Influence of a 12.8-km military load carriage activity on lower limb gait mechanics and muscle activity

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Abstract

The high stress fracture occurrence in military populations has been associated with frequent load carriage activities. This study aimed to assess the influence of load carriage and of completing a load carriage training activity on gait characteristics.

Thirty-two Royal Marine recruits completed a 12.8 km load carriage activity as part of their military training. Data were collected during walking in military boots, pre- and post-activity, with and without the additional load (35.5 kg). Ground contact time, lower limb sagittal plane kinematics and kinetics, and electromyographic variables were obtained for each condition. When carrying load, there was increased ground contact time, increased joint flexion and joint moments, and increased plantar flexor and knee extensor muscle activity. Post-activity, there were no changes to kinematic variables, knee extensor moments were reduced, and there was evidence of plantar flexor muscle fatigue. The observed gait changes may be associated with stress fracture development.

KEYWORDS: load carriage; EMG; kinetics; kinematics; stress fracture

Practitioner Summary:

This study identified gait changes due to load carriage and after a military load carriage training activity. Such activities are associated with lower limb stress fractures. A pre-post study design was used. Gait mechanics changed to a greater extent when carrying load, than after completion of the activity when assessed without load.

Introduction

Regular load carriage activities are a key feature of military recruit training programmes. The high injury rates (Birrell, Hooper, and Haslam 2007; Knapik et al. 1992; Orr et al. 2014) and specifically the high stress fracture rates in military populations (Knapik, Reynolds, and Harman 2004; Orr et al. 2014) have been associated with these load carriage activities. Royal Marine recruits (United Kingdom) have a high rate of lower limb stress fracture, with the tibia and metatarsals the most commonly fractured sites (Davey et al. 2015). Mechanisms by which load carriage activities may increase the risk of lower limb stress fracture are unclear.

A review of the influence of carrying additional load on walking gait characteristics revealed a decrease in step length, an increase in the time spent in double support (time during which both feet are in contact with the ground), and an increase in forward lean of the trunk when carrying load (Seay 2015). These mechanisms are suggested to increase stability during load carriage activity. Increased trunk lean coincides with an increase in maximal hip flexion. Other findings regarding the influence of load carriage on lower limb kinematics are conflicting. Increased knee flexion angle has been

observed during overground and treadmill walking whilst carrying load (Harman et al. 2000; Kinoshita 1985; Silder, Delp, and Besier 2013). However, sagittal plane knee range of motion has been reported both to increase (Knapik, Harman, and Reynolds 1996), and to decrease (Birrell and Haslam 2009; Harman et al. 2000; Polcyn et al. 2000) with load, while Polcyn et al (2000) found no change. Similarly, ankle dorsiflexion angle has been found to increase (Silder, Delp, and Besier 2013) and not to change (Birrell and Haslam 2009; Ghori and Luckwill 1985; Harman et al. 2000) with the addition of load. These conflicting findings may be influenced by the different load carriage protocols, including overground and treadmill walking, speeds ranging from 1.1 $m.s^{-1}$ to 1.7 $m.s^{-1}$, different footwear conditions, and the inclusion or exclusion of weapon carriage. This highlights the need for population-specific assessment of gait changes as a result of load carriage.

Load carriage (at least 30% body weight) has been shown to increase ankle plantarflexor, knee extensor and hip extensor moments, where the observed increases in knee moments were considerably greater than the increases in ankle and hip moments during early stance (Seay 2015; Quesada et al. 2000). With a smaller load of 15 kg, only the knee extensor moment increased, with no change in the ankle plantarflexor and hip extensor moments observed (Seay 2015). The knee appears to be most sensitive to increased load during early stance, whereas the ankle and hip contribute more to the push-off phase (Seay 2015). Increased plantar flexor and knee extensor muscle activity have also been reported during load carriage activities (Harman et al. 2000; Ghori and Luckwill 1985). It was suggested that increased lower

limb muscle activity may be a mechanism to prevent changes in kinematics during load carriage (Ghori and Luckwill 1985). However, increased muscle activity has also been reported to accompany increased sagittal plane ankle and knee flexion (Harman et al. 2000), contradicting this suggestion.

