Joint moment strategies during stair descent in patients with peripheral arterial disease and intermittent claudication

Stephanie L.King, NatalieVanicek, Thomas D. O'Brien

Abstract

Objective: To determine the lower limb joint kinetic strategies during stair descent in claudicants with peripheral arterial disease (PAD-IC). **Design:** Cross-sectional observation study. **Setting:** University laboratory. **Participants:** A total of 22 participants; 10 healthy controls and 12 patients diagnosed with PAD-IC. **Main Outcome Measures:** Between-group comparisons of ground reaction force (GRF) and, hip, knee and ankle kinetics during steady-state stair descent. **Results:** The claudicating-limb group demonstrated reduced vertical and posterior GRF compared to healthy controls (ES=-1.46 [-2.32,-0.69] and ES=-1.08 [-0.42,-0.26] as well as demonstrating a greater contribution to support moment from the ankle and trends towards a smaller hip contribution (42±14% vs 28±7%, P=.005 and Hip 16±8% vs 21±11%, P=.056, respectively). A unique sub-group was identified within the PAD-IC cohort demonstrating different hip moment strategies during weight acceptance: a novel hip extensor group (PAD-IC HExt) and stereotypical hip flexor group (PAD-IC HFlex). Compared to both healthy controls and the PAD-IC HFlex groups, the PAD-IC HExt group demonstrated increased hip extensor moment (ES=3.05 [1.67,4.42] and ES=3.62 [1.89,5.35]) and reduced knee extensor moment (ES=-2.00 [-3.15,-0.85] and ES=-1.36 [-2.60,-0.11] respectively) during weight acceptance. **Conclusions:** A novel hip extensor strategy was identified in a sub-group of claudicants which acts to reduce the demand on the knee extensors, but not the plantarflexors. Weakness in the knee extensors may prevent redistribution of the task demand,

typically seen in older adults in stair descent, away from the functionally limited plantarflexor muscle group. Further investigation into multi-level joint strength and the relationship to functional tasks is warranted to inform targeted intervention programmes.

Keywords: Claudicant, PAD-IC, hip moment strategy, stair descent, joint kinetics

Introduction

Peripheral arterial disease (PAD) refers to a chronic disease of the peripheral arteries, primarily in the legs, that negatively impacts functional ability[1], physical activity levels[2] and quality of life[3]. The calf is a frequently reported site of intermittent claudication pain (IC)[4] and strength of the plantarflexors has been associated with mortality in those with PAD-IC[5]. Furthermore, there is clear evidence of functional impairments of the calf musculature during level walking[6,7]. There are also indications of reduced strength in other lower limb muscle groups[8-10] with further associations between knee extensor strength and all-cause mortality[11] and some impairments in these more proximal muscle groups (i.e. knee and hip extensors) during level gait[6,7]. Given that ground reaction forces (GRF)[12,13], as well as joint range of motion[14], are larger during stair descent than level walking, the functional impairments may be even more pronounced during this daily task that is vital for active and independent living.

Stair climbing is a challenging, and often-times hazardous activity with descent in particular posing a high risk for falls in the elderly[14,15]. The moments acting about the knee and ankle joints of healthy elderly are close to maximal capacity determined from isokinetic dynamometry[16], or exceed the maximum isometric capacity of their muscle strength[17], and are greater than the relative demands experienced by younger counterparts[16]. The hip flexors are active primarily in the leg pull-through phase[18] and both hip flexors and extensors reportedly function between at only 40-50% of their maximum isometric capacity during stair walking[17]. Therefore, they may well possess a strength reserve to compensate for strength deficits distally. Recent research has demonstrated that those with PAD-IC have velocity-dependent limitations in plantarflexor strength, that are also correlated to disease severity at high contraction speeds[19], which likely necessitates compensatory strategies at more proximal muscle groups. Combined with impaired balance[20-22] and a greater prevalence for falls[22] it is likely that those with PAD-IC face significant functional challenges and risks when negotiating stairs. No study has examined the stair descent biomechanics of this population, therefore compensatory strategies and any implications for risk of falling are unknown.

The purpose of the study was to determine whether individuals with PAD-IC adopt alternative gait mechanics during stair descent. This was achieved by drawing comparisons to a control group consisting of healthy older adults and exploring relationships between gait parameters and disease severity. It was hypothesised that those with PAD-IC would demonstrate a redistribution of kinetics at weaker joints that may be working closer to their maximum capacity (plantarflexors and knee extensors) towards the joint that may possess greater strength capacity (hip musculature). Our second hypothesis was that these alterations in joint moments and powers would be associated with increased disease severity among the PAD-IC participants.

Methods

Participants

Ethical approval was granted by the National Health Service Research Ethics Committee (REC reference: 11/YH/0335). A total of 22 participants were recruited consisting of 12 individuals with PAD-IC (six unilateral, six bilateral; ten males, two females) and 10 healthy controls (four males and six females). Those with PAD-IC were recruited via consultant referral from a local outpatient vascular clinic. Male and female participants aged between 55-80 years who were diagnosed with Grade 1 Chronic Limb Ischemia[23] with the primary arterial stenosis located in the superficial femoral artery were considered for inclusion. Healthy aged-matched controls were recruited from the local community. Individuals deemed to have severe or acute cardiovascular, musculoskeletal or pulmonary illness were excluded along with those with a previous lower-limb joint replacement and observable gait abnormalities or who required a walking aid. Any individuals with a history of neurological disorders, stroke, myocardial infarction or life-limiting diseases (such as cancer), were also excluded.

Disease severity

Disease severity was determined using the ankle brachial pressure index (ABPI) at a local outpatient vascular clinic. ABPI measures for both lower limbs were taken pre- and post-standardised exercise protocol performed on a motorised treadmill (5 minutes at 2.5km/h at 10% incline)[24]. Systolic blood pressure was measured in the posterior tibial and dorsalis pedis arteries of each leg and the brachial pressure of both arms, separately, using a sphygmomanometer cuff and a hand held Doppler instrument (Parks Medical Electronics Inc, Oregon, USA). In accordance with standard protocol, the ABPI for both legs was then calculated as the higher of the two leg artery pressures normalised to the higher brachial pressure of the two arms[4]. The post-exercise ABPI was subsequently used to categorise the limbs of the participants as well as quantify disease severity. The asymptomatic limb of the unilateral claudicants was identified and excluded from further analysis; therefore from the six unilateral and six bilateral claudicants, a total of 18 claudicating-limbs were analysed. Whilst this creates a statistical imbalance between groups, all symptomatic limbs were included to better understand movement patterns within this cohort, and their relationships with disease severity. Control participants also undertook the exercise protocol to determine ABPI values and to confirm the absence of PAD-IC. Previous investigations revealed no significant differences in plantarflexor strength or power between dominant and non-dominant limbs[25] therefore, for brevity, the data for the dominant limb only, determined through a ball-kicking preference task[26], were presented.

Experimental protocol

Ten Qualisys Oqus 400 cameras (Qualisys, Gothenburg, Sweden) and 2 Kistler force plates (model 9286AA, Kistler, Winterthur, Switzerland), sampling at 100Hz and 1000Hz, respectively, were synchronised to collect kinematic and kinetic data. To avoid distal error propagation associated with the Helen Hayes model[27], a total of 47 retro-reflective passive markers (14mm diameter) were positioned according to the six Degrees-of-Freedom marker set[28] on the lower limbs and pelvis. Functional movements were used to define the hip joint centre[29]. Participants were asked to descend a custom-made five-step wooden staircase (step height; 20cm, step tread; 25cm, step width; 80cm) in a step-over-step manner. The staircase was instrumented with force plates embedded into steps two and three (step five being the top landing), and the top landing of the staircase was 1-metre in length. This allowed for approximately two steps to be taken before participants descended the staircase at a self-selected pace and participants continued walking for ~3-metres on level ground. The staircase was equipped with a safety handrail and participants were instructed to use it only when necessary. Even light handrail influences lower limb kinetics during stair ascent and descent[30], therefore trials in which the participant used the handrail were excluded from further analysis. This staircase has previously been shown to be rigid with negligible artefact or power lost from the force plate signals at physiologically relevant frequencies[31].

Data analysis

3D coordinate data were tracked using Qualysis Track Manager (V2.8, Qualysis, Gothenburg, Sweden) then exported for further processing in Visual 3D (V4, C-motion, Rockville, MD, USA). Coordinate data were interpolated using a cubic spline algorithm and both marker and kinetic data were filtered using a 2nd order low-pass Butterworth filter with a cut-off frequency of 6Hz for marker data[32] and 15Hz for kinetic data[33]. Relevant gait events were identified (foot strike and foot off) using vertical ground reaction force (≥20N threshold) and were normalised to 100% gait cycle. The focus of the present study was continuous, steady state stair descent therefore one gait cycle was defined from initial foot strike on step three to the subsequent foot strike of the same limb on step one; and from initial foot strike on step two to the subsequent foot strike of the same limb on the floor. Sub-phases of gait were specified according to McFadyen & Winter (1998) with weight-acceptance defined as 0-25% and controlled-lowering as 40-60% of the gait cycle. Inverse dynamics was used to calculated joint moments[34] and normalised to body mass (Nm/kg). The support moment was calculated as the summed moments of the hip, knee and ankle joints[35], with peak support moment quantified in both the weight-acceptance and controlled-lowering phases, and the relative contribution from these joints expressed as a percentage of peak support moment. An average of 10 trials for each participant was computed for subsequent between-group comparisons.

Between-group statistical analysis

Data were exported into SPSS v21.1 (SPSS Inc, Chicago, IL, USA), assessed for normality violations using Shapiro-Wilk's test for normality and histogram plots, and assessed for outliers through box plot analysis. Demographic data were non-parametric so an independent samples Kruskal-Wallis test was performed with subsequent pairwise comparisons where appropriate. Initial analysis detected trends towards slower walking speeds in those with PAD-IC compared to controls ($P=.060$; Table I) with a mean difference of 0.14m/s, which was a clinically meaningful difference in gait speed[36]. A univariate-analysis of variance was performed with walking speed as a co-variate, with significance accepted at $P=\geq .05$ and trends towards significance accepted at $P=<.10$.

Exploratory sub-group analysis

Joint moment profiles at the hip level in the PAD-IC group exhibited very wide variability (Figure 1) and prompted further investigation. This subsequent exploratory sub-group analysis revealed an alternate internal hip extensor moment strategy, as opposed to the stereotypical internal hip flexor moment profile[35], was being adopted by a large proportion of the claudicating-limb group (73% or 11/15 claudicating-limbs; Figure 1). Additionally, two healthy controls demonstrated similar profiles however; as these were predominantly within the first 10% of the gait cycle and not a sustained strategy throughout early stance, they were subsequently excluded from this sub-group analysis. As such, the following sub-groups were categorised: PAD-IC hip extensor strategy (PAD-IC HExt), PAD-IC hip flexor strategy (PAD-IC HFlex) and controls. Due to the exploratory nature of the research, effect sizes with 95% confidence intervals were reported.

[Figure 1]

Disease severity correlations

A Pearson's partial product-moment correlation was performed to assess relationships between disease severity (as assessed by ABPI and controlled for the influence of age) and gait parameters for the claudicant group only. A moderate relationship was accepted as R=.40–.59, a strong relationship was accepted as R=.60–.79 and a very strong relationship was accepted as R=.80–1[37].

Results

No significant overall between-group differences were found in participant characteristics (Table I).

[Table I]

Between-group differences

Ground reaction forces

During weight-acceptance, the claudicating-limb group had significantly reduced peak vertical force (1.44±0.23N/kg vs 1.79±0.24N/kg, P=.021, ES=-1.46 [-2.32,-0.69]) and reduced peak posterior force (-0.12±0.03N/kg vs -0.15±0.02N/kg, P=.016, ES=-1.08 [-0.42,-0.26]) compared to the control group.

[Figure 2]

Joint kinetics

During weight-acceptance, the claudicating-limb group demonstrated trends towards reduced hip moment (P=.079) and reduced knee power absorption in early stance compared to the control group (P=.073; Table II). Despite a 19% increase in plantarflexor moment and 27% decrease in knee extensor moment between the claudicating-limb group and healthy controls during weight-acceptance, neither were significantly reduced (P=.208 and P=.214 respectively; Table II). During controlled-lowering, the claudicating-limb group demonstrated a trend towards reduced hip flexor moment compared to healthy controls (P=.092; Table II)

[Table II]

[Figure 2]

During weight-acceptance, and in relation to contribution to peak support moment, the claudicating-limb group displayed a greater proportion of ankle moment contribution and demonstrated trends towards a smaller hip moment contribution compared to controls (ankle 42±14% vs 28±7%, P=.005 and Hip 16±8% vs 21±11%, P=.056, respectively) with comparable contribution from the knee extensors (42±17% vs 53±8%, P≥.10). During controlled-lowering, the claudicating-limb group used a greater proportion of ankle moment contributing to peak support moment and reduced hip contribution compared to controls (Ankle 41±10% vs 32±3%, P=.022 and Hip 14±5% vs 20±7%, P=.005, respectively) with a comparable contribution from the knee extensors (44±10% vs 47±8%, P≥10)

Disease severity correlations

Increased disease severity was moderately associated with trends towards reduced ankle power absorption (R=-.497, P=.100), reduced knee extensor moment (R=.557, P=.060), reduced hip flexor moment/increased hip extensor moment (R=-.453, P=.100) and a significant, strong association identified with increased hip contribution to peak support moment (R=-.760, P=.017) during weight-acceptance.

