



LEEDS
BECKETT
UNIVERSITY

Citation:

Hanley, B and Bissas, A (2013) Analysis of lower limb internal kinetics and electromyography in elite race walking. *Journal of sports sciences*, 31 (11). 1222 - 1232. ISSN 0264-0414 DOI: <https://doi.org/10.1080/02640414.2013.777763>

Link to Leeds Beckett Repository record:

<https://eprints.leedsbeckett.ac.uk/id/eprint/192/>

Document Version:

Article (Accepted Version)

The aim of the Leeds Beckett Repository is to provide open access to our research, as required by funder policies and permitted by publishers and copyright law.

The Leeds Beckett repository holds a wide range of publications, each of which has been checked for copyright and the relevant embargo period has been applied by the Research Services team.

We operate on a standard take-down policy. If you are the author or publisher of an output and you would like it removed from the repository, please [contact us](#) and we will investigate on a case-by-case basis.

Each thesis in the repository has been cleared where necessary by the author for third party copyright. If you would like a thesis to be removed from the repository or believe there is an issue with copyright, please contact us on openaccess@leedsbeckett.ac.uk and we will investigate on a case-by-case basis.

Analysis of lower limb internal kinetics and electromyography in elite race walking

Brian Hanley and Athanassios Bissas

School of Sport, Carnegie Faculty, Headingley Campus, Leeds Metropolitan University,
United Kingdom

Correspondence details:

Brian Hanley,

Fairfax Hall,

Headingley Campus,

Leeds Metropolitan University,

LS6 3QS,

United Kingdom.

Telephone: +44 113 812 3577

Fax: +44 113 283 3170

Email: b.hanley@leedsmet.ac.uk

Running title: Lower limb internal kinetics and EMG in elite race walking

Keywords: Elite athletes, gait, inverse dynamics, joint kinetics, track and field

ABSTRACT

The aim of this study was to analyse lower limb joint moments, powers and electromyography patterns in elite race walking. Twenty international male and female race walkers performed at their competitive pace in a laboratory setting. The collection of ground reaction forces (1000 Hz) was synchronised with two-dimensional high-speed videography (100 Hz) and electromyography of seven lower limb muscles (1000 Hz). As well as measuring key performance variables such as speed and stride length, normalised joint moments and powers were calculated. The rule in race walking which requires the knee to be extended from initial contact to midstance effectively made the knee redundant during stance with regard to energy generation. Instead, the leg functioned as a rigid lever which affected the role of the hip and ankle joints. The main contributors to energy generation were the hip extensors during late swing and early stance, and the ankle plantarflexors during late stance. The restricted functioning of the knee during stance meant that the importance of the swing leg in contributing to forward momentum was increased. The knee flexors underwent a phase of great energy absorption during the swing phase and this could increase the risk of injury to the hamstring muscles.

INTRODUCTION

Race walking is part of the athletics programme at the Olympic Games and other major championships. Its rules state that no visible loss of contact with the ground should occur and that the knee must be straightened from the moment of first contact with the ground until the ‘vertical upright position’ (Rule 230.1) (IAAF, 2011). Race walkers need to develop performance parameters while perfecting their technique so they remain within the rules and avoid disqualification (Hanley, Bissas & Drake, 2011). The technical precision demanded by the event’s rules means that understanding the underlying biomechanical factors is particularly important.

The patterns of joint internal kinetics and electromyography (EMG) in normal walking and running have been researched extensively (Zajac, Neptune & Kautz, 2003). To date, similar research on race walking has been sparse although with its increasing popularity (e.g. a women’s 20 km race was first held as part of the Olympic programme in 2000) such research is now invaluable to elite athletes and their coaches. It is important to note that the present rule governing the knee during stance was only introduced in 1995. As a result, earlier studies no longer describe either the joint kinetics (White & Winter, 1985; Cairns, Burdette, Pisciotta & Simon, 1986) or EMG patterns (Murray, Guten, Mollinger & Gardner, 1983) in modern race walkers. With regard to more recent studies, Hoga, Ae, Enomoto & Fujii (2003) used video footage of men’s competitions to measure swing leg moments and powers. While the sampling rate of 60 Hz used is sufficient for basic kinematic analysis, it might not be suitable for calculation of internal kinetics, especially given the short duration of the swing phase. Hoga, Ae, Enomoto, Yokozawa & Fujii (2006) analysed the race walking stance phase separately in a laboratory study using sub-elite men. Because the swing and stance phases are not exclusive gait events but are instead interdependent, analysis of a complete gait cycle is