Although there is a wealth of evidence regarding gait changes that occur during load carriage activity, there is only limited evidence regarding the changes that occur following completion of a prolonged load carriage activity. Increased peak ankle dorsiflexion and knee flexion, and reduced knee extensor moments have been reported following 40 minutes of treadmill marching (1.67 m.s^{-1}) whilst carrying a load of up to 30% of body weight (Quesada et al. 2000). The altered kinematics are indicative of an inability to maintain the pre-activity gait pattern, likely as a result of fatigue, while the reduced knee extensor moments may be the result of quadriceps muscle fatigue. Direct evidence regarding the influence of a prolonged load carriage activity on muscular fatigue is limited. Following a 2 km treadmill march (2.22 m.s^{-1}), reduced median frequency of the gastrocnemius lateralis and peroneus longus muscles was observed (Gefen 2002), indicative of muscular fatigue (Winter 2009). These muscles act as plantar flexors during the push-off phase of walking, and play a protective role against increasing tibial (Milgrom et al. 2007) and metatarsal (Arndt et al. 2002; Sharkey et al. 1995) strain. This protective ability may be impaired as a result of fatigue.

Identification of lower limb gait changes that occur following a load carriage activity in military populations is required. Such information may be used to inform interventions

designed to minimise these potentially detrimental gait changes. The aim of this study was to assess the influence of load carriage, and of completing a prolonged load carriage activity, on lower limb kinematics, kinetics and EMG activity, following a real-life Royal Marine recruit training activity. It was hypothesised that carrying load would result in increased knee flexion, but would not change ankle dorsiflexion angle. Furthermore, it was hypothesised that carrying load would result in greater plantar flexor and knee extensor moments, and greater plantar flexor and knee extensor muscle activity. It was hypothesised that following a military load carriage training activity, recruits would display increased peak ankle dorsiflexion and knee flexion, as well as reduced knee extensor moments and evidence of plantar flexor muscle fatigue.

Methods

Participants

Thirty-two Royal Marine recruits volunteered to participate in the study which took place in week-21 of their 32-week recruit training programme. All volunteer recruits were male, as the Royal Marines only allow male recruits. All recruits were injury-free and had not been removed from training for any reason at the time of data collection. Within these criteria, recruits were randomly selected from four troops (eight per troop). Recruits had a mean (SD) age: 23.8 (3.5) years, body mass: 79.2 (6.5) kg, height: 1.8 (0.1) m and body mass index: 25.8 (1.9) kg.m^{-2}. The study was approved by the Ministry of Defence Research Ethics Committee (Reference: 367/GEN/12), and all recruits gave voluntary informed consent.

Protocol

As part of the Royal Marines recruit training programme, all recruits are required to complete regular prolonged load carriage activities. One such activity was selected as the protocol for this study, allowing high ecological validity. The selected activity involved 'yomping', which is predominantly a walking activity, interspersed with short periods of running, over uneven terrain. The total distance covered was 12.8 km and recruits were required to carry an additional load of 35.5 kg. This activity occurred in week-21 of the 32-week training programme. The approximate duration of the activity was 150 minutes, thus the average speed was 5.12 $km.h^{-1}$ (1.4 $m.s^{-1}$). The pace of the activity was determined by members of a training team. Recruits wore military issue clothing and boots throughout the activity. The activity paused approximately every hour for water breaks, which were less than five minutes in duration. Recruits were encouraged to eat regularly throughout the activity. The load consisted of a Bergen (large rucksack) and webbing (worn like a belt to carry additional military kit) with a combined mass of 31.3 kg, and a weapon (4.2 kg), which was attached to a sling and held in front of the body throughout the activity.

Data collection

Synchronised force, kinematic and EMG data were collected from the left leg of recruits during walking in military boots, both pre- and post- the load carriage activity. Post-activity, recruits replaced their combat trousers with shorts and wore their military boots

