Gait & Posture 108 (2024) 208–214

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Acute biomechanical responses to wearing a controlled ankle motion (CAM) Walker boot during walking

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ARTICLE INFO

Keywords:
Gait
Rehabilitation
Leg length discrepancy
Achilles tendon
Joint immobilisation
Orthotic walker

ABSTRACT

Background: Controlled ankle motion (CAM) boots are often prescribed during the rehabilitation of lower limb injuries and pathologies to reduce foot and ankle movement and loading whilst allowing the patient to maintain normal daily function.
Research question: The aim of this study was to quantify the compensatory biomechanical mechanisms undergone by the ipsilateral hip and knee joints during walking. In addition, the compensatory mechanisms displayed by the contralateral limb were also considered.
Methods: Twelve healthy participants walked on an instrumented treadmill at their preferred walking speed. They underwent kinematic and kinetic analysis during four footwear conditions: normal shoes (NORM), a Malleo Immobil Air Walker on the right leg (OTTO), a Rebound® Air Walker on the right leg with (EVEN) and without (OSS) an Evenup Shoelift™ on the contralateral leg.
Results: CAM boot wear increased the relative joint contribution to total mechanical work from the ipsilateral hip and knee joints ($p < 0.05$), which was characterised by increased hip and knee abduction during the swing phase of the gait cycle. EVEN increased the absolute work done and relative contribution of the contralateral limb. CAM boot wear reduced walking speed ($p < 0.05$), which was partially compensated for during EVEN.
Significance: The increased hip abduction in the ipsilateral leg was likely caused by the increase in effective leg length and limb mass, which could lead to secondary site complications following prolonged CAM boot wear. Although prescribing an even-up walker partially mitigates these compensatory mechanisms, adverse effects to contralateral limb kinematics and kinetics (e.g., elevated knee joint work) should be considered.

1. Introduction

Controlled ankle motion (CAM) boots are often prescribed for post-surgical immobilisation following traumatic injuries to the foot and ankle such as Achilles tendon rupture, and for wound offloading for conditions such as diabetic foot ulceration [1–4]. CAM boots allow patients to continue ambulation and other activities of daily living while unloading and protecting the foot and ankle complex [2]. As such, they offer functional benefits over a plaster cast (e.g., allowing controlled joint movement), in addition to other benefits (e.g., removability making them more hygienic). However, the continuation of normal, unperturbed gait is challenging while wearing a CAM boot for various reasons. They are designed to reduce ankle joint range of motion (ROM) during walking, although there is inconsistency in the literature around their ability to do this effectively [1,5]. Nonetheless, assuming ROM is reduced, this would also negate the ankle's potential to produce

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https://doi.org/10.1016/j.gaitpost.2023.12.011
Received 3 July 2023; Received in revised form 9 December 2023; Accepted 14 December 2023
Available online 17 December 2023

effective joint moments, which have been shown to be major contributors to overall mechanical work during walking at various speeds [6,7].

The second alteration that might affect gait biomechanics is the artificial increase in leg length caused by the CAM boot's sole [2]. This discrepancy varies depending on the thickness of the sole on the contralateral side, which has led to commercially available appliances such as "even-up walkers", designed to equalise leg length. Leg length asymmetry via CAM boots or other means, has been shown to compromise balance [8], alter kinematics [9,10], and increase the metabolic cost of locomotion [11], the risk of stress fractures, and the incidence of lower back pain [12]. The third alteration is that CAM boots increase the mass of the limb, thus increasing the mechanical demand of the muscles responsible for lifting the limb during early- and mid-swing. There is little research that has attempted to quantify the kinetic demand placed on these muscles, although given the other two alterations discussed (reduced ankle ROM and increased leg length) would only amplify this demand, changes are likely to be clinically meaningful. This has somewhat been considered in prosthetics [13] and CAM boot-specific studies [10], but this remains an under-appreciated consideration in most gait research.

The reduced capacity for the ankle to perform work, along with the increased leg length and mass, must be compensated for by neighbouring joints if gait patterns are to be maintained [10]. This might be in the form of an increased joint contribution to total mechanical work done during gait, or by accentuated joint kinematics (e.g., hip abduction) to mitigate the effects of CAM boot wear. This has been shown previously through altered ipsilateral hip and knee joint moments compared with normal footwear [10,14]. Additionally, no information is available for the kinematics and kinetics of the contralateral (unaffected) leg during walking with an even-up walker during CAM boot wear. The alterations caused by the CAM boot likely affect contralateral joint biomechanics as much as those of the ipsilateral joints, potentially by increasing their total mechanical work done. Investigations into the effectiveness of even-up walkers are warranted here, as CAM boots have been shown to induce asymmetries in spatiotemporal variables such as step length and step width [15], and are reported to lead to long-term contralateral hip pain, alongside lower-back and ipsilateral knee pain [2]. Therefore, the aim of the current study was to investigate the acute biomechanical responses to wearing CAM boots during walking at a preferred walking speed in both the ipsilateral and contralateral legs, accounting for reduced ankle motion, increased leg length, and increased segment mass. We considered the effect of wearing an even-up walker on the contralateral limb, which might mitigate any kinematic or kinetic response to walking during CAM boot wear.

