Accepted Manuscript

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PII:	S0966-6362(17)30747-6
DOI:	http://dx.doi.org/doi:10.1016/j.gaitpost.2017.07.044
Reference:	GAIPOS 5728
To appear in:	*Gait & Posture*
Received date:	7-11-2016
Revised date:	12-7-2017
Accepted date:	16-7-2017

Please cite this article as: Roberts Andrew, Roscoe David, Hulse David, Bennett Alexander N, Dixon Sharon.Biomechanical differences between cases with chronic exertional compartment syndrome and asymptomatic controls during walking and marching gait.*Gait and Posture* http://dx.doi.org/10.1016/j.gaitpost.2017.07.044

Biomechanical differences between cases with chronic exertional compartment syndrome and asymptomatic controls during walking and marching gait

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Highlights:

- Patients with CECS are shorter and take relatively longer strides than controls.
- Kinematic differences are found at the ankle, but not at more proximal joints.
- These differences may play an important role in the development of CECS.

Abstract

Chronic exertional compartment syndrome is a significant problem in military populations that may be caused by specific military activities. This study aimed to investigate the kinematic and kinetic differences in military cases with chronic exertional compartment syndrome and asymptomatic controls.

20 males with symptoms of chronic exertional compartment syndrome of the anterior compartment and 20 asymptomatic controls were studied. Three-dimensional lower limb kinematics and kinetics were compared during walking and marching.

Cases were significantly shorter in stature and took a relatively longer stride in relation to leg length than controls. All kinematic differences identified were at the ankle. Cases demonstrated increased ankle plantarflexion from mid-stance to toe-off. Cases also demonstrated less ankle inversion at the end of stance and early swing phases. Lower ankle inversion moments were observed during mid-stance.

The anthropometric and biomechanical differences demonstrated provide a plausible mechanism for the development of chronic exertional compartment syndrome in this

population. The shorter stature in combination with the relatively longer stride length observed in cases may result in an increased demand on the anterior compartment musculature during ambulation. The results of this study, together with clinical insights and the literature suggest that the suppression of the walk-to-run stimulus during group marches may play a significant role in the development of chronic exertional compartment syndrome within a military population. The differences in joint angles and moments also suggest an impairment of the muscular control of ankle joint function, such as a reduced effectiveness of tibialis anterior. It is unclear whether this is a cause or consequence of chronic exertional compartment syndrome.

Keywords: exercise-induced leg pain; chronic exertional compartment syndrome; biomechanics; anthropometry; military training.

Funding: This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Introduction

Chronic exertional compartment syndrome of the leg was first described in 1956 [1]. It is an overuse condition presenting as pain in the lower limb, associated with the muscles contained within the myofascial compartments of the shank. The anterior compartment is most frequently affected [2]. While numerous studies have attempted to understand the pathophysiology of CECS [3-6], few studies have identified potential risk factors. Chronic exertional compartment syndrome poses a significant clinical burden in the military making this population suitable for investigating these potential factors [7].

The North Atlantic Treaty Organization (NATO) recently identified walking and marching as key common tasks performed in recent and current military missions [8]. As such these activities are also commonly performed during military training. These tasks have also previously been associated with CECS [7,9,10]. The exact definition of a march varies; however it typically requires a fast walking gait with a set stride length and cadence to allow the movement of a group of individuals at a set pace. Personnel often undertake organised group marches that prepare them for deployment and the completion of the annual fitness tests that, for example in the Army and Royal Marines, require 2-3 hours of marching at 1.8m/s [11,12]. A large proportion of military training also involves walking between other planned activities [13,14].

Chronic exertional compartment syndrome has been defined as a condition where elevated intramuscular compartment pressure (IMCP) during exercise impedes local blood flow leading to ischaemia and impaired neuromuscular function within the compartment [15,16]. Recent evidence has reported improved diagnostic criteria over existing methods for CECS using continuous IMCP measurement during exercise [17].

IMCP can be increased through changes in compartment compliance, compartment fluid content or muscle activity [3,18,19]. Recent evidence has shown that in CECS, IMCP is elevated above that of controls immediately on standing at rest [17]. This suggests that there is an increased stiffness of a passive structure, presumably the fascia, which results in reduced compartment compliance. This divergence is amplified during a symptom-provoking exercise challenge [17].