for all trials. Recruits first completed familiarisation trials until successful trials were repeated. A successful trial was within the correct speed range (1.4 m.s^{-1} (±5%)), with the left foot fully contacting a force plate without adjusting stride length. The walking speed represented the average speed of the training activity. Ten trials were collected per recruit pre-activity, and a further ten post-activity. Five of the ten trials involved carriage of 35.5 kg of additional load, and five were unloaded. Unloaded trials involved recruits walking in military boots, having removed their Bergen, their webbing, and their weapon. During loaded trials, recruits held their weapon in the position required of them throughout the activity. During unloaded trials, recruits simulated this 'holding' position, without the weapon. Order of loading condition was counterbalanced, such that half the study population completed the trials with additional load first, and the remaining half completed unloaded trials first. Data collection post-activity commenced immediately upon completion of the activity for each troop. Based on existing literature (Bisiaux and Moretto 2008; Horita et al. 1999; Tsai et al. 2009) and pilot work, 90 minutes was considered an appropriate maximum time between completion and data collection. This was achieved with 31 of 32 recruits (range 6 – 98 min, mean (SD): 49 (26) min). Energy and water intake were not controlled post-activity, as they were not controlled during the activity.

Kinematic and force data were collected at 200 Hz with two Coda Mpx30 units (CodaMotion, Charnwood Dynamics, UK) and an AMTI (OR6-7-2000, Waterway, MA, USA) force plate. Eleven active markers were positioned as shown in Figure 1. Markers were secured using Micropore^{TM} tape (3M, USA). Positions were identified with

permanent pen to allow reliable replacement post-activity, with the exception of the greater trochanter marker, which was repositioned by palpation. The lateral malleolus of the ankle was palpated through the military boot, which was securely fastened on the foot. A standing trial was recorded both pre- and post-activity in the unloaded condition, allowing adjustment of joint angles during both loaded and unloaded walking, relative to a 'neutral' standing position.

EMG data (Trigno Wireless System, Delsys, Boston, MA, UK) were collected at 4000 Hz from the vastus lateralis (VL); a knee extensor muscle, and the gastrocnemius lateralis (GL), and peroneus longus (PL) muscles, both of which are plantar flexors during walking. SENIAM guidelines were followed, and wireless EMG sensors were secured using purpose-made adhesive interfaces (Biosense Medical Ltd, Essex, UK) and Micropore™ tape. The position of each sensor was marked with permanent pen to allow reliable replacement post-activity. EMG sensors were necessarily moved between data collection sessions, thus both within- and between-session reliability of EMG variables were assessed. Intraclass correlation coefficients (ICC) were calculated using a single measure, two-way random effects model. Only variables which had ICC values ≥ 0.7 were included in further analyses. Both the VL and GL EMG data had poor between-session reliability, and therefore could only be used to compare loaded and unloaded conditions.

Data analysis

Data were analysed using customised MATLAB scripts (R2012a, The MathWorks Inc. Natick, MA, USA). Kinematic data were filtered with a cut-off frequency of 12 Hz. Force data were filtered at the same frequency as kinematic data (Bisseling and Hof 2006). Ground contact time was determined from force data (vertical force > 10 N indicating stance). Ankle kinematic variables were: touchdown and peak dorsiflexion angle, and range of motion (from peak plantar flexion to peak dorsiflexion). Knee kinematic variables were: touchdown and peak flexion angle, and range of motion (from touchdown to peak). Peak flexion angle during the first two-thirds of stance was reported, in order to exclude the flexion angle at push-off. Peak ankle plantar flexor and knee extensor moments were obtained. Joint moments were normalised to body mass.

EMG data were filtered with a fifth-order Butterworth band-pass filter between 5 and 500 Hz (according to Seniam recommendations (Rose 2011)) and full-wave rectified, then low-pass filtered at 10 Hz (Konrad 2005). The EMG frequency analysis was conducted using a fast Fourier transform algorithm. This was obtained for ground contact time, and the period of time equal to ground contact time that occurred prior to initial contact (total time of 2 x ground contact time). Maximum amplitude and integrated EMG (iEMG) provided an indication of muscular activity. Maximum amplitude was normalised to an ensemble averaged mean value taken from each 5% of stance, during unloaded pre-activity walking trials. iEMG values were normalised to the maximum iEMG value from each recruit, obtained during five unloaded pre-activity walking trials. Mean and median frequencies were used as gold standard indicators of muscular fatigue (Phinyomark et al. 2012), thus only the influence of time (pre- vs. post-activity) on these variables was

assessed. Mean frequency was calculated as the sum of the product of the power density values and the frequency, divided by the total sum of the power spectrum (Phinyomark et al. 2012). Median frequency was the value such that the sum of all power densities below and above the median were equal (Winter 2009).