2. Methods

2.1. Participants

Twelve participants (eight males, four females; age: [mean ± S.D.] 29 ± 8 y; stature: 1.81 ± 0.86 m; body mass: 81.6 ± 13.7 kg) were recruited for this study. At the time of data collection, participants were healthy and free of any musculoskeletal injury or neurological condition that might impact gait. Prior to participation, participants completed health screening and provided written informed consent. The study was approved by the Local Research Ethics Committee (project number 66465), and was conducted in accordance with the Declaration of Helsinki [16].

2.2. Data collection

Participants were required to walk during four footwear conditions: (1) wearing a "hard-shell" CAM boot (Rebound® Air Walker, Össur, Iceland) on the right leg (OSS); (2) wearing a "soft-shell" CAM boot (Malleo Immobil Air Walker, Otto Bock, Germany) on the right leg (OTTO); (3) wearing the OSS CAM boot with a 28-cm Evenup Shoelift™ (Össur, Iceland) on the contralateral leg (EVEN); and (4) wearing their own trainers bilaterally as a normal footwear condition (NORM). Participants wore their own trainers on the left leg during all CAM boot conditions. The CAM boot was always worn on the right leg to quasi-randomise for limb dominance, and the size of the boots (small, medium, or large) were determined by the participant's shoe size.

During each condition, participants walked at their preferred walking speed (PWS). PWS was determined using the final eight out of ten overground trials, with the first two trials being used as familiarisation, measured using photocell timing gates (WittyGATE, Microgate, Italy). The overground trials were conducted around a ~30-m loop, where a central 2-m linear portion was used to obtain PWS. All experimental walking trials were then conducted on a motorised treadmill (Gaitway3D, h/p/cosmos, Germany) instrumented with four load cells to measure three-dimensional ground reaction forces at 2 kHz (Arsalis, Belgium). Motion capture was conducted using a 14-camera optoelectronic system (Oqus 7 +, Qualisys AB, Sweden) operating at 250 Hz. Kinetic and kinematic systems were synchronised via digital integration of force signals into the motion capture software, as per manufacturer recommendations. Using an adapted 'CAST' (Calibrated Anatomical Systems Technique) methodology [17], 24 retroreflective markers were placed on key anatomical locations (bilaterally the anterior and posterior superior iliac spines, iliac crest, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, posterior aspect of the calcaneus, head of the 1st and 5th metatarsals, and base of the 2nd metatarsal), with non-collinear four-marker rigid clusters placed on the thigh and shank segment. Marker placement on the CAM boots was done before the boots were 'closed', meaning key anatomical landmarks could be palpated. The calcaneus marker was placed using the height of the metatarsal markers and palpation as a guide [10]. All markers were left on the participants during static (segment definition) and dynamic (walking) trials. Participants had two minutes of familiarisation at PWS in each condition, before five gait cycles (between consecutive ipsilateral ground contacts) were collected for analysis. The order of gait condition was randomised for each participant.

2.3. Data processing

Instrumented treadmill data (ground reaction forces and centres of pressure) were filtered using a recursive second-order (zero phase-lag), low-pass Butterworth filter with a cut-off frequency of 17.7 ± 3.6 Hz. Cut-off frequencies were determined with residual analyses [18] using a custom-written Matlab script (R2023a, MathWorks Inc., USA). Filtered kinetic data then underwent signal decomposition [19,20] before being exported to Visual3D (v6.01.36, C-Motion Inc., Canada) along with motion capture data. Kinematics were filtered using a recursive second-order (zero phase-lag), low-pass Butterworth filter with cut-off frequency of 6.0 Hz. Kinematic modelling was conducted using six degrees-of-freedom, where the CAM boot was modelled as standard foot and shank segments. The mass of the boot (*small*: OSS = 0.97 kg, OTTO = 1.06 kg; *medium*: OSS = 1.11 kg, OTTO = 1.22 kg; *large*: OSS = 1.35 kg, 1.27 kg) was added to the foot segment during modelling. Joint kinetics (joint moments and mechanical work) were estimated using Inverse Dynamics. The sum of ankle, knee, and hip joint mechanical work was used to estimate total mechanical work done by ipsilateral and contralateral limbs individually. The relative contribution of each joint to total mechanical work was therefore calculated by normalising joint mechanical work to total mechanical work. This relative contribution was used to account for any differences in gait speed between conditions. This method maintained ecological validity by allowing participants to walk at their self-selected gait speed. Spatiotemporal variables were also computed in Visual3D for left and right legs.

2.4. Statistical analysis

All statistical analyses were conducted using SPSS (version 27, IBM, USA) and Matlab. PWS was compared between gait conditions using a one-way repeated measures analysis of variances (ANOVA) with Bonferroni adjustment for post-hoc tests. Data were tested for sphericity, and Greenhouse-Geiser corrections were applied where required. Discrete spatiotemporal and kinetic variables were also analysed using one-way ANOVA for left and right legs independently, meaning inter-limb differences were not observed in the current study. Time-series joint kinematics and kinetics, normalised to a percentage of the gait cycle (kinematic) or stance phase (kinetic), were analysed for effects of gait condition with statistical parametric mapping (SPM) using the open-source SPM1D Matlab package (version M.0.4.10) for one-way repeated measures ANOVA (‘anova1rm’) [21]. No post-hoc tests were conducted using SPM as these are not yet validated. As such, any observations between individual conditions were descriptive only. For all tests, significance level was set at $p < 0.05$.

3. Results

There was a main effect of gait condition on PWS ($F_{3,36} = 8.65$, $p < 0.001$). Compared to NORM, PWS was reduced in OSS and OTTO, but not EVEN (Table 1). Post-hoc testing showed no other significant differences for PWS between gait conditions. There were no effects of condition on spatiotemporal characteristics in either limb (Table 1), except cadence ($F_{3,36} = 4.88$, $p = 0.020$) and swing time ($F_{3,36} = 6.74$, $p = 0.012$) in the right leg. Post-hoc testing showed no differences between individual conditions for cadence ($p \geq 0.073$). However, swing time was significantly lower for OSS and OTTO, but not EVEN, when compared to NORM (Table 1).

Table 1
Preferred walking speed (PWS), spatiotemporal data, and joint-level mechanical work done averaged over the stance phase, for left and right legs during the four experimental conditions.

Parameter	Limb	NORM	OSS	OTTO	EVEN
PWS (m/s)	***N/A***	1.37 ± 0.16	**1.26 ± 0.19** *	**1.19 ± 0.19** * *	1.28 ± 0.09
Step length (m)	***Left***	0.73 ± 0.09	0.71 ± 0.08	0.68 ± 0.07	0.72 ± 0.06
	Right	0.73 ± 0.08	0.72 ± 0.07	0.70 ± 0.09	0.71 ± 0.06
Stance time (ms)	***Left***	692 ± 104	730 ± 89	734 ± 64	741 ± 73
	Right	682 ± 97	708 ± 81	703 ± 67	709 ± 86
Swing time (ms)	***Left***	420 ± 37	428 ± 31	424 ± 36	418 ± 33
	Right	430 ± 43	**450 ± 37** * *	**457 ± 41** *	453 ± 34
Cycle time (ms)	***Left***	1111 ± 133	1158 ± 110	1158 ± 97	1159 ± 98
	Right	1111 ± 133	1159 ± 110	1160 ± 98	1162 ± 99
Cadence (Hz)	***Left***	1.86 ± 0.21	1.76 ± 0.14	1.78 ± 0.15	1.78 ± 0.15
	Right	1.83 ± 0.23	1.73 ± 0.17	1.71 ± 0.14	1.70 ± 0.12
Ankle work (J/kg)	***Left***	0.43 ± 0.05	0.40 ± 0.06	0.41 ± 0.08	0.39 ± 0.07
	Right	0.38 ± 0.13	**0.06 ± 0.02** * *	**0.07 ± 0.03** * **	**0.08 ± 0.08** * **
Knee work (J/kg)	***Left***	0.46 ± 0.16	0.40 ± 0.15	0.39 ± 0.17	**0.48 ± 0.09**$^{\#}$
	Right	0.49 ± 0.13	0.45 ± 0.12	0.39 ± 0.11	0.42 ± 0.10
Hip work (J/kg)	***Left***	0.48 ± 0.11	0.47 ± 0.11	0.46 ± 0.12	0.49 ± 0.07
	Right	0.49 ± 0.09	0.45 ± 0.12	0.39 ± 0.08	0.43 ± 0.07
Total work (J/kg)	***Left***	1.37 ± 0.28	1.27 ± 0.27	1.25 ± 0.34	1.36 ± 0.18
	Right	1.37 ± 0.25	**0.95 ± 0.22** * *	**0.85 ± 0.19** * **	**0.93 ± 0.19** * **

* Significantly different from NORM at p < 0.05 level.
* *Significantly different from NORM at p < 0.01 level.
* **Significantly different from NORM at p < 0.001 level. $^{\#}$Significantly different from OSS at p < 0.05 level. $^{\#\#}$Significantly different from OSS at p < 0.01 level.

There was an overall main effect of condition on total mechanical work done by the right leg ($F_{3,36} = 20.35$, $p < 0.001$), which was lower in all CAM boot conditions when compared to NORM (Table 1). Ankle work in the right leg was also affected by gait condition ($F_{3,36} = 39.11$, $p < 0.001$), with all CAM boot conditions (OSS, OTTO, and EVEN) being lower than NORM (Table 1). Overall main effects were also found at the knee and hip joints, but post-hoc tests showed only a difference between OSS and EVEN in the left knee ($p = 0.048$).

When displayed as a proportion of total mechanical work, the work done by the right ankle joint was reduced ($F_{3,36} = 35.89$, $p < 0.001$) from 27.6 ± 8.9% in NORM to 5.7 ± 1.1%, 8.5 ± 3.3%, and 7.6 ± 5.3% in OSS, OTTO, and EVEN, respectively (Fig. 1B). Conversely, both the right knee ($F_{3,36} = 14.37$, $p < 0.001$) and hip ($F_{3,36} = 17.20$, $p < 0.001$) joint showed significant increases in relative joint contribution in OSS (knee: 46.9 ± 7.4%; hip: 47.4 ± 3.8%), OTTO (knee: 44.9 ± 5.7%; hip: 46.7 ± 6.0%), and EVEN (knee: 45.3 ± 5.9%; hip: 47.0 ± 8.2%), compared to NORM (knee: 36.2 ± 7.1%; hip: 36.2 ± 4.6%) (Fig. 1B). In the left ankle ($F_{3,36} = 5.48$, $p = 0.010$), joint contribution was lower in EVEN compared to OTTO ($p = 0.036$) (Fig. 1A). In the left knee ($F_{3,36} = 9.33$, $p < 0.001$), joint contribution was higher in EVEN compared to OSS ($p = 0.013$) and OTTO ($p = 0.008$), but not NORM ($p = 0.504$) (Fig. 1A). No main effects were found for left hip joint (Fig. 1A).

Fig. 2 and Fig. 3 show time-series joint kinematic data for the knee and hip joints, respectively. Suprathreshold clusters were detected for left knee sagittal-plane (Fig. 2A) and left hip sagittal- and frontal-plane angles, with OSS and OTTO conditions showing less knee and hip flexion, with less hip abduction (Fig. 3A,B). EVEN appeared to show more hip abduction around late-swing compared to the other conditions (Fig. 3B). In the right leg, suprathreshold clusters were mainly detected for frontal- and transverse-plane knee kinematics, with OSS, OTTO, and EVEN displaying more knee abduction and internal rotation (Fig. 2E,F) as well as more hip abduction during swing, although this was reduced somewhat in EVEN (Fig. 3E).

Fig. 4 and Fig. 5 show time-series joint moment data for the knee and hip joints, respectively. Suprathreshold clusters were mainly detected for left knee sagittal- and transverse plane (Fig. 4A,C) and left hip frontal-plane moments (Fig. 5B), with EVEN causing increased knee extension moments, and OSS and OTTO increasing knee external rotation moments. In the right leg, suprathreshold clusters were detected for sagittal- and frontal-plane knee joint moments, with OSS, OTTO, and EVEN showing reduced knee extension and adduction moments (Fig. 4E, F) as well as some reduced hip adduction moments (Fig. 5E).

4. Discussion

The aim of this study was to describe acute biomechanical changes in gait during CAM boot wear with and without the implementation of a contralateral even-up walker. The reduction in the ankle joint’s contribution to total mechanical work during CAM boot wear led to increases in the relative contribution of the ipsilateral hip and knee joint to total mechanical work in both “hard-shell” and “soft-shell” boots. The even-up walker affected knee and ankle joint contributions on the contralateral leg but did partially mitigate some biomechanical responses on the ipsilateral leg. Despite this, few significant spatiotemporal changes, besides PWS, were reported.

The alterations in relative joint contribution to total mechanical work in the ipsilateral limb (Fig. 1B) were characterised by increased hip abduction during most of the gait cycle and increased knee abduction in the early and swing phase of the gait cycle. Reductions in knee adduction and extension moments were observed in the CAM boot conditions, along with reduced hip adduction moments during stance, when compared to NORM. These changes during the different CAM boot

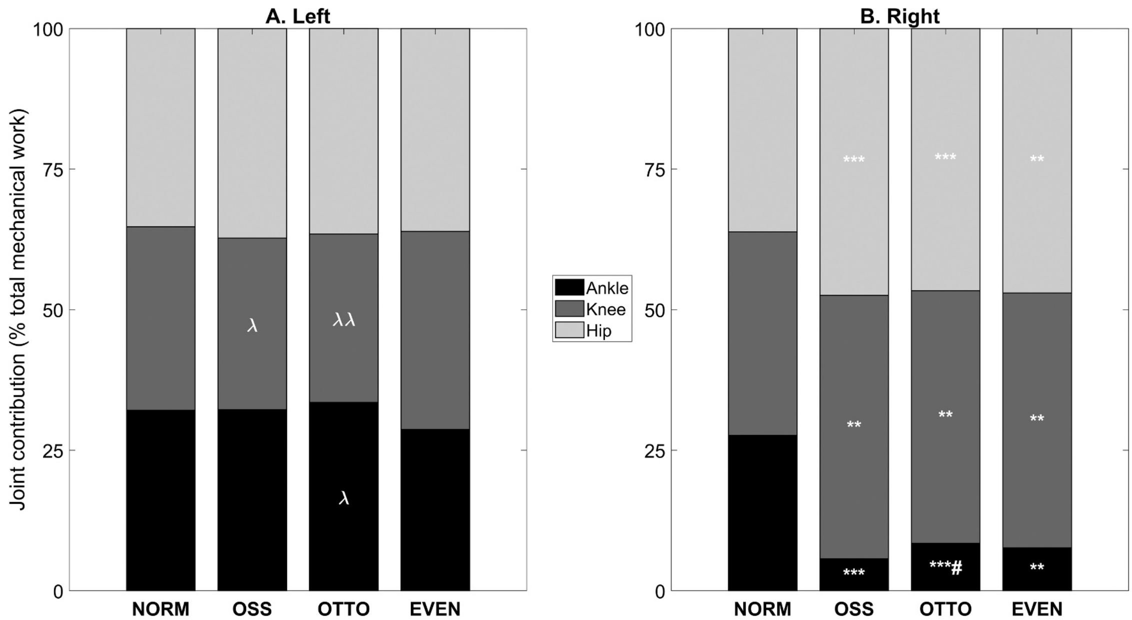

Fig. 1. Stacked bar of percentage work contribution for (A) the left leg, and (B) the right leg. * * and * ** denote conditions are significantly different from NORM at $p < 0.01$ and $p < 0.001$ level, respectively. # denotes condition is significantly different from OSS at $p < 0.05$ level. λ and λλ denote conditions are significantly different from EVEN at $p < 0.05$ and $p < 0.01$ level, respectively.

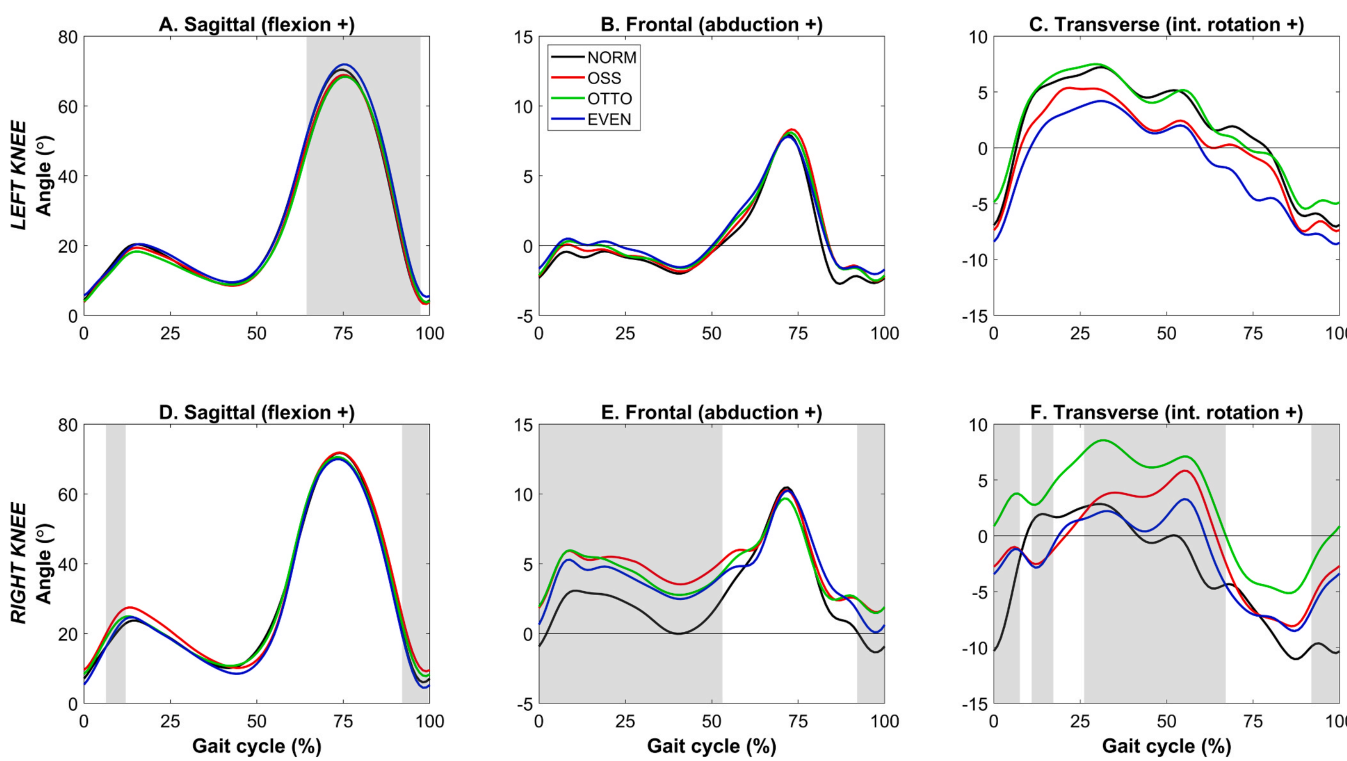

Fig. 2. Knee joint angle data for left (top row, A-C) and right (bottom row, D-F) legs during the four experimental conditions. Curves are normalised as a percentage of gait cycle. Grey, shaded areas represent suprathreshold clusters as determined by SPM analysis, which indicate an overall main effect of condition ($p < 0.05$), although no post-hoc comparisons were made.

conditions suggest the hip and knee produce a compensatory mechanism. These findings agree with previous literature that observed kinematic and kinetic alterations during CAM boot wear [10,14]. The compensatory mechanisms might explain the secondary site pain reported in the ipsilateral knee, which is among the most commonly-reported sites of pain following CAM boot wear [2]. The increased hip abduction shown here is likely a compensatory mechanism to the increased effective leg length, as well as the extra mass placed on the foot-ankle complex on the ipsilateral leg. Although the ipsilateral hip is not commonly reported as a site of pain following prolonged CAM boot wear, this increased joint contribution with altered kinematics might have a long-term detrimental impact on hip joint health. This should be considered when prescribing CAM boots to populations who might inherently be at an elevated risk of hip pathologies, such as those

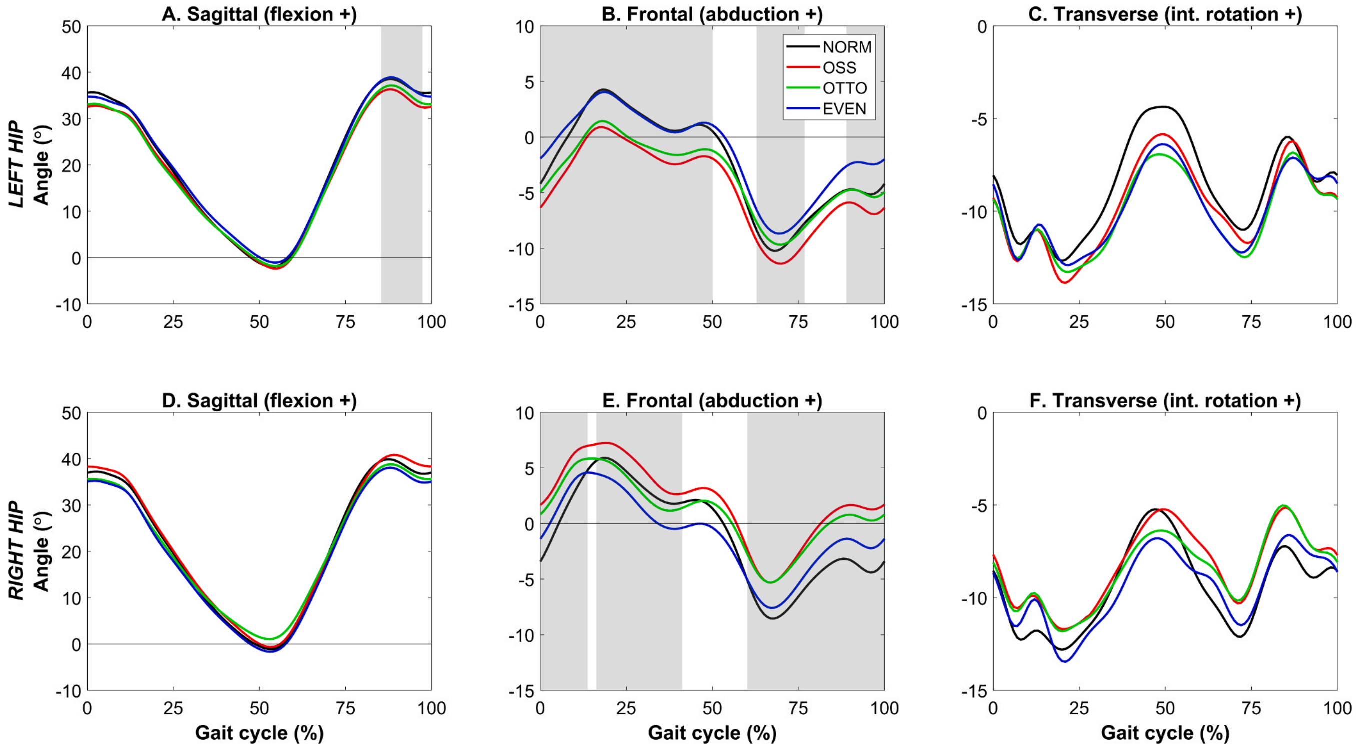

Fig. 3. Hip joint angle data for left (top row, A-C) and right (bottom row, D-F) legs during the four experimental conditions. Curves are normalised as a percentage of gait cycle. Grey, shaded areas represent suprathreshold clusters as determined by SPM analysis, which indicate an overall main effect of condition ($p < 0.05$), although no post-hoc comparisons were made.

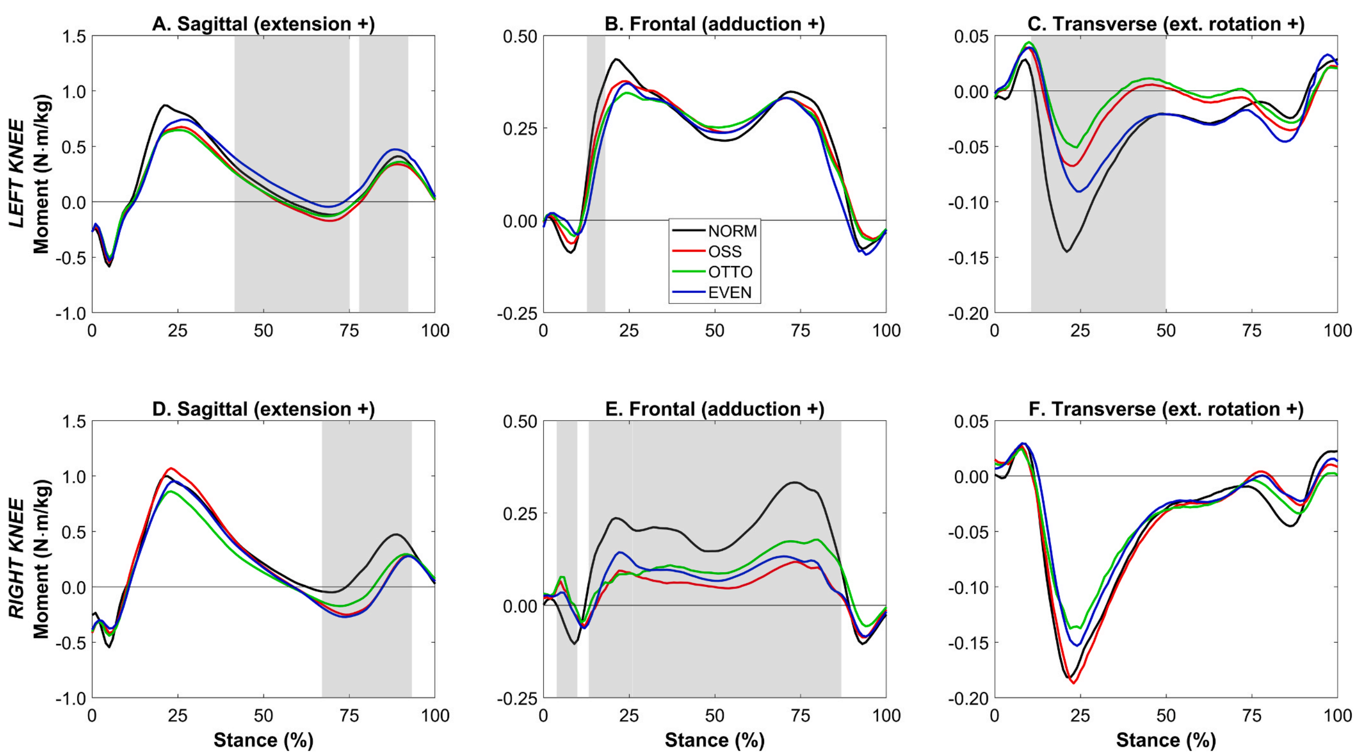

Fig. 4. Knee joint moments for left (top row, A-C) and right (bottom row, D-F) legs during the four experimental conditions. Curves are normalised as a percentage of the stance phase. Grey, shaded areas represent suprathreshold clusters as determined by SPM analysis, which indicate an overall main effect of condition ($p < 0.05$), although no post-hoc comparisons were made.

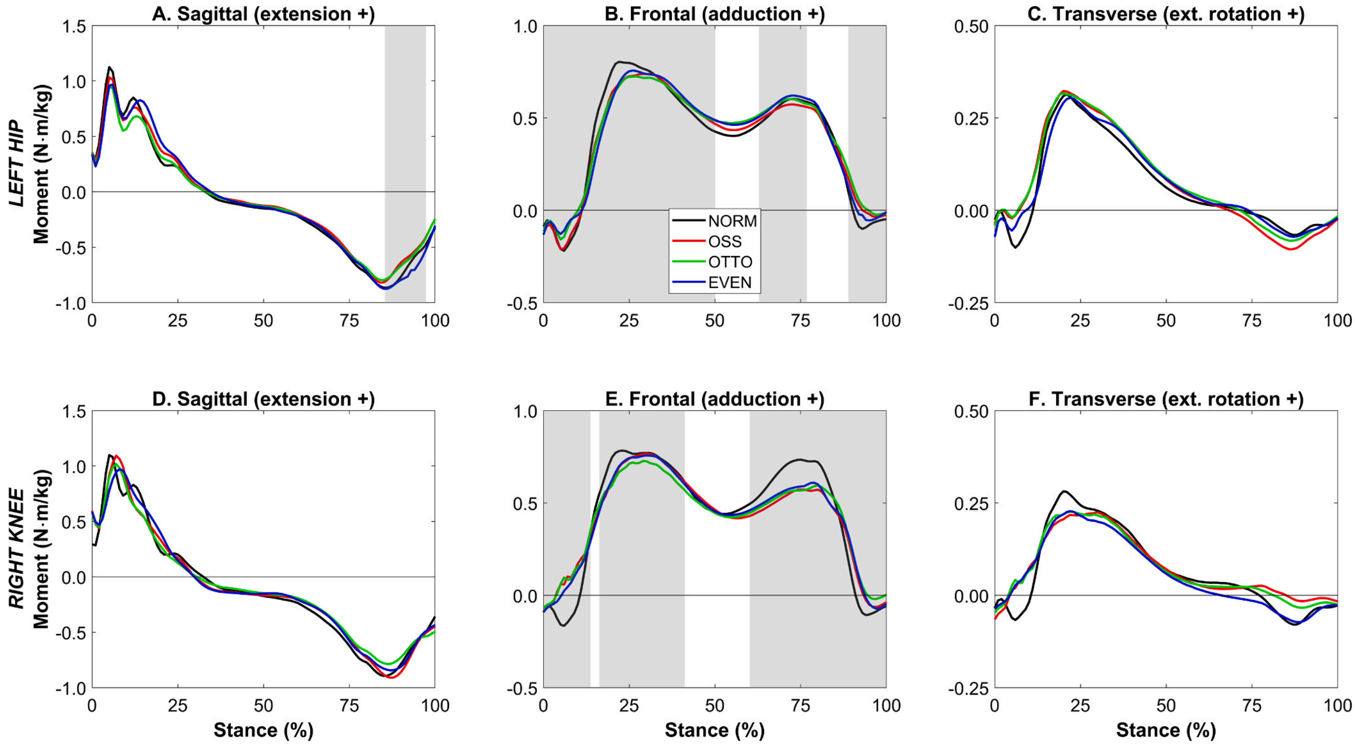

Fig. 5. Hip joint moments for left (top row, A-C) and right (bottom row, D-F) legs during the four experimental conditions. Curves are normalised as a percentage of the stance phase. Grey, shaded areas represent suprathreshold clusters as determined by SPM analysis, which indicate an overall main effect of condition ($p < 0.05$), although no post-hoc comparisons were made.

with high body mass [22]. The implementation of the even-up walker reduced hip abduction angles during swing in the ipsilateral leg (Fig. 3E), demonstrating the effectiveness of prescribing this alongside CAM boots to mitigate the impacts seen in the other gait conditions (OSS and OTTO).

The even-up walker also partially mitigated impacts on the contralateral leg, specifically the altered hip abduction angles during stance in OSS and OTTO (Fig. 3B). Contralateral hip and knee are also reported as sources of secondary-site pain following CAM boot wear [2]; the even-up walker might reduce secondary site complications here by reducing the compensatory mechanisms adopted by patients wearing CAM boots, while also partially maintaining PWS (Table 1). However, EVEN also showed an increase in absolute work done and relative contribution of the left knee and reduced the ankle's contribution to total mechanical work (Fig. 1A). This means that, even though the even-up walker offers some mitigation to the ipsilateral leg because of the reduced leg length discrepancy, it does not allow the contralateral leg to maintain normal biomechanical function. One consideration here is that we did not control for gait speed as such. We opted to compare gait parameters between conditions at PWS as this is a true effect of CAM boot wear, however, this affects the absolute joint work requirements (i.e., lower PWS = lower total mechanical work done). Future research on compensatory mechanisms could account for this in work calculations (e.g., normalise to gait speed) or control gait speed in data collection, but this was beyond the purpose of this study. Nonetheless, we feel that normalising joint work done to total lower limb mechanical work (Fig. 1) accounts for differences in gait speed, meaning comparisons between conditions remain valid.

The findings of this study might be impacted by several limitations. It should be noted that the compensatory mechanisms described above are acute effects of CAM boot wear, which do not provide any indication of whether these effects would decrease (i.e., patient habituation) or increase (i.e., lead to further increases in secondary-site injury risk) following extended periods of use. As such, future research should seek to understand the habituation effects of prolonged CAM boot wear, both with and without the implementation of an even-up walker. Additionally, the participants in this study were healthy, with no current lower-limb musculoskeletal injury or disorder. This allowed us to isolate the effects of the gait condition without needing to consider the biomechanical impact of the injury itself. However, it does affect ecological validity, as patients who have been prescribed a CAM boot due to a specific injury or condition might respond differently to healthy participants. It is entirely possible that secondary-site complications are not caused by biomechanical compensations at all but are simply a result of reduced physical activity as a comorbidity following injury or pathology, so this should be explored further by researchers. Our decision against controlling gait speed can also be considered a limitation of the current study. Although we aimed to understand the biomechanical implications of CAM boot wear, one of which was a reduced PWS, this does somewhat limit direct comparisons between conditions in terms of joint-level kinematics and kinetics, as gait speed might alter these.

The current study also offered an insight into the generic response to CAM boot wear without considering the specific modifications often applied to CAM boots for specific injuries or pathologies. For example, their prescription for Achilles tendon ruptures is supplemented with heel wedges, or specific CAM boots are used to control the ankle's ROM and incrementally load the Achilles tendon during healing. So, although these findings lack specificity to a particular injury, they provide novel insight into how the responses at neighbouring joints may occur. Finally, as we modelled the CAM boot as a six degrees-of-freedom foot and shank, we have no information regarding in-boot kinematics or kinetics. As the aim of the study was to understand compensatory mechanisms, we feel that this was not a major limitation given that forces within the boot would be unlikely to result in external forces that would substantially impact our analyses. Nonetheless, previous research [5] showed CAM boots do not fully reduce ankle ROM, although significant modifications to the boot (e.g., holes cut around key anatomical landmarks) were made using this marker-based motion capture method. Therefore, the structural integrity of the boot might have been compromised, as another study, using video fluoroscopy, showed significant reductions in talocrural joint ROM [1]. As such, more advanced kinematic and kinetic methods [23] are required to further understand the ankle's true contribution to total mechanical work, even when inside a CAM boot. This will have implications for research in this area and in clinical practice.

5. Conclusion

CAM boot wear increases the relative contribution of the ipsilateral hip and knee joint to total mechanical work during walking at PWS, which is a compensatory mechanism for the reduced work capacity of the ankle joint. However, the reduced PWS during CAM boot wear did results in reduced overall lower-limb mechanical work done. Whilst the kinematic effects of CAM boot wear are partially mitigated by the implementation of an even-up walker on the contralateral foot to reduce leg length discrepancies, this can adversely affect the contralateral limb's biomechanical function. In summary, increased hip abduction during swing on the ipsilateral leg, likely caused by the increased effective leg length and mass of the foot, could lead to secondary site injuries following chronic exposure to daily function when wearing a CAM boot.

Funding

This research was supported by the Carnegie School of Sport Staff Research Development Fund.

CRediT authorship contribution statement

Stolycia Mason L.: Writing – review & editing, Methodology, Investigation, Data curation. **Wilkins Richard A.:** Writing – review & editing, Supervision, Investigation, Conceptualization. **Lunn David E.:** Writing – review & editing, Writing – original draft, Validation, Supervision, Software, Resources, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Walker Josh:** Writing – review & editing, Writing – original draft, Visualization, Validation, Software, Resources, Project administration, Methodology, Investigation, Funding acquisition, Formal analysis, Data curation, Conceptualization. **Thomas Aaron:** Writing – review & editing, Software, Resources, Methodology, Investigation, Formal analysis, Data curation, Conceptualization.

Declaration of Competing Interest

The named authors have no conflict of interest to disclose, financial or otherwise.

Acknowledgements

The authors would like to acknowledge Samuel Dickinson for their contributions towards data collection.

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