Biomechanical factors have been considered to play a role in the development of CECS for a long time [20,21]; however these have never been directly studied. We therefore aimed to examine potential biomechanical differences during walking and marching between cases and controls to provide evidence regarding the role of biomechanical factors in the aetiology of this condition. The anterior compartment musculature is responsible for movements at the ankle; these angles and moments were therefore of prime interest. As this was the first study to examine the biomechanics of CECS patients we also explored the angular and moment data of joints further up the kinetic chain. These more proximal joints have also been the subject of recent biomechanical interventions for CECS [21-23].

Materials and methods

20 male cases (PT) with bilateral symptoms consistent with CECS of the anterior compartment of the leg and 20 asymptomatic controls (CON) participated following informed consent. The diagnosis of CECS was established from typical symptoms, with clinical examination and MRI excluding other pathologies. All participants were recruited from the UK armed forces with significant experience of marching. Cases were recruited from the XXXXX clinic at the XXXXX. Ethical approval was granted by the MOD Research Ethics Committee.

The inclusion criteria were: Male; Aged 18-40 (representing the typical age-range of UK military service personnel); BMI<35; and, no true leg length discrepancy >2cm. Cases required the following: symptoms of exercise-induced leg pain consistent with a diagnosis of anterior compartment CECS; a negative MRI of the affected limb(s) and lumbar spine; no diagnosis other than CECS more likely; absence of multiple lower limb pathologies; and, no previous lower limb surgery. Cases had higher IMCP than controls (114±32mmHg vs 68.7±22mmHg) and reported pain (scale: 0-10) in the anterior compartment of 5.1±2.6 within 10 minutes of loaded marching as previously reported [17]. Controls were included when they were able to run for a minimum of 20 minutes and had: no lower limb pain in the previous 12 months; no current pain at any site, including during exercise activities; and no reliance on orthotics.

Measurements of leg length, height and body mass were performed using a tape measure, stadiometer (SECA, UK) and medical grade scales (SECA, UK) respectively. The same operator, using the same landmarks and techniques assessed all subjects.

Motion capture

Fifteen body segments (feet, shank, thigh, pelvis, trunk, head, upper arm, forearm and hand) were defined using retro-reflective markers placed on specific anatomical landmarks by the same operator. The head, upper arm, forearm and hand were not analysed as part of this study. Data were collected using a 10 camera (4xT160, 4xT40-S, 2xT10) 3D motion analysis system (Vicon MX system, Oxford Metrics Ltd., Oxford, England) at a sampling frequency of 120 Hz. Ground reaction forces were collected using three force plates (AMTI, OR-6, USA) at a sampling frequency of 1200 Hz.

Following a static calibration trial, participants performed traverses of the laboratory while walking and marching until a minimum of 10 complete cycles for each leg had been captured [24]. Following familiarisation, participants were asked to walk at their natural pace (expected speed c.1.4m/s) and march 'as if they were doing their military fitness test' (expected speed c.1.8m/s). They were then asked to adjust their speed between trials if they were outside (±0.1m/s) of the expected pace.

Shod and barefoot trials were captured resulting in a total of 4 conditions: walk and march; and, shod and barefoot. Participants wore military issue training shoes (Silver Shadow, Hi-Tec™) for collection of shod trials over the force plates. Training shoes were chosen for testing over military boots primarily to allow direct marker placement on the ankle malleoli. Participants were discouraged from targeting the force plates. A recorded trial was deemed suitable if it had minimal marker dropout, full clean contact of the foot within the boundary of the force plates (minimum 5 clean strikes for each side) and no major gait inconsistency on the part of the subject as judged by an observer, e.g. stopping or stumbling.

The pelvis and thigh segments were defined according to Wu [25], the shank segments were defined according to Peters [26] and tracked using the marker cluster recommended by Manal [27], the foot segments were a modified version of the foot flat option defined according to Pratt [28]. The thorax was defined according to Gutierrez [29]. An additional foot segment was created for the calculation of joint moments based on a modified Helen Hayes set [30]. This segment is considered better suited for inverse dynamics calculations as it follows the dissection positions of Dempster [31]. It is defined with the proximal point at the ankle joint centre and removes the foot flat offset used in the kinematic foot. Internal moments were calculated for each lower limb joint.

Data processing and statistical analysis

Gaps smaller than 14 frames in the raw marker data were interpolated using a 3rd order least squares fit [32]. In the case of larger gaps the whole segment was excluded from analysis at these time points. The marker data was then filtered using a 6 Hz low pass bidirectional Butterworth filter [33]. Force plate data were filtered using a 50 Hz low pass Butterworth filter [34]. Gait data were normalised to body size as recommended by Hof [35] and Pierrynowski [36] (Table 1).

Kinematic and kinetic data were normalized to 100% of the gait cycle and stance phase respectively. This resulted in 101 individual time points for each movement plane where heel strike occurs at time points 0 and 100. Bootstrapped t-tests on each individual normalised time point were carried out to identify regions within the gait cycle that were significantly different [37]. Reference values for peak joint angles, and time to peak for sagittal lower limb angles are presented as supplementary data.

An attempt was made to control for speed a priori, however technical failure of light gates meant that this was not completely successful. Consequently, speeds were higher and more variable between participants than intended. Military training typically involves walking and marching at a fixed pace. ANCOVA was therefore used to control for speed in the temporal-spatial data. Multiple ANCOVAs were also carried out to cross-check that controlling for the variations in speed would not alter the interpretation of the original analysis of gait curves. Independent t-tests were carried out to compare height, body mass and speed. Alpha for all analyses was set to 0.05. SPSS (v18; SPSS Inc, USA) and Matlab (v2014a; MathWorks, USA) were used for all analyses.

Results

Cases ranged in age between 21-40 years (mean=27.5 years, sd=4.9 years); controls between 19-40 years (mean=28.3 years, sd=7.4 years). No pain was reported by cases or controls during testing demonstrating sufficient rest was provided between traverses. Cases (mean height 1.71m; sd 0.13) were significantly shorter (p=0.002) than controls (1.81m; 0.06) although there were no differences in weight or height-to-leg length ratio.

Kinematics

The mean (sd) speed was 1.8 (0.2) m/s for walking and 2.1 (0.2) m/s for marching. Cases were 0.08-0.14m/s faster than controls; although this difference was only significant for barefoot walking ($p=0.02$). There were no differences in normalised step time, stance time or swing time. A significantly longer stride length (relative to leg length) was observed for cases in the shod condition only (Table 2).

Toe-off occurred between 58-60% of the gait cycle for both the walking and marching conditions. The position of toe-off is therefore marked at 59% on all gait curves. Each kinematic and kinetic variable is presented graphically (Figure 1) highlighting regions of data that differ significantly ($p<0.05$) between the two groups.

Significantly greater ankle plantarflexion was measured in cases from mid-stance to toe-off with a maximum difference of 6.3° at 55% of the gait cycle (Tables 3 and 4). When the effect of speed was controlled for (using the ANCOVA) the difference observed was less (maximum difference 5.5° at 52% of the gait cycle) but the relationship to the gait cycle described above remained.

Significantly less ankle inversion was observed in cases at the end of stance and beginning of swing with a maximum difference of 3.8° at 64% of the gait cycle. This difference persisted, after statistical control for the effect of speed, for almost 10% of the gait cycle with a maximum difference of 4.9° at 63% of the gait cycle . A summary of the significant differences for kinematic data is presented in Table 4.

In view of the consistency of the results reported in Table 4, graphs of the original data (i.e. unadjusted for speed) are presented (Figure 1). Graphs for left-sided shod-marching are presented as there were no differences between left and right-sided data.

Forces and joint moments

When controlled for speed there were no consistent differences in any of the ground reaction forces. Consistent differences were found in the joint moments and are summarised in Table 4. Cases demonstrated lower ankle inversion moments during the majority of mid-stance; and greater ankle dorsiflexion moments during small sections of early stance and around the time of heel-off. Hip abductor moments were lower in cases during early mid-stance and during terminal stance. Representative graphs are presented in Figure 1.

Discussion

This study demonstrates a number of differences in biomechanical measurements between cases with CECS and asymptomatic controls. These differences were consistent during walking, marching, barefoot and shod gait. The shorter stature, with no differences in body proportions, seen in this cohort has not previously been discussed in a biomechanical context. This difference has not been demonstrated in a civilian population [38] and has only been reported in the military once [17]. The implications of the observed shorter stature are discussed throughout this section.

During the completion of this study many participants reported having previously experienced the urge to transition to run in order to alleviate their pain. The transition from walking to running (WRT) has been suggested to transfer the work from the dorsiflexor muscles to the larger proximal muscles (such as gluteus maximus, rectus femoris, vastus lateralis, and vastus medialis) [40]. The speed at which both humans and quadrupeds begin to transition to running also appears to be dependent on stature; resulting in a transition at the same Froude number (c.0.5; speed in relation to leg length). In humans, this corresponds to a WRT speed of around 2m/s (the marching speed in this study) depending on leg length [41]. The degree of tibialis anterior activation has been identified as a key determinant of the speed at which the WRT is triggered [39]. The short stature found for cases in this study may therefore have important implications on the requirements of tibialis anterior and the subsequent development of CECS.

The muscles of the anterior compartment in healthy individuals perform close to maximum capacity during fast walking [40]. This is amplified in shorter individuals during level walking [42] and is even less advantageous during ambulation on an incline as greater propulsion and toe clearance are required. The shorter stature found for cases in this study therefore likely demands increased activation of tibialis anterior, that may be represented here as an increased ankle dorsiflexor moment, and plays a significant role in the development of CECS. Further work to test this hypothesis is needed.

The relatively longer stride of shorter personnel, when normalised to leg length, may reflect ingrained changes induced by military training; whereby all personnel are required to move at a uniform cadence and speed. In order to maintain this relationship, shorter personnel can only achieve this through an increase of stride length relative to taller peers. This is likely to be the adaptation to allow ambulation at a higher Froude number. Cases increase speed through an increase in ankle plantarflexion at toe-off, resulting in an increased stride length, rather than a change in cadence. Experimental data also indicates that increasing stride length at a fixed speed increases the stress to the dorsiflexor musculature [43]. The greater ankle dorsiflexion moment observed in this study further supports this theory. This homogenisation of gait may be a key factor in the development of CECS in the military population.

The results of the current study suggest that the homogenisation of marching gait also has a similar effect on the slower walking gait. The over-striding in late stance, which is required by shorter personnel during marching, may then become ingrained into everyday walking gait in this population. It is likely that due to this learned adaptation cases have learned to override the normal stimulus to trigger the WRT.

Discomfort in the anterior compartment muscles has been reported in healthy individuals during fast walking [44] and pain has also been described when the WRT stimulus is overridden but subsides on transitioning to running or cessation of activity [45]. The continued excessive demand on the anterior compartment due to over-riding of the WRT may be the trigger for the development of CECS and/or symptoms of CECS in this population and could account for the higher reported prevalence of CECS in the military. The observation that kinematic differences of the lower limb only occur at the ankle joint and, that kinetic differences are present at the ankle joint also supports the suggestion of excessive demand on the anterior compartment. This is also in line with the symptomatology. The lower ankle inversion moments in cases from mid-stance to toe-off suggests that, in CECS, tibialis anterior is operating at a mechanical disadvantage. This may be due to an inability to generate the force required for inversion, due to intrinsic weakness [46,47] or neuromuscular fatigue. It could also occur as a result of failure to effectively transfer the force generated by tibialis anterior contraction due to tendon lengthening or stretch. This effect could also be produced by an external barrier, such as through compressive footwear, to the normal stretch of the extensor retinaculum at the ankle during tibialis anterior contraction [48]. Reduced force transmission through the tibialis anterior tendon would reasonably account for the differences observed in the ankle angles in this study.

The reduced hip abductor moments observed in this study have also been identified in other patient populations including those with iliotibial band syndrome and knee osteoarthritis [49,50]. This has been suggested to be as a result of weakness of the hip musculature. Of note, hip joint moments are the most susceptible to errors in the calculation of joint moments as these are propagated up the kinetic chain [51]. The reasons for these differences are therefore unclear and warrant further investigation to determine the role of the hip abductors, if any, in this condition.

This study provides evidence of biomechanical factors associated with CECS in males that are unlikely to be a protective mechanism; further investigation is required to confirm that these same factors apply to females. Furthermore the assessment of biomechanical differences in non-military populations is also required. Future work is also needed to investigate the activity of tibialis anterior in these populations. The technical issues experienced in the control of speed also could have affected the results. It was therefore reassuring that both the bootstrapped t-test and ANCOVA gave predominantly the same results. Finally, the case-control design of the current study identifies key aspects of gait specific to those with CECS. Acknowledging the limitations of this approach, the results of this study identify height, stride length and ankle biomechanics for potential inclusion in a prospective study. Nevertheless, this study is the first to describe biomechanical differences in this population, and the results provide new insights into this condition.

Perspectives

In summary, this study demonstrates differences between cases and controls, which are present prior to the onset of painful symptoms, in height and biomechanical measurements during ambulation. These data can not confirm whether the biomechanical differences are a cause or consequence of CECS. However, these data do provide potential mechanisms underpinning the development of CECS in this population. The changes in joint angles and moments may indicate an impairment of either the muscle or tendon of tibialis anterior. The shorter height necessitates an increased stride length that likely results in an increased demand on tibialis anterior during ambulation. This disadvantage may be further amplified when individuals override the urge to transition to a run or when ambulating on gradients. This may be a vital factor in the development of CECS in the military. Shorter personnel in military populations will continue to be required to march at prescribed speeds to fulfil occupational requirements; biomechanical interventions for CECS are therefore unlikely to be efficacious within this population.

Conflict of interest statement: The authors report no conflicts of interest. The authors alone are responsible for the content and writing of the paper.

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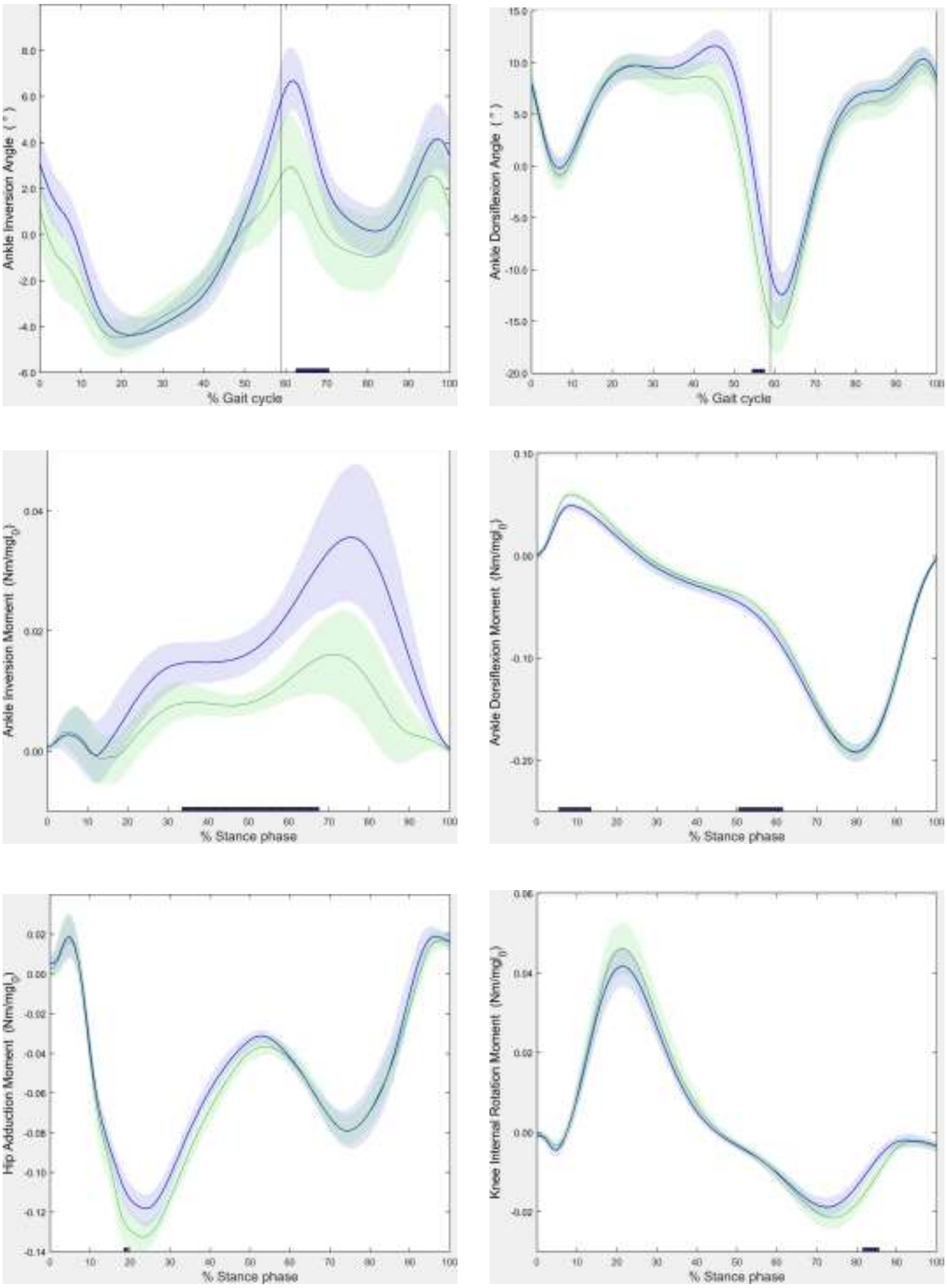

Figure 1. Differences in ankle angles (top row) during the gait cycle and ankle, knee and hip moments (2nd and 3rd rows) during stance phase from left shod-marching data. Blue lines represent CON group, green PT group. Shaded areas represent bootstrapped 95% confidence intervals. The bar along the x-axis indicates those time points where all conditions were significantly (P<0.05)

different. The vertical line on the angle sub figures indicates the time of toe off. The graphs show both dorsiflexion and inversion but the movements in these planes are predominantly negative indicating plantarflexion and eversion respectively.

Table 1 Normalisation of gait parameters. Symbols: l_0, leg length; m, body mass; g, acceleration due to gravity (9.81 m/s^2).

Quantity	Dimensionless number
Length, distance (l)	$\frac{l}{l_0}$
Time (t)	$\frac{t}{\sqrt{l_0/g}}$
Force (F)	$\frac{F}{mg}$
Moment (M)	$\frac{M}{mgl_0}$

Table 2. Comparison of differences in temporal-spatial data between groups (*P<0.05). Differences for the left-side only are presented as there were no differences between left and right-sided data. N.B. All variables are normalised according to Hof and therefore do not have units.

Hof-norm'd variable	BF/ SHOD	Condition	F	P	Mean (CON)	SE (CON)	Mean (PT)	SE (PT)	Mean Diff
Stride Length	BF	Walk	1.6	0.22	1.76	0.02	1.80	0.022	0.042
		March	2.7	0.11	1.90	0.03	1.96	0.026	0.063
	SHOD	Walk	6.6	0.014*	1.79	0.02	1.86	0.018	0.068
		March	4.3	0.046*	2.00	0.03	2.08	0.026	0.076
Step time	BF	Walk	0.06	0.80	1.52	0.01	1.51	0.013	0.005
		March	0.01	0.91	1.40	0.02	1.39	0.015	0.002
	SHOD	Walk	0.03	0.88	1.61	0.01	1.61	0.011	0.003
		March	0.01	0.93	1.46	0.02	1.46	0.015	0.002
Stance time	BF	Walk	1.42	0.24	1.78	0.02	1.75	0.015	0.027
		March	3.02	0.09	1.60	0.02	1.56	0.015	0.04
	SHOD	Walk	0.34	0.56	1.96	0.02	1.95	0.015	0.013

		March	0.48	0.49	1.72	0.02	1.71	0.016	0.016
Swing time	BF	Walk	0.23	0.64	1.26	0.02	1.27	0.016	0.011
		March	0.72	0.40	1.20	0.02	1.22	0.020	0.024
	SHOD	Walk	1.69	0.20	1.25	0.02	1.28	0.015	0.028
		March	0.54	0.47	1.20	0.02	1.21	0.019	0.020

Table 3. Angular measurements from bootstrapped t-test data (Left shod-marching) of maximal significant (P<0.05) differences between CON and PT. Angles are reported in degrees; moments are reported normalised to body mass and leg length (Nm/kg.LL)

Joint	Movement (% of gait cycle)	Angle / moment	Mean (sd) CON	Mean (sd) PT	Difference
Ankle	Inversion (64%)	Angle	6.5 (3.1)	2.7 (4.8)	-3.8
Ankle	Plantarflexion (55%)	Angle	1.6 (6.8)	-4.7 (5.9)	+6.3
Ankle	Inversion (83%)	Moment	0.034 (0.026)	0.012 (0.004)	0.022
Ankle	Dorsiflexion (81%)	Moment	0.047 (0.009)	0.058 (0.008)	-0.011
Hip	Abduction (19%)	Moment	-0.099 (0.016)	-0.115 (0.024)	0.016
Knee	Internal rotation (83%)	Moment	-0.012 (0.005)	-0.017 (0.005)	0.005

Table 4. Time points of significant differences (all conditions) in angular and moment data.

Joint	Movement	Angle/Moment	Bootstrapped t-test (%)	ANCOVA (%)	Direction of effect
Ankle	Inversion	Angle	63-70	62-69	CON>PT
Ankle	Plantarflexion	Angle	55-57	-	PT>CON
Ankle	Inversion	Moment	34-67	37-65	CON>PT
Ankle	Dorsiflexion	Moment	6-13, 51-61	9-10, 59-61	PT>CON
Hip	Abduction	Moment	19	31-39, 92-94	CON>PT
Knee	Internal rotation	Moment	82-85	-	CON>PT