Exploratory sub-group analysis

A number of between-group differences of interest emerged (Figure 3, Table IIIa and Table IIIb). A trend towards slower walking speed was evident in the PAD-IC HExt group compared to control (0.64±0.06 vs 0.79±0.14m/s, P=.082, ES -1.46 [-2.52,-0.40]). During weight-acceptance, the PAD-IC HExt group had reduced peak vertical force (ES=-2.02 [-3.18,-0.87]), a shift from a hip flexor to hip extensor moment (ES=3.05 [1.67,4.42]), reduced knee extensor moment (ES=-0.98 [-1.98,0.02]) and increased ankle contribution to peak support moment (45±14.2% vs 28.1±7.1%, ES=1.34 [0.29,2.38]) compared to the control group. During controlled-lowering, the PAD-IC HExt group had a reduced hip extensor moment (ES=1.68 [0.59,2.78]), reduced knee power absorption (ES=1.95 [0.81,3.09]), increased ankle contribution (43.9±8.34% vs 31.2±2.7%, ES=1.78 [0.66,2.89]) and reduced hip contribution to peak support moment (12.6±3.3% vs 21.4±6.9%, ES=-1.69 [-2.78,-0.59]).

[Figure 3]

[Table IIIa and IIIb]

Discussion

This is the first study to investigate the gait adaptations of those with peripheral arterial disease and intermittent claudication (PAD-IC) during stair descent. Contrary to our hypothesis, claudicants did not redistribute joint moments away from the ankle and knee towards the hip. In fact, the plantarflexors contributed a larger percentage to the peak support moment (~40%) with reduced contribution from the hip (~15%) in the claudicating-limb group compared to controls. The exploratory analysis of sub-groups within the PAD-IC cohort revealed a shift towards utilisation of a hip extensor strategy, as opposed to the typical hip flexor moment profiles[35]. The impact of this alternate hip strategy seemed to reduce knee moments, but not ankle moments, suggesting this strategy is not a compensatory mechanism for the well-known functionally limited plantarflexors[4-7]. Our findings suggest claudicants place a greater reliance on the plantarflexor muscles than the more proximal musculature, highlighting the importance of preserving the strength of this muscle group.

The finding that claudicants do not redistribute joint kinetics away from the ankle, but rather towards, in comparison to controls, was surprising. The healthy controls in our study utilised the strength of the knee extensors much more than either the hip or ankle muscles (~50% of peak support moment), which is consistent with previous reports that this strategy allows them to operate within safer limits of maximum strength[16]. In contrast, the claudicants utilised the ankle and knee almost equally to meet the functional demands of the task (both ~ 40% of peak support moment). Given that velocity-dependent weakness in the ankle plantarflexors have previously been identified in this population[19], the greater reliance on these muscles seems counter-intuitive. This is most notable in the weight-acceptance phase that is characterised by fast, eccentric muscle action of the plantarflexors in particular, to absorb the falling body mass[35]. It could be postulated that claudicants may evade the recruitment of the larger, and therefore more metabolically demanding, knee extensor muscle group as a means of conserving energy. Recently observed remodelling of the plantarflexors, towards muscle designs favouring length changes and energy conservation[38], indicate metabolic efficiency is paramount in patients with PAD-IC. Alternatively, weakness in the knee extensors may be greater than in the plantarflexors, preventing the redistribution that is typically seen in healthy elderly[16]. When knee strength is impaired, particularly eccentric strength, it appears to have important consequences, as weakness is associated with all-cause mortality[11] and reduced functional ability[10,39]. The latter is consistent with the present results.

In partial support of our second hypothesis, trends towards associations with disease severity were identified. The moderate negative correlation between hip moment during weight-acceptance with ABPI indicates that those limbs with higher ABPI values (less severe disease) utilise the hip flexors and those limbs with lower ABPI (more severe disease) utilise the hip extensors with these hip extensors contributing more towards the support moment during weight-acceptance. Previous studies have already identified that as disease severity increases, plantarflexor[39], knee extensor[8,10] and hip extensor strength[8] all decline. The present study indicates that the result of reduced strength in these muscle groups, particularly the knee extensors, has a substantial effect on the strategies employed to descend stairs. We have previously demonstrated that the function of the plantarflexors is impaired, both in comparison to healthy controls and in relation to disease severity[19,38]. However, it is unclear whether hip extensor strength has been maintained in the full cohort, allowing for an alternate, compensatory gait strategy to be adopted. Alternatively, if hip extensor strength has declined at a comparable rate to previous reports[8], the adoption of this alternate strategy may place those with PAD-IC at a greater risk for falls.

Exploratory sub-group analysis

The identification of an alternate hip moment strategy, primarily during the weight-acceptance phase of stair descent is interesting and indicates some form of compensatory mechanism was being adopted by those with PAD-IC. There are some reports of the alternate hip moment profile in healthy young adults in previous literature[,40,41], and some data presenting group mean hip moments that appear stereotypical but with large variability in older adults[17], suggesting a few individuals may have used an alternate strategy which was not distinct within the group mean. The discrepancies between studies is possibly due to variations in trunk position resulting in a shift in the centre of pressure (CoP) and centre of mass (CoM), and therefore the location and orientation of the ground reaction force vector relative to the hip joint centre. The reduced vertical GRF during weight-acceptance, combined with this alternate hip extensor strategy, are indicative of a more upright posture and a greater reliance on the trailing limb during controlled-lowering in the PAD-IC HExt group. This is commensurate with previous reports suggesting that healthy elderly utilise the trailing limb (in the controlled-lowering phase) to minimise the downwards acceleration of the CoM rather than relying on the leading limb to decelerate the falling CoM during weight-acceptance[42]. The more upright posture would likely affect the relationship between the CoM and CoP with a reduced CoM-CoP separation, reinforcing the notion that these mechanisms are a means to increase dynamic stability in the elderly. Buckley et al. (2013)[42] suggested that these compensations act to reduce the rapid eccentric demand on the plantarflexors of the leading limb during weight-acceptance, thereby maintaining stability in the face of reduced muscular capacity. It seems that this notion still applies to those with PAD-IC. However, contrary to our hypothesis, the knee extensors appeared to benefit more from these compensations rather than the ankle plantarflexors.

Limitations and further research

It must be acknowledged that our sample size was small and grouped the symptomatic limb from unilateral claudicants with both symptomatic limbs from bilateral claudicants. This method of analysis was undertaken to better explore the movement patterns within the cohort and understand the relationship with disease severity. Despite the small sample, our findings identified interesting and functionally relevant movement strategies as well as trends towards significant differences between key variables that may well be siginifcant given a larger sample. The present study also classified unilateral and bilateral claudicants into one group, however future research should be adequately powered to evaluate the potential variations in movement patterns with these differing disease presentations. We have also made assumptions regarding upper-body posture, CoM control and strength around more proximal joints. Further research using a full-body model, supplemented by comprehensive assessments of the lower limb muscular requirements and subsequent redistribution of the task demands, are required to determine the safety and/or effectiveness of these compensations. These more detailed investiagtions could further inform targeted exercise interventions in this population.

Conclusions

This novel study has identified specific adaptations in the stair descent strategies utilised by those with PAD-IC that have important implications for function and safety. Firstly, even though the ankle plantarflexors are more functionally-limited in claudicants than healthy controls, the claudicants placed greater reliance on them during weight-acceptance. These findings strongly indicate targeted exercise interventions to maintain and/or improve plantarflexor strength are vital in this population. Secondly, exploratory analysis revealed an alternate hip moment strategy was being adopted by 73% of the PAD-IC cohort who relied on the hip extensors during weight-acceptance, as opposed to the stereotypical use of the hip flexors. This unconventional strategy acts to reduce knee extensor moments but not ankle plantarflexor moments.

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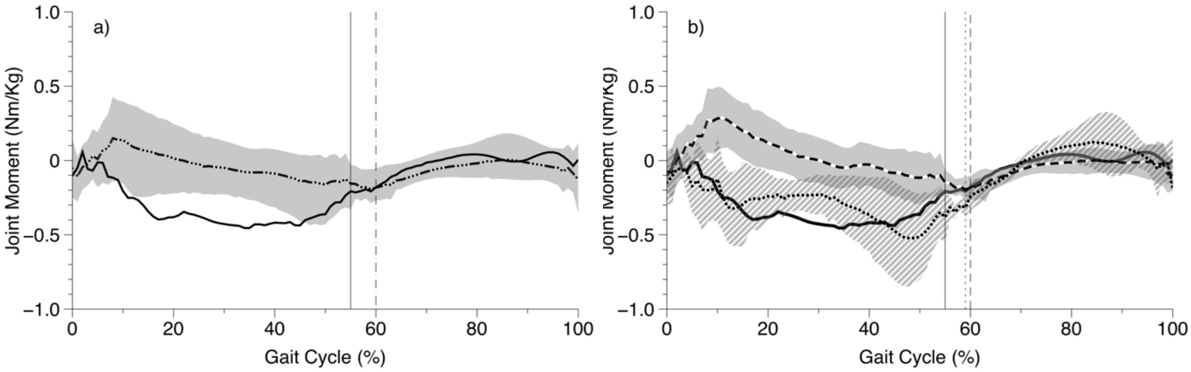

Figure 1. Group mean hip moment across 100% gait cycle with standard deviation bands and vertical lines representing toe-off for a) healthy controls (solid line) and the full PAD-IC group (dotted-dashed line); and b) healthy controls (solid), PAD-IC hip flexor strategy (dotted) and PAD-IC hip extensor strategy (dashed). Positive values indicate internal hip extensor moment.

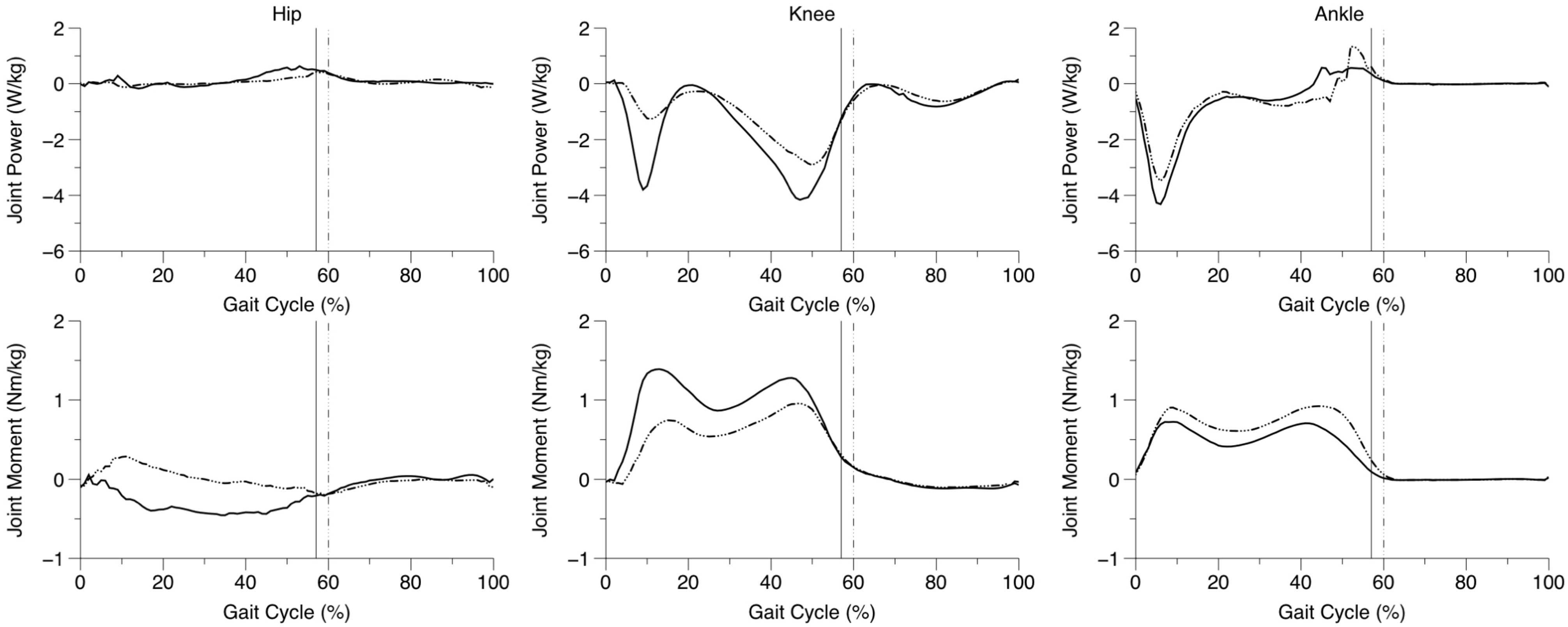

Figure 2. Group mean internal joint power (top row) and joint moment (bottom row) for the hip, knee and ankle across 100% gait cycle for claudicating-limb (dotted-dashed) and healthy controls (solid). Positive values indicate hip and knee extensor and ankle plantarflexor internal joint moment and power generation. Vertical lines represent toe-off for claudicating-limb (dashed) and healthy controls (solid).

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Figure 3. Group mean internal joint power (top row) and joint moment (bottom row) for the hip, knee and ankle across 100% gait cycle with vertical lines representing toe-off for PAD-ICHExt (dashed line), PAD-IC HFlex (dotted line), and healthy control groups (solid line). Positive values indicate hip and knee extensor and ankle plantarflexor internal joint moments and power generation.

Table I. Participant characteristics and limb analysis breakdown. Data are presented as group mean (SD).

	PAD-IC	Control
N limbs total	18	10
Breakdown	1 limb from 6 unilateral claudicants 2 limbs from 6 bilateral claudicants	10 dominant limbs
Age (years)	64.7 (7.1)	60.0 (4.5)
Height (m)	1.72 (0.08)	1.60 (0.06)
Mass (Kg)	83.3 (18.8)	68.6 (10.1)
ABPI pre-exercise	0.80 (0.21)	1.01 (0.11)
ABPI post-exercise	0.56 (0.20)	0.99 (0.18)
Stair descent speed (m/s)	0.65 (0.08)	0.79 (0.14)

Table II. Group mean (SD) sagittal plane peak joint kinetics with effect sizes [95% confidence intervals]. Positive values indicate ankle plantarflexor, knee extensor and hip extensor moment. ^ represent between-group trends towards significance (P<.10).

Weight acceptance	**Claudicating limb**	**Control**	**ES [95% CI]**
Hip moment (Nm/kg)	-0.35 (0.22) ^	-0.53 (0.22)	0.79 [-0.01,1.59]
Hip power absorption (W/kg)	-0.28 (0.23)	-0.44 (0.27)	0.64 [-0.16,1.43]
Knee moment (Nm/kg)	1.11 (0.56)	1.52 (0.57)	-0.71 [-1.50,0.09]
Knee power absorption (W/kg)	-2.38 (1.66) ^	-5.23 (2.96)	2.00 [1.07,2.93]
Ankle moment (Nm/kg)	1.01 (0.28)	0.83 (0.26)	0.64 [-0.15,1.43]
Ankle power absorption (W/kg)	-3.78 (1.42)	-4.67 (2.03)	0.52 [-0.26,1.31]
Support moment (Nm/kg)	2.58 (0.57)	2.95 (0.98)	-0.49 [-1.27,0.30]
Controlled lowering	**Claudicating limb**	**Control**	**ES [95% CI]**
Hip moment (Nm/kg)	-0.35 (0.20) ^	-0.56 (0.18)	1.03 [0.22,1.85]
Hip power generation (W/kg)	0.70 (0.39)	0.88 (0.41)	-0.44 [-1.22,0.34]
Knee moment (Nm/kg)	1.22 (0.50)	1.34 (0.39)	-0.25 [1.03,0.53]
Knee power absorption (W/kg)	-4.17 (1.42)	-4.73 (1.62)	0.36 [-0.41,1.14]
Ankle moment (Nm/kg)	1.04 (0.18)	0.84 (0.29)	0.87 [0.06,1.67]
Ankle power generation (W/kg)	2.26 (0.72)	2.04 (1.23)	0.23 [-0.55,1.01]
Support moment (Nm/kg)	2.49 (0.42)	2.76 (0.59)	-0.54 [-1.33,0.25]

Table IIIa. Group mean (SD) ground reaction forces (GRF), sagittal plane peak joint kinetics and percentage joint contributions to total support moment alongside effect sizes and 95% confidence intervals during weight acceptance. Positive values indicate vertical and anterior GRF, ankle plantarflexor, knee extensor and hip extensor moments.

Weight acceptance	PAD-IC HExt	PAD-IC HFlex	Control	PAD-IC HExt vs Control ES [95% CI]	PAD-IC HFlex vs Control ES [95% CI]	PAD-IC HExt vs PAD-IC HFlex ES [95% CI]
Peak vertical GRF (N/Kg)	1.29 (0.26)	1.56 (0.21)	1.85 (0.27)	-2.02 [-3.18,-0.87]	-1.05 [-2.36,0.25]	-1.02 [-2.22,0.18]
Peak posterior GRF (N/Kg)	-0.10 (0.31)	-0.14 (0.04)	-0.14 (0.23)	-0.11 [-1.06,0.84]	0.00 [-1.23,1.23]	-0.11 [-1.26,1.03]
Hip moment (Nm/kg)	0.37 (0.19)	-0.39 (0.22)	-0.50 (0.37)	3.05 [1.67, 4.42]	0.31 [-0.93,1.54]	3.62 [1.89,5.35]
Hip power absorption (W/kg)	-0.48 (0.27)	-0.26 (0.25)	-0.28 (0.14)	-0.83 [-1.81,0.16]	0.10 [-1.13,1.33]	-0.78 [-1.96,0.40]
Knee moment (Nm/kg)	0.88 (0.55)	1.60 (0.27)	1.48 (0.64)	-0.98 [-1.98,0.02]	0.20 [-1.03,1.43]	-1.36 [-2.60,-0.11]
Knee power absorption (W/kg)	-1.44 (1.37	-2.51 (1.65)	-5.21 (2.34)	2.00 [0.85,3.15]	1.16 [-0.16,2.48]	0.70 [-0.47,1.87]
Ankle moment (Nm/kg)	0.93 (0.27)	1.08 (0.27)	0.75 (0.26)	0.64 [-0.33,1.61]	1.15 [-0.17,2.46]	-0.52 [-1.68, 0.64]
Ankle power absorption (W/kg)	-3.82 (1.90)	-5.05 (1.49)	-4.69 (2.11)	0.42 [-0.54,1.38]	-0.17 [-1.40,1.06]	0.64 [-0.53,1.81]
Support moment (Nm/kg)	2.13 (0.61)	3.07 (0.23)	2.73 (0.79)	-0.84 [-1.82,0.15]	0.47 [-0.77,1.72]	-1.62 [-2.90,0.34]

Table IIIb. Group mean (SD) ground reaction forces (GRF), sagittal plane peak joint kinetics and percentage joint contributions to total support moment alongside effect sizes and 95% confidence intervals during controlled lowering. Positive values indicate vertical and anterior GRF, ankle plantarflexor, knee extensor and hip extensor moments.

Controlled lowering	**PAD-IC HExt**	**PAD-IC HFlex**	**Control**	**PAD-IC HExt vs Control ES [95% CI]**	**PAD-IC HFlex vs Control ES [95% CI]**	**PAD-IC HExt vs PAD-IC HFlex ES [95% CI]**
Peak vertical GRF (N/Kg)	1.01 (0.16)	0.99 (0.15)	0.99 (0.16)	0.12 [-0.83,1.07]	0.00 [-1.23,1.23]	0.12 [-1.03,1.26]
Peak anterior GRF (N/Kg)	0.10 (0.05)	0.14 (0.06)	0.18 (0.22)	-0.54 [-1.51,0.42]	-0.20 [-1.43,1.03]	-0.72 [-1.89,0.46]
Hip moment (Nm/kg)	-0.28 (0.09)	-0.66 (0.30)	-0.53 (0.20)	1.68 [0.59,2.78]	-0.50 [-1.75,0.75]	2.18 [0.79,3.56]
Hip power generation (W/kg)	0.52 (0.28)	0.66 (0.21)	0.71 (0.26)	-0.66 [-1.64,0.31]	-0.19 [-1.42,1.04]	-0.50 [-1.65,0.66]
Knee moment (Nm/kg)	1.02 (0.41)	1.55 (0.73)	1.30 (0.37)	-0.66 [-1.64,0.31]	0.44 [-0.80,1.68]	-0.98 [-2.18,0.22]
Knee power absorption (W/kg)	-3.15 (0.83)	-6.12 (1.37)	-4.93 (0.93)	1.95 [0.81,3.09]	-0.99 [-2.29,0.30]	2.85 [1.32,4.38]
Ankle moment (Nm/kg)	0.99 (0.20)	1.01 (0.15)	0.84 (0.14)	0.79 [-0.19,1.78]	1.66 [0.25,3.07]	-0.55 [-1.77,0.62]
Ankle power generation (W/kg)	2.24 (1.36)	2.72 (0.74)	2.03 (1.41)	0.15 [-0.80,1.09]	0.51 [-0.73,1.76]	-0.36 [-1.51,0.79]
Support moment (Nm/kg)	2.32 (0.54)	3.22 (0.87)	2.71 (0.51)	-0.70 [-1.68,0.27]	0.71 [-0.55,1.98]	-1.34 [-2.58,0.10]