required to assess overall performance. Furthermore, sub-elite athletes have been shown to differ from elite athletes with regard to key biomechanical variables and therefore the analysis of sub-elite athletes does not adequately describe elite performances (e.g. Leskinen, Häkkinen, Virravirta, Isolehto & Kyröläinen, 2009). In another laboratory study, Donà, Preatoni, Cobelli, Rodano & Harrison (2009) only reported moments at the knee, and while the knee is of particular interest, its contribution to race walking gait needs to be considered in the context of other lower limb joints.

It is also worth noting that no previous study on race walking has combined the measurement of joint moments and powers with EMG, a technique which increases the validity of the power data by adding a biological perspective to the mechanical analysis. Therefore, literature on race walking biomechanics is sparse and as a result no clear relationships have been established between key biomechanical parameters and performance criteria. New research on internal kinetics and EMG of elite race walkers which incorporates both swing and stance phases in a single experimental set-up, and includes both men and women, is required to fully appreciate this unique form of locomotion. The findings of this research can assist coaches and athletes to understand the key biomechanical factors affecting technique and what measures can be taken to improve performance. The aim of this study was to measure and analyse the lower limb joint moments, powers and EMG patterns in elite international male and female race walkers.

METHODS

Participants

The study was approved by the university's Research Ethics Committee and twenty international race walkers gave written informed consent. The twenty athletes comprised ten men (23 ± 5 yrs, $1.79 \pm .06$ m, 67.0 ± 9.4 kg) and ten women (22 ± 5 yrs, $1.69 \pm .05$ m, 53.9 ± 5.6 kg). Five of the men and five women had competed at the Olympic Games or World Championships or had qualified for the 2012 Olympic Games at the time of testing. The remaining younger athletes included one woman who was World Youth Champion, and two others had competed at the World Junior Championships. Two other athletes competed at the European Junior Championships and one at the European Youth Olympic Trials, while the other young athletes had competed at the European and World Cups.

Data collection

Each athlete race walked along a 45 m indoor running track at a speed equivalent to their season's best time (10 km for juniors, 20 km or 50 km for seniors dependent on specialism). Timing gates were placed 4 m apart around two force plates (Kistler, Winterthur) which recorded both left and right foot contact phases and flight time. Athletes completed at least ten trials and the three closest to the target time were analysed (provided they were within 3% of the target time). The force plates recorded at 1000 Hz and were placed in a customised housing in the centre of the track. The force plates were covered with a synthetic athletic running surface so that the force plate area was flush with the rest of the runway to preserve ecological validity (Bezodis, Kerwin & Salo, 2008).

In order to analyse the sagittal plane movements of the hip, knee and ankle, two-dimensional video data were collected at 100 Hz using a high-speed camera (RedLake, San Diego).

Sagittal plane movements were the focus of this study as they are the most important movements in race walking and 2D measurements are therefore appropriate for this type of analysis (Alkjaer, Simonsen & Dyhre-Poulsen, 2001). The shutter speed was 1/500 s, the *f*-stop was 2.0, and there was no gain. The camera was placed approximately 12 m from and perpendicular to the line of walking. The resolution of the camera was 1280 x 1024 pixels. Extra illumination was provided by 26 lights providing 4 kW each of overhead floodlighting. Four reference poles were placed spread out across the 4 m data collection area in the centre of the running track. The poles provided twelve reference points for vertical and horizontal calibration.

Surface EMG signals were recorded from seven muscles of the right leg: gluteus maximus (GM), biceps femoris (BF), rectus femoris (RF), vastus lateralis (VL), gastrocnemius (lateral head) (GL), soleus (SO) and tibialis anterior (TA). Skin preparation involved shaving and cleansing of the surface with alcohol swabs (Okamoto, Tsutsumi, Goto & Andrew, 1987). The single differential electrodes (DelSys, Boston) consisted of two silver bars 10 mm long, 1 mm wide, and 10 mm apart in a polycarbonate casing. The reference electrode was placed over the fourth lumbar vertebra. After identifying the appropriate site by palpating the contracted muscle, each electrode was placed over the muscle belly, aligned parallel to the underlying muscle fibre direction (Clarys & Cabri, 1993). A telemetry unit (DelSys, Boston) was used to collect the data at 1000 Hz. The Common Mode Rejection Ratio (CMRR) was 92 dB, the bandwidth frequency between 20 and 450 Hz, the gain was 1000 V/V, and the input impedance greater than 100 G Ω . EMG data collection lasted 5 s and began approximately 2 s before initial contact with the force plates. The EMG collection software was synchronised with both the force plate software (Kistler, Winterthur) and the camera system (RedLake, San

Diego) using a Trigger Output Module (Wireless) (National Instruments, Austin) and a Kistler connection box (Kistler, Winterthur).

Data analysis

The video files were manually digitised by a single experienced operator to obtain kinematic data (SIMI Motion, Munich). Digitising was started at least 10 frames before the beginning of the stride and completed at least 10 frames after to provide padding during filtering (Smith, 1989). Reliability of the digitising process was estimated by repeated digitising of one race walking sequence with an intervening period of 48 hours. The results showed minimal systematic and random errors and therefore confirmed the high reliability of the digitising process. Whole body digitisation was conducted and de Leva's (1996) body segment parameter models used to obtain data for the whole body centre of mass, right thigh, right lower leg, and right foot. A cross-validated quintic spline was used to smooth the data prior to displacement calculations whereas a recursive second-order, low-pass Butterworth digital filter (zero phase-lag) was employed to filter the calculations of the 1st and 2nd derivatives (Giakas & Baltzopoulos, 1997a,b). The cut-off frequencies were calculated using residual analysis (Winter, 2005) and ranged from 7.0 to 11.6 Hz.

Race walking speed was determined as the average horizontal speed of the centre of mass during one complete gait cycle, which was defined from the start of swing to the end of stance. Stride length was measured as the distance between successive right foot contacts using the digitised data. Stride length was also expressed as a percentage of the participants' statures, and referred to as stride length ratio. Cadence was calculated as the reciprocal of stride time. To calculate stride time, the two contact phases plus the flight phase between them were measured using the force plates. These times were added to the duration of the

first flight phase which was measured using the video data (as the initial right toe-off did not occur on a force plate). ‘Foot ahead’ was used to describe the horizontal distance from the right foot to the centre of mass at initial contact. Similarly, ‘foot behind’ was the horizontal distance from the right foot to the centre of mass at the final instant of contact. Both of these distances were also expressed as a proportion of stature and referred to as foot ahead ratio and foot behind ratio respectively. The hip angle was defined as the sagittal plane angle between the trunk and thigh segments. The knee angle was calculated as the sagittal plane angle between the thigh and leg segments. Both hip and knee angles were considered to be 180° in the anatomical standing position and angles beyond 180° as hyperextension. The ankle angle was calculated using the lower leg and foot segments and considered to be 110° in the anatomical standing position (Cairns et al., 1986). With regard to kinetic data, peak braking and propulsive anteroposterior forces were expressed as bodyweights (BW) and the braking-to-propulsion value represented the time spent in the braking phase as a percentage of contact time.

The kinetic data were matched with the kinematic data (Bezodis et al., 2008) and extracted at 100 Hz. The kinetic data included anteroposterior and vertical ground reaction forces and centre of pressure data; the kinematic data included segment and joint linear and angular velocities and accelerations. These data were used to calculate net joint moments using a link segment rigid body model (Winter, 2005). Power was calculated by multiplying the moment by the joint angular velocity; positive power indicated that mechanical energy was being generated, while negative power indicated energy absorption (White & Winter, 1985). In order to account for different body sizes in this study, and in order to compare results with similar previous research, muscle moments were normalised using body mass and stature, while powers were normalised using body mass only (Hoga et al., 2006). Because the mean

swing duration was 54.1% of total stride time, each athlete's swing data were interpolated to 55 points and their stance data to 46 points (for a total of 101 points) using a cubic spline to normalise both periods of the gait cycle. The raw EMG signals were processed using average rectified EMG (AREMG), with a moving time window of 50 ms and an overlap of 25 ms. These data were also interpolated to 101 points using a cubic spline. In order to facilitate the identification of key events during the gait cycle, specific peaks on each moment and power trace are labelled (A1, K2, etc.) in a similar fashion to previous studies (White & Winter, 1985; Bezodis et al., 2008).

Statistical analysis

Independent *t*-tests were conducted to compare values between men and women, with adjustments made if Levene's test for equality of variances was less than 0.05. Pearson's product moment correlation coefficient was used to find associations between normalised joint moments and powers and key variables in race walking; an alpha level of 5% was set.

RESULTS

The values for key kinematic, kinetic and temporal variables are shown in Table 1. Race walking speed was correlated with stride length ($r = .66, P = .002$), stride length ratio ($r = .52, P = .018$), cadence ($r = .66, P = .002$), contact time ($r = -.59, P = .006$), and foot behind (absolute: $r = .53, P = .017$; ratio: $r = .45, P = .047$). The men were faster than the women, and while men's mean stride length was longer when expressed as an absolute value, stride length ratio was not different. Similarly, men had longer foot ahead and foot behind distances, but neither of these differed between men and women when expressed as a ratio. There were no differences for any temporal variable, joint angle, or normalised kinetic variable. Several absolute moment and power peaks were found to differ between men and women, but the vast majority of differences were not present after normalising (a similar outcome was found when comparing senior and junior athletes). Because there were so few significant differences between genders when normalised, these elite athletes have been considered a single group to show typical movement patterns at their high level of performance. The traces of the averaged joint angles, normalised moments, and normalised powers of the ankle, knee and hip (with standard deviations) as well as associated EMG patterns are shown in Figures 1, 2 and 3 respectively.

Table 1. Mean ($\pm s$) values for key kinematic and kinetic variables in elite male and female race walkers. Between subject effects were significant at $P < .05$ (shown in bold).

	Men	Women	<i>t</i>	<i>P</i>
Speed ($\text{km}\cdot\text{h}^{-1}$)	13.66 \pm 0.39	12.49 \pm 0.89	3.82	.001
Stride length (m)	2.44 \pm 0.04	2.24 \pm 0.08	7.60	< .001
Stride length ratio (%)	135.4 \pm 5.3	131.1 \pm 5.7	1.72	.103
Foot ahead (m)	0.39 \pm 0.03	0.35 \pm 0.02	3.89	.001
Foot ahead ratio (%)	21.8 \pm 1.4	20.8 \pm 1.1	1.71	.104
Foot behind (m)	0.50 \pm 0.02	0.46 \pm 0.02	4.09	.001
Foot behind ratio (%)	27.7 \pm 1.2	26.9 \pm 1.4	1.41	.174
Cadence (Hz)	1.55 \pm 0.05	1.55 \pm 0.11	0.09	.927
Contact time (s)	.297 \pm .021	.299 \pm .027	0.13	.899
Flight time (s)	.027 \pm .014	.026 \pm .008	0.17	.865
Peak braking force (BW)	-0.37 \pm 0.09	-0.37 \pm 0.05	0.24	.812
Peak propulsive force (BW)	0.24 \pm 0.04	0.26 \pm 0.04	1.13	.274
Braking-to-propulsion (%)	42.5 \pm 7.0	42.4 \pm 5.4	0.04	.966

Ankle

At toe-off, the ankle was plantarflexed at an angle of $132^\circ (\pm 6)$ which decreased by approximately 40° during early swing. The energy generating dorsiflexor moment which caused this action decreased considerably during mid- and late swing but noticeable tibialis anterior activity continued throughout the swing phase while the activity of the triceps surae remained relatively low. During early stance, the average ankle dorsiflexion angle decreased by approximately 20° to reach the anatomical standing position (110°). This movement was initiated by a short plantarflexion moment (A1) but was quickly interrupted by an energy absorbing dorsiflexion moment (A2) before the plantarflexor moment continued. In the midstance phase, the ankle dorsiflexed as the lower leg rotated about the ankle joint. This motion coincided with an energy absorbing plantarflexor moment (A3) which lasted for approximately 15% of total stride time. The EMG amplitudes of gastrocnemius and soleus increased considerably during this period, indicating an eccentric loading of the triceps surae muscle-tendon unit. Moving into late stance, the activity of the triceps surae gradually decreased while the preparation for toe-off was controlled by a powerful energy generating plantarflexion moment which peaked at about 80% of total stride time. Thereafter, this plantarflexor moment reduced rapidly and a small dorsiflexor moment occurred with a reversal of the EMG pattern from triceps surae activity to tibialis anterior activity. There were no correlations between any of the peak values for ankle moments or powers and the variables featured in Table 1.

Knee

In early swing, the knee flexed from a toe-off angle of $145^\circ (\pm 5)$ to a maximum flexion value of $107^\circ (\pm 5)$. The knee experienced an extensor moment during this phase which actually began in late stance (K1) and decreased in magnitude until approximately 20% of stride time (midswing) before experiencing an energy absorbing flexor moment during late swing (K2) as the knee extended to $180^\circ (\pm 3)$ at initial contact. Peak knee extension angular velocity ($10.14 \text{ rad/s} \pm 0.81$) occurred during the late swing phase at 38% of stride time. The EMG traces showed slight rectus femoris activity during early swing and a high activity of biceps femoris during late swing, while vastus lateralis activity gradually increased towards heel strike. The peak knee moments and powers during swing (K1 / K2) were correlated with race walking speed (Table 2). The early swing moment was the only one correlated with stride length, but both peak swing moments and powers were correlated with cadence. Furthermore, these swing moments were both correlated with foot behind ($r = .54, P = .015$ and $r = -.48, P = .033$ respectively), but only the early swing moment (K1) was correlated with foot ahead ($r = .54, P = .013$).

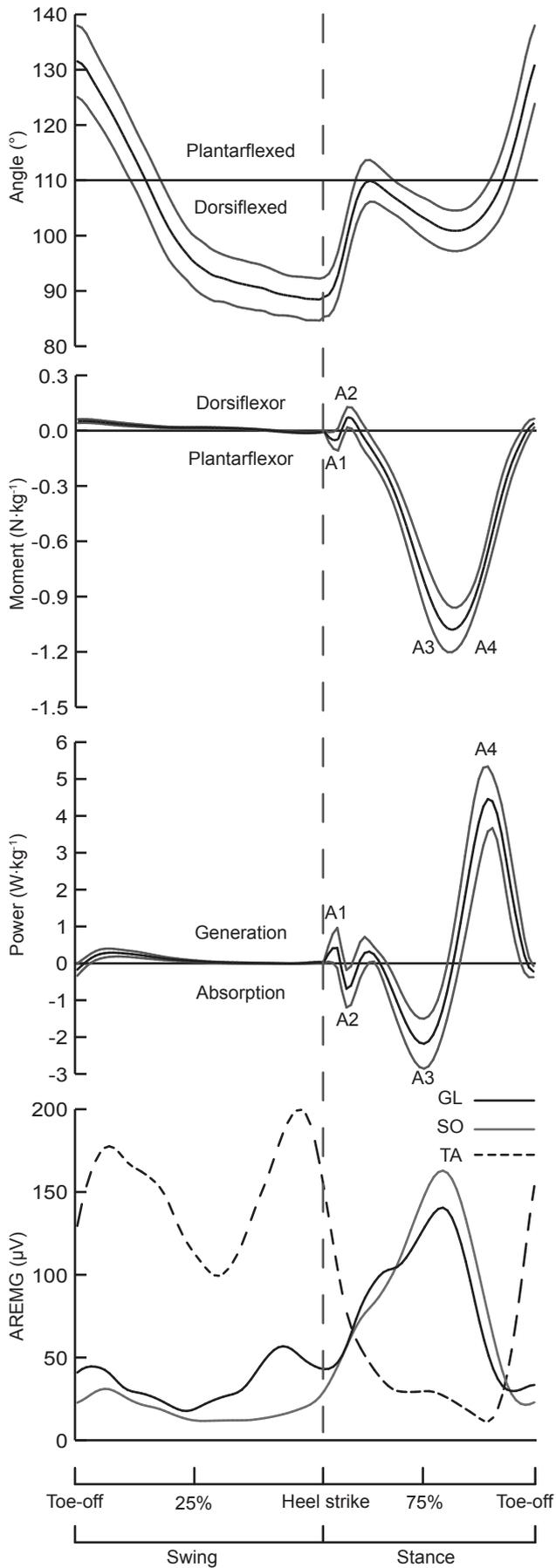


Figure 1. Ankle joint angle, moment, and power means (\pm s) and AREMG during a race walking stride. The vertical dashed line represents the transition from swing to stance.

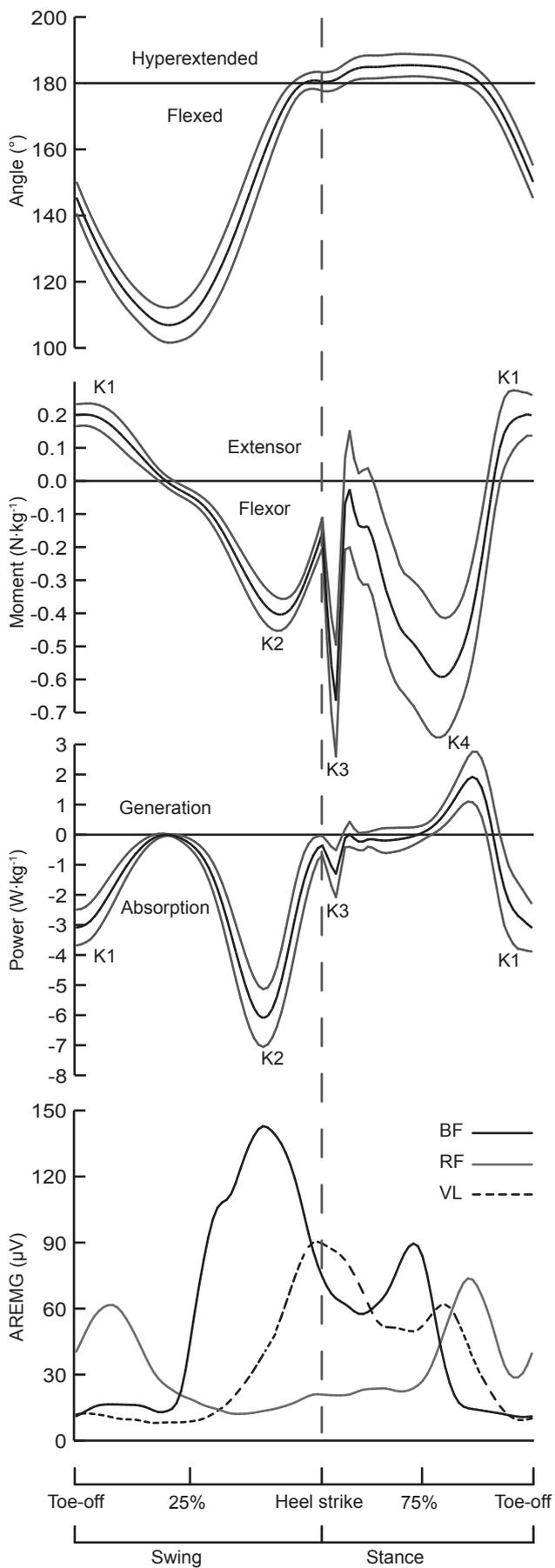


Figure 2. Knee joint angle, moment, and power means (+ s) and AREMG during a race walking stride. The vertical dashed line represents the transition from swing to stance.

Table 2. Correlation analysis of key race walking variables with selected joint moments and powers. Correlations were significant at $P < .05$ (shown in bold).

	Speed	Stride length	Stride length ratio	Cadence	Knee at contact	Knee at midstance
Knee						
K1 moment	$r = .78$	$r = .55$	$r = .20$	$r = .46$	$r = -.54$	$r = -.36$
	$p < .001$	$p = .012$	$p = .398$	$p = .041$	$p = .014$	$p = .119$
K1 power	$r = -.77$	$r = -.37$	$r = -.11$	$r = -.63$	$r = .47$	$r = .32$
	$p < .001$	$p = .109$	$p = .652$	$p = .003$	$p = .035$	$p = .169$
K2 moment	$r = -.63$	$r = -.39$	$r = -.13$	$r = -.47$	$r = .38$	$r = .27$
	$p = .003$	$p = .094$	$p = .580$	$p = .036$	$p = .097$	$p = .244$
K2 power	$r = -.49$	$r = .05$	$r = -.13$	$r = -.65$	$r = .27$	$r = .09$
	$p = .029$	$p = .834$	$p = .600$	$p = .002$	$p = .249$	$p = .705$
Hip						
H3 moment	$r = .70$	$r = -.13$	$r = .12$	$r = .74$	$r = -.45$	$r = -.20$
	$p = .001$	$p = .589$	$p = .615$	$p < .001$	$p = .047$	$p = .397$
H3 power	$r = .35$	$r = -.27$	$r = .06$	$r = .69$	$r = -.23$	$r = -.46$
	$p = .128$	$p = .260$	$p = .813$	$p = .001$	$p = .321$	$p = .039$
H4 moment	$r = .32$	$r = .33$	$r = -.04$	$r = .07$	$r = -.51$	$r = .11$
	$p = .171$	$p = .151$	$p = .877$	$p = .777$	$p = .023$	$p = .632$
H4 power	$r = .38$	$r = .43$	$r = .01$	$r = .04$	$r = -.53$	$r = .14$
	$p = .098$	$p = .060$	$p = .969$	$p = .873$	$p = .017$	$p = .569$

A brief energy absorbing flexor moment (K3) occurred within the first 0.05 s of stance which was followed by a larger flexor moment as the knee hyperextended to a maximum of $185^\circ (\pm 3)$; the athletes spent 66% (± 14) of stance time with the knee hyperextended. After midstance, the flexor moment generated energy (K4) to allow the knee to flex and prepare for swing; this moment was correlated with cadence ($r = .58, P = .007$). During late stance, an energy absorbing extensor moment occurred as the knee flexed (K1). Vastus lateralis activity was noticeable during the knee extension phase, although rectus femoris activity was not. The vastus lateralis activity continued to be present for the majority of the stance phase. Biceps femoris activity was also relatively high, peaking during midstance before diminishing prior to toe-off. During this late stance phase, rectus femoris activity increased to its highest magnitude during the gait cycle.

Hip

During early swing, the hyperextended hip ($185 \pm 4^\circ$) flexed due to an energy generating flexor moment (H1), coincident with activity of rectus femoris. Peak hip flexion angular velocity ($4.79 \text{ rad/s} \pm 1.18$) occurred at 15% of stride time (during the early swing phase). During midswing, an extensor moment absorbed energy (H2) with increased gluteus maximus activity observed from 20 to 35% of stride time. Following this the hip flexed to its furthest point of $150^\circ (\pm 6)$. During late swing, hip extension occurred due to the extensor moment becