.11, -4.02	7.87 \pm 8.03	12.72, -3.02	0.12	0.19
Flexion peak angle (°)	-11.28 \pm 8.47	-16.40, -6.16	-14.13 \pm 7.87	-18.89, -9.38	-11.84 \pm 7.15	-16.16, -7.52	-14.75 \pm 6.61	-18.78, -10.72	0.02	0.83
Pose at heel strike (°)	-8.52 \pm 8.20	-13.47, -3.57	-11.55 \pm 7.40	-16.01, -7.07	-9.24 \pm 7.34	-13.47, -3.57	-12.19 \pm 7.27	-16.58, -7.8	0.03	0.81
Extension/flexion mean angle (°)	-7.93 \pm 8.40	-13.00, -2.86	-10.84 \pm 7.70*	-15.49, -6.18	-8.74 \pm 7.44	-13.23, -4.24	-11.60 \pm 7.10	-15.89, -7.31	0.03	0.85

Table 2

Mean \pm standard deviation magnitudes for external joint moment and power variables. Joint moment values are reported as N·m/kg⁻¹ and power values are reported as W·kg⁻¹. *Indicates a significant main effect of distance, #indicates a significant main effect of training, †indicates a significant interaction effect ($p < 0.05$).

Variable	Pre-training				Post-training				Effect size (η_p^2)	
	Pre-march		Post-march		Pre-march		Post-march			
	Mean \pm SD	95% CI (lower, upper)	Mean \pm SD	95% CI (lower, upper)	Mean \pm SD	95% CI (lower, upper)	Mean \pm SD	95% CI (lower, upper)	Training	Distance
<i>Hip</i>										
Hip extension moment	-2.12 \pm 0.34	-2.33, -1.92	-2.30 \pm 0.33*	-2.50, -2.01	-2.00 \pm 0.32	-2.19, -1.81	-2.18 \pm 0.31	-2.36, -1.99	0.17	0.75
Hip flexion moment	1.37 \pm 0.39	1.13, 1.60	1.33 \pm 0.35	1.12, 1.55	1.43 \pm 0.39	1.19, 1.67	1.47 \pm 0.46	1.19, 1.75	0.07	0.00
Positive hip power	0.67 \pm 0.14	0.59, 0.75	0.60 \pm 0.12*	0.64, 0.76	0.63 \pm 0.12	0.53, 0.68	0.63 \pm 0.11	0.56, 0.69	0.20	0.21
Negative hip power	-0.82 \pm 0.15	-0.91, -0.73	-0.83 \pm 0.20	-0.95, -0.71	-0.81 \pm 0.18	-0.92, -0.71	-0.79 \pm 0.16	-0.89, -0.69	0.08	0.01
<i>Knee</i>										
Knee extension moment	-1.04 \pm 0.26	-0.88, -1.19	-1.10 \pm 0.26*	-0.95, -1.26	-0.97 \pm 0.17	-0.86, -1.07	-1.05 \pm 0.21 [#]	-0.93, -1.18	0.11	0.48
Knee flexion moment	2.30 \pm 0.61	2.70, 1.94	2.36 \pm 0.64	2.75, 1.98	2.31 \pm 0.58	2.66, 1.96	2.33 \pm 0.56	2.66, 1.99	0.00	0.03
Positive knee power	0.39 \pm 0.09	0.33, 0.44	0.44 \pm 0.16	0.34, 0.54	0.37 \pm 0.08	0.32, 0.41	0.40 \pm 0.11	0.33, 0.46	0.25	0.21
Negative knee power	-0.82 \pm 0.15	-0.91, -0.73	-0.83 \pm 0.20	-0.95, -0.71	-0.81 \pm 0.18	-0.92, -0.71	-0.79 \pm 0.16 [#]	-0.89, -0.69	0.08	0.01
<i>Ankle</i>										
Ankle dorsiflexion moment	0.50 \pm 0.06	0.46, 0.54	0.45 \pm 0.12	0.38, 0.53	0.48 \pm 0.10	0.42, 0.54	0.47 \pm 0.10	0.41, 0.53	0.00	0.09
Ankle plantarflexion moment	-4.1 \pm 0.19	-4.21, -3.98	-4.03 \pm 0.22	-4.16, -3.89	-4.08 \pm 0.21	-4.20, -3.95	-4.10 \pm 0.23	-4.23, -3.96	0.03	0.05
Positive ankle power	0.79 \pm 0.09	0.73, 0.84	0.75 \pm 0.11*	0.69, 0.82	0.78 \pm 0.09	0.73, 0.84	0.79 \pm 0.13*	0.71, 0.87	0.06	0.02
Negative ankle power	-0.30 \pm 0.13	-0.38, -0.22	-0.28 \pm 0.87	-0.33, -0.28	-0.33 \pm 0.13	-0.41, -0.26	-0.33 \pm 0.08	-0.38, -0.29	0.55	0.03
<i>Net</i>										
Positive power total	1.84 \pm 0.15	1.75, 1.93	1.90 \pm 0.25	1.75, 2.05	1.75 \pm 0.18	1.64, 1.86	1.81 \pm 0.23	1.68, 1.95	0.26	0.15
Negative power total	-1.38 \pm 0.17	-1.48, -1.27	-1.36 \pm 0.21	-1.49, -1.23	-1.41 \pm 0.18	-1.52, -1.30	-1.40 \pm 0.18	-1.51, -1.29	0.10	0.03

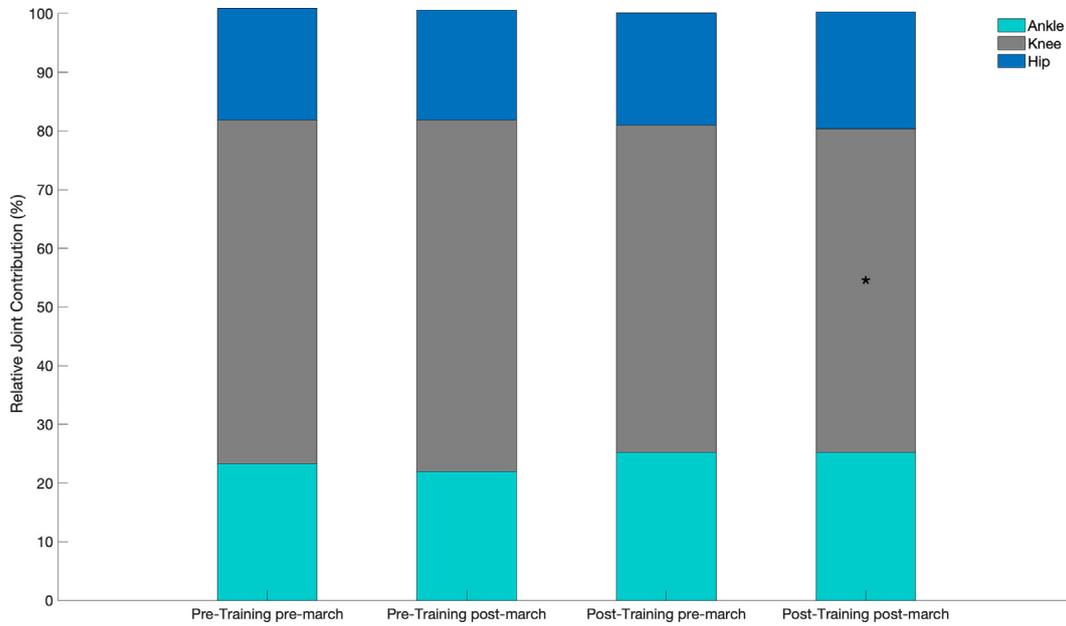


Fig. 2. Relative contributions of hip, knee, and ankle joints to stance phase total mechanical negative power over the 5 km load carriage walking task, before and after the 10-week physical training intervention. Asterisks (*) indicate significant difference in knee joint contribution to total negative power after training.

increasing from pre-march to post-march measures. Percentage contribution of the hip to total positive power increased from pre-march to post-march, while ankle joint contribution towards total positive power decreased ($p < 0.05$, $\eta_p^2 = 0.35$) from pre-march to post-march (Table 2).

Knee extension moment peak values at initial contact of the stance phase (0–40%) significantly increased ($p < 0.05$, $\eta_p^2 = 0.28$) from pre-march to post-march after training. Negative ankle power significantly increased from pre-march and post-march and this

effect was consistent before ($p < 0.05$, $\eta_p^2 = 0.03$) and after training ($p < 0.05$, $\eta_p^2 = 0.55$). Percentage contribution of the knee towards total negative power decreased post-march compared to pre-march after training (Fig. 2). Ankle joint contribution towards total positive power significantly increased at the post-march measurement of the loaded walk after training, increasing from 39.9% (pre-training post-march) to 43.6% (post-training post-march) (Fig. 3). Joint work variables demonstrated no significant interactions or main effects of distance or training.

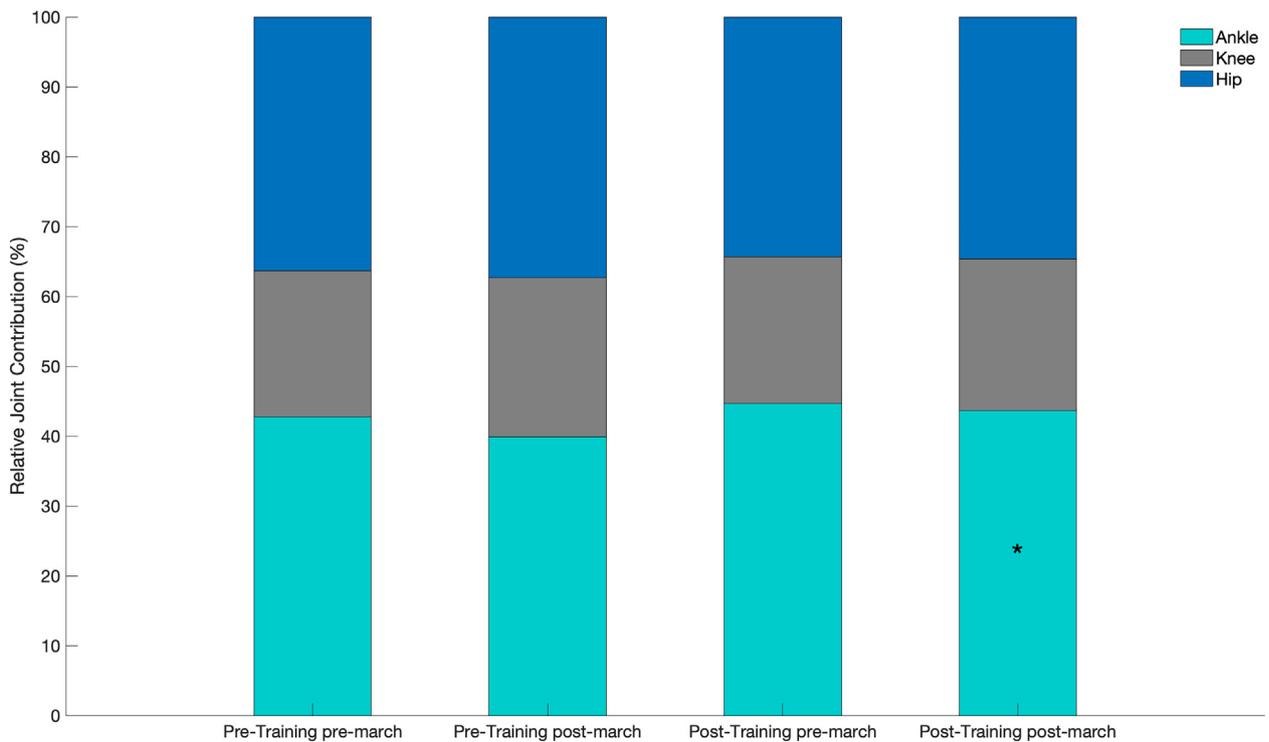


Fig. 3. Relative contributions of hip, knee, and ankle joints to stance phase total mechanical positive power over the 5 km load carriage walking task, before and after the 10-week physical training intervention. Asterisks (*) indicate significant difference in ankle joint contribution to total positive power after training.

9. Discussion

The purpose of this study was to determine lower-limb biomechanical changes during a load carriage task and in response to a 10-week evidence-based physical training program. We found that the main effects of training elicited responses in kinematic and kinetic variables primarily at the knee and ankle joints, with limited changes observed at the hip joint. Joint power contribution shifted distally after training, whereby negative knee power contribution decreased, and positive ankle power increased. To our knowledge, this study is the first to identify and quantify neuromuscular adaptations of the lower-limbs in response to a load carriage physical training program.

Consistent with prior research, hip and knee joint extensor moments increased over the 5 km load carriage task (Harman et al., 2000; Wang et al., 2013; Seay et al., 2014). Changes in extensor moments appear to correspond with observed changes in estimated hip and knee joint level kinematics. Knee flexion increases at heel strike, likely pre-stretching the knee extensors and increasing the quadriceps extension moment arm, resulting in an increased knee extensor moment for a given muscle activation (Wang et al., 2013). After training, hip moments remained stable suggesting training did not impair normal hip biomechanics and may have helped to sustain normal hip motion. In agreement with our first hypothesis, knee joint moments were larger before training compared to after training. Despite larger relative increases in knee moments from pre-march to post-march after training compared to before, these findings may indicate a reduced risk of injury as joint moments were lower throughout the load carriage task after the 10-weeks of training. No other study has compared joint-level changes during a standardized load carriage task in response to a physical training program, meaning further research is required to independently corroborate our findings.

Consistent with our second hypothesis, lower-limb net joint negative powers were unchanged from pre-march to post-march across the load carriage task after training compared to before training. Increases in total positive ankle joint power from 39.9% to 43.6% were observed after the 5 km march after training. The distal shift in power production accompanied by an increase in peak ankle dorsiflexion angle indicates an ankle driven strategy was adopted (Attwells et al., 2006; Majumdar et al., 2010; Silder et al., 2013) after the 10-week training in order to effectively meet the demands of loaded walking (Huang and Kuo, 2014). These findings contrast those reported by Lenton et al. (2019), and are interpreted as a means to accelerate soldier's centre of mass (COM) when performing loaded walking. However, making direct comparisons with previous research is difficult as these studies compare changes in joint power when walking with load vs. without load (Huang and Kuo, 2014), or different load configurations (Lenton et al., 2019), and were not measured over an extended time period. Despite the primary focus of the 10-week training intervention being focused on training the lower-limb musculature, with a specific focus on the hip extensor and flexor muscles, the main effects of training were realised at the ankle. Indeed, shifting relative joint power contributions distally is an efficient strategy to assist with forward progression when carrying evenly distributed load configurations, as increased ankle push-off propels the COM forward and upward (Lewis and Ferris, 2008). This may explain the decreased hip extension moments as the COM is adequately accelerated via an ankle driven mechanism. These findings combined with the maintenance of hip and knee joint powers over the 5 km load carriage task after training may indicate an increased capacity of sustained performance.

Step width decreased from pre-march to post-march during the load carriage task before training and increased over the walk duration after training. This interaction effect suggests after train-

ing, participants increased the base of support to actively increase stability and decreased lower-limb internal loading during load carriage (Birrell and Haslam, 2009; Kinoshita, 1985). Hip and knee kinematics demonstrated the most changes over the duration of the 5 km load carriage walking task. A decrease in peak hip flexion combined with increases in peak knee flexion and mean trunk flexion angles suggests a more upright posture at the hip is adopted over the 5 km walk duration before and after training. These findings contrast previous studies where increases in peak hip flexion were observed, which suggests a lower COM is facilitated to increase stability when carrying external loads (Harman et al., 2000; Birrell and Haslam, 2009). Differences may be accounted for by variations in experimental load configurations as increased peak hip flexion often occurs with greater trunk lean when loads are posteriorly donned compared to the current study where load was evenly donned (Harman et al., 2000; Majumdar et al., 2010). In response to the 10-week training program, changes in lower-limb kinematics were only found at the ankle joint. Ankle dorsiflexion angle increased at pre- and post-march measurements after training, though no significant changes in ankle excursion were observed. Combined, these findings suggest that the 10-week training program elicited no effects on ankle control during the load carriage walking task.

The current study has some limitations that should be acknowledged. Knee flexion and extension DOF were used to determine non-sagittal knee joint motions (abduction/adduction, internal/external rotations, as well as tibial translations) using the same base functions which were then scaled for each subject. This method was chosen as secondary knee motion measures taken from skin-surface marker data is error prone (Benoit et al., 2006). Although the current study recruited civilian participants who were representative of male Australian Army recruits, participants had no prior load carriage experience. We acknowledge that as a result of this, some differences in responses may be apparent between a civilian and an experienced military population. The application of findings may therefore be limited to initial recruits compared to a more experienced soldier population.

In conclusion, this study was the first to investigate and quantify the biomechanical changes in response to a physical training program for a military specific load carriage task. Results demonstrate an evidence-based training program may facilitate injury risk reduction through favourable changes in knee and ankle joint moments and powers, enabling an individual to sustain performance during a load carriage task. As combat-centric roles are now open to female soldiers, the current findings could help direct future research in investigating sex-specific responses to load-carriage tasks. Further investigations could provide insight into lower-limb biomechanical changes specific to males and females in response to evidence-based training for load carriage. Understanding these responses could provide direction in the improvement of Military training and the integration of female soldiers into physically demanding combat roles within the Australian Defence Force.

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Declaration of competing interest

The authors declare no conflicts of interest that could bias this research, including financial and/or personal relationships with other people or organisations.

Appendix A. Supplementary material

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.jbiomech.2019.109341>.

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