Statistical analysis

Statistical analyses were conducted (α = 0.05) using SPSS software (Version 16.0, SPSS Inc., Chicago, IL, USA). Two-way repeated measures ANOVAs were conducted to determine the effect of load carriage (loaded and unloaded) and time (pre- and post-activity). The assumption of sphericity was met as there were only two levels for each within-subject factor. Order of loading was the between-subjects variable. Where there was no significant effect of order of loading, the variable was removed before repeating the ANOVA. Effect size (r, $0.01 \leq$ small < 0.3, $0.3 \leq$ medium < 0.5, $0.5 \leq$ large (Cohen 1988)) was calculated using the equation below where $F(1, dfR)$ is the F-ratio for the effect and dfR is the degrees of freedom for the error term on which the F-ratio is based (Field 2005).

$$r = \frac{F(1, dfR)}{F(1, dfR) + dfR}$$

For those variables where there was a main effect for load or time, bivariate correlation analyses were conducted to identify any associations between the amount of recovery time and the change in variable from pre- to post-activity. This was used to indicate whether recovery time influenced the findings.

Results

There was no effect of the order of trials on any variables. Mean (SD) values are displayed for variables where there was a main effect in Table 1. Main effects for load are presented in Table 2. When loaded compared with unloaded, there was a longer ground contact time, greater dorsiflexion range of motion, and greater knee flexion angle at touchdown and peak. There were also greater peak ankle plantar flexor and knee extensor moments. iEMG of the GL muscle was greater when loaded, as was the maximum VL amplitude, indicating greater activity of these muscles. Main effects for time are also presented in Table 2. Post-activity there was no difference in sagittal plane ankle or knee kinematics compared with pre-activity. Post-activity, knee extensor moments were reduced, as was PL mean frequency, indicating PL muscle fatigue. Main effects were associated with large effect sizes. Bivariate correlation analyses revealed that the association between recovery time and change in value from pre- to post-activity for all variables was $r \leq \pm 0.31$ ($P > 0.05$), suggesting recovery time did not influence the findings.

Discussion

This was the first known study to assess changes to lower limb kinematics, kinetics and muscle activity as a result of a real-life prolonged load carriage training activity. The study benefitted from a homogeneous population all of whom had completed the same 20 weeks of military training within a highly-controlled living and working environment,

prior to data collection. The effects of carrying load and the effects of completing the activity were both assessed.

Effects of load carriage

The increased knee flexion angle when carrying load is consistent with previous findings (Harman et al. 2000; Kinoshita 1985; Polcyn et al. 2000; Silder, Delp, and Besier 2013) and with the hypothesis. There is now strong evidence to suggest that knee flexion increases when carrying load, within a range of walking speeds (1.1 m.s^{-1} – 1.5 m.s^{-1}), during both overground and treadmill walking, and with or without weapon carriage. It is suggested that this is a mechanism to maintain postural stability by lowering the body centre of mass (Harman et al. 2000). This could also be achieved through increased ankle dorsiflexion when carrying load, which was observed in the present study, in contrast to the hypothesis and to previous studies (Birrell and Haslam 2009; Ghori and Luckwill 1985; Harman et al. 2000). This difference in findings may be influenced by the particular military boot worn in each study, where military boot stiffness has been reported to predominantly influence ankle joint kinematics (Cikajlo and Matjacić 2007). In addition to helping to maintain postural stability, increased knee and ankle joint flexion will increase the dissipation of reactive forces, thereby minimising the demand on the structures of the lower limb (de Fonseca et al. 2007).

In order to control the increased flexion at the ankle and knee joints, there was a corresponding increase in ankle plantar flexor and knee extensor moments (40% and

72% respectively) as hypothesised. Harman (2000) reported similar increases in plantar flexor and knee extensor moments during loaded compared with unloaded walking (38% and 98% respectively) with an additional load of 49% of body weight (compared with an additional load of approximately 44% of body weight in the present study). The results from the present study, along with previous findings (Harman et al. 2000) suggest that the knee extensor muscles assume a greater proportion of the burden of load carriage than the ankle plantar flexor muscles. Seay (Seay 2015) further explains that the knee extensor moments act to resist the effects of additional load in early stance, before the musculature at the ankle respond.

EMG data indicated increased ankle plantar flexor and knee extensor muscle activity whilst carrying load, in support of the hypotheses, and consistent with earlier findings (Harman et al. 2000). The quadriceps muscles act eccentrically during early stance to control knee flexion, whilst the plantar flexors act concentrically towards the end of stance to contribute to push-off (Harman et al. 2000; Seay et al. 2014). The increased gastrocnemius lateralis muscle activity suggests a greater plantar flexor force production was required for push-off when carrying a load of 35.5 kg, consistent with earlier findings which reported an increase in gastrocnemius activity with loads up to 33 kg (Harman et al. 2000). Interestingly, in this earlier study it was found that muscular activity of the gastrocnemius did not increase further when load was increased to 47 kg, whereas muscular activity of the quadriceps continued to increase (Harman et al. 2000). This suggests that carrying heavier loads is more demanding on the knee joint musculature than that of the ankle. The increases in joint moments and muscular

activity that occur when carrying load will exacerbate plantar flexor and knee extensor muscle fatigue during load carriage.

An observed difference that was not hypothesised was the longer ground contact time during loaded walking than unloaded walking. Loading of the lower limb is increased when carrying load (Harman et al. 2000; Quesada et al. 2000). Therefore a longer ground contact time would increase the amount of time the limb is subjected to increased loading. This may result in increased loading of lower limb structures, such as increased metatarsal compression (Arndt et al. 2002). Alternatively, a longer ground contact time may be a protective mechanism to lower the rate of loading on the internal structures in order to reduce the potential negative effects of increased load magnitude. It is therefore unclear whether a longer ground contact time is associated with the development of injury, or if it is a mechanism to minimise the risk of injury. Assessment of loading rates under the metatarsals in future work would help to improve this understanding.

The results discussed so far present evidence that walking whilst carrying load is more physically demanding than unloaded walking, requiring greater plantar flexor and knee extensor moments, and consequently increased plantar flexor and knee extensor muscle activity. A weight-bearing activity of 12.8 km would likely result in fatigue without the addition of load. The increased demand of carrying load would result in earlier fatigue onset and greater levels of fatigue upon completion of the activity, than would be

observed when completing an activity of the same distance and in the same time, without load carriage. Interventions to increase muscular strength and endurance of the plantar flexor and knee extensor muscles may be beneficial to this population. This is in support of a previous suggestion that military populations consider training to improve the endurance of muscles "related to the 'marching' pattern of fatigue" (Gefen 2002).

Effect of completing the load carriage activity

Contrary to the hypotheses, there were no differences in sagittal plane kinematics as a result of the load carriage activity. This conflicts with previous findings of reduced peak ankle dorsiflexion and knee flexion following just 40 minutes of walking with load (Quesada et al. 2000). This previous study involved male Army Reserve Officer Training Corps volunteers, who may be less accustomed to load carriage activities than Royal Marine recruits, and therefore less able to maintain the kinematic patterns demonstrated pre-activity. Following the activity, Royal Marine recruits were able to maintain the gait kinematics displayed prior to the activity, when unloaded, but these could not be maintained when carrying load. Load carriage is therefore a greater contributor to altered lower limb mechanics in this Royal Marine recruit population, than the 'fatiguing' effects of completing a prolonged load carriage activity.

The observation that knee extensor moments reduced post-activity was in support of the hypothesis, and earlier findings following a 4 km load carriage treadmill march (1.7 m.s^{-1} with 15% and 30% of body weight carried (Quesada et al. 2000)).This change may be the result of quadriceps muscle fatigue (Quesada et al. 2000), although this

could not be assessed in the present study, due to poor between-session reliability of vastus lateralis EMG variables. This reduction in knee extensor moments presents evidence of a reduced ability of the quadriceps muscles to dissipate energy, which may result in increased demand at other parts of the kinetic chain (de Fonseca et al. 2007). As both knee extensor and plantar flexor moments are key contributors to walking (Kepple, Lohmann Siegel, and Stanhope 1997), the reduction in knee extensor moments post-activity, whilst ankle plantar flexor moments were maintained, suggests an increased reliance on the ankle plantar flexor moments to maintain postural stability. This may increase the level of fatigue or the rate of fatigue development in the plantar flexor muscles.

A reduction in mean frequency of the peroneus longus muscle post-activity is indicative of muscular fatigue, although this evidence would be stronger with the addition of reliable gastrocnemius lateralis EMG frequency spectrum data. Gastrocnemius lateralis muscle activity was greater during load carriage which would be likely to accelerate the onset of muscle fatigue. The observed peroneus longus muscle fatigue may have been the result of a greater contribution of the plantar flexor muscles while knee extensor moments were reduced. Plantar flexor muscle fatigue has previously been associated with both tibial (Milgrom et al. 2007) and metatarsal (Arndt et al. 2002; Sharkey et al. 1995) stress fractures. Given its contributing role to ankle plantar flexion, peroneus longus muscle fatigue following a prolonged military load carriage activity may therefore be associated with the high rate of lower limb stress fracture in military populations. Although the hypothesised plantar flexor muscle fatigue was only partially supported,

this adds weight to the suggestion that interventions to reduce fatigability of the plantar flexor muscles during load carriage activity should be considered.

Limitations

Assessment of muscular fatigue was important in the present study, but was limited by the difficulties in collecting reliable EMG data between sessions. The EMG sensors necessarily had to be removed and replaced between sessions, and this could not be reliably performed for all the muscles of interest. Furthermore, the quality of EMG data collection is greatly influenced by the data collection conditions (Konrad 2005). Perspiration levels would have been greater post-activity, making differences between conditions more difficult to observe.

The use of a real-life military training activity in the present study allowed identification of gait changes that are likely to occur in the wider Royal Marine recruit population, giving ecologically valid findings. This may help to explain some of the mechanisms for the prevalent injuries within this population. In order to assess the real-life influences of the training activity, it was not modified in any way, such that all recruits within each troop completed the activity together. Therefore recovery time varied within the eight recruits per troop from whom data were collected in each session. Bivariate correlation analyses indicated that the findings of this study were not influenced by recovery time.

This study only included recruits who had not been injured during the first 20 weeks of recruit training, thus they were arguably more resistant to injury than many Royal Marine recruits. More pronounced or additional changes may have been observed following this prolonged load carriage activity in recruits who were more susceptible to lower limb injury. Recruits at greater risk may have already sustained an injury prior to week-21, when the protocol activity for this study took place, and as such would have been excluded from the study. However, stress fractures tend to occur later throughout the Royal Marine training programme than other military training programmes (Davey et al. 2015), thus the volunteer recruits likely provided a useful representation.

Females were not included in this study, as the Royal Marine recruit population is all-male. The influence of load carriage on gait characteristics and muscle activity has been reported to be similar between males and females (Silder, Delp, and Besier 2013). However, the demand of completing a prolonged load carriage activity may differ between males and females, and further investigation is warranted.

Summary

Load carriage during a military training activity resulted in increased ground contact time and altered sagittal plane lower limb mechanics. Greater dorsiflexion range of motion and knee flexion angle, greater plantar flexor and knee extensor moments, and greater activity of the plantar flexor and knee extensor muscles were observed during walking with load compared with unloaded walking. Following the activity, there were no

differences in kinematic variables, there was a reduction in knee extensor moments, and evidence of peroneus longus muscle fatigue. These observed gait changes may partly explain the association between military load carriage activities, and lower limb stress fracture development. Interventions to increase ankle plantar flexor and knee extensor muscle strength and endurance should be assessed.

Acknowledgements

Conflict of Interest Statement

We are not aware of any conflict of interest related to the manuscript and its publication.

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Table 1: Mean (SD) values and main effects for kinematic, kinetic and EMG variables

Variable	Pre-Unloaded	Pre-Loaded	Post-Unloaded	Post-Loaded	Effects
Kinematic variables					
Dorsiflexion range of motion (degrees)	18.7 (5.8)	20.2 (5.9)	16.8 (3.4)	18.1 (3.6)	Load
Peak knee flexion (degrees)	17.6 (4.7)	19.1 (5.9)	18.7 (4.6)	21.1 (4.9)	Load
Knee flexion touchdown (degrees)	-1.8 (4.2)	0.6 (5.0)	-0.2 (3.3)	2.3 (4.3)	Load
Kinetic variables					
Ground contact time (ms)	655 (30)	674 (31)	642 (27)	673 (27)	Load
Peak plantar flexor moment (Nm.kg^{-1})	1.73 (0.60)	2.46 (0.72)	1.77 (0.20)	2.46 (0.36)	Load
Peak knee extensor moment (Nm.kg^{-1})	1.59 (0.39)	2.17 (0.50)	1.37 (0.34)	1.89 (0.39)	Load Time
EMG variables					
VL max (multiples of mean)	3.99 (1.12)	6.00 (1.30)	4.52 (2.96)	6.79 (2.18)	Load
GL iEMG (multiples of max)	0.86 (0.08)	1.39 (0.35)	2.12 (5.78)	2.79 (6.71)	Load
PL mean frequency (Hz)	126.00 (21.59)	136.95 (22.10)	87.20 (51.04)	89.75 (53.71)	Load Time

Note: A greater knee flexion angle indicates a more flexed knee
VL: vastus lateralis; BF: biceps femoris; GL gastrocnemius lateralis; TA: tibialis anterior; PL: peroneus longus

Table 2: F values, p values and effect sizes of those variables for which there was a main effect for load carriage and time

Main effect for load carriage				
Variable	**Change with load**	**F value**	**P**	**r**
Kinematics				
Ground contact time	Longer	$F(1,30) = 97.332$	< 0.001	0.874
Dorsiflexion range of motion	Greater	$F(1,14) = 31.474$	0.039	0.832
Knee flexion touchdown	More flexed	$F(1,13) = 15.883$	0.002	0.742
Peak knee flexion	More flexed	$F(1,14) = 13.078$	0.003	0.695
Kinetics				
Peak plantar flexor moment	Greater	$F(1,21) = 251.983$	< 0.001	0.961
Peak knee extensor moment	Greater	$F(1,19) = 117.683$	< 0.001	0.928
EMG				
VL max	Greater	$F(1,9) = 14.831$	0.004	0.662
GL iEMG (multiples of max)	Greater	$F(1,24) = 23.669$	< 0.001	0.704
Main effect for time				
Variable	**Change with time**	**F value**	**p**	**r**
Kinetics				
Peak knee extensor moment	Reduced post-	$F(1,19) = 8.540$	0.008	0.557
EMG				
PL mean frequency	Lower post-	$F(1,12) = 5.989$	0.031	0.577

Posterior View   **Anterior View**

Greater trochanter

Medial epicondyle
of the femur

Lateral epicondyle
of the femur

Posterior
lower leg

Proximal third
metatarsal

Anterior midshaft
of the tibia

Lateral malleolus
of the ankle

Superior
calcaneus

Fifth metatarso-
phalangeal joint
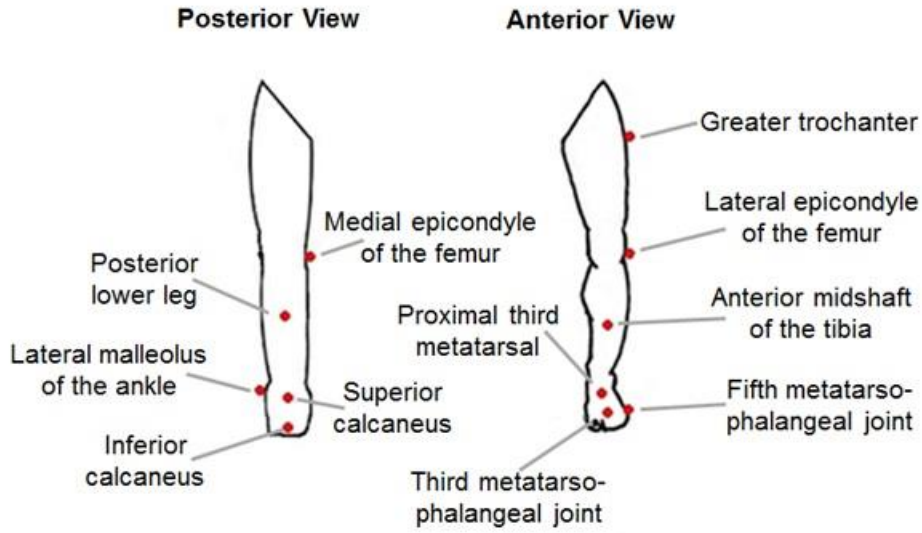
Inferior
calcaneus

Third metatarso-
phalangeal joint

Figure 1: Coda marker positions displayed on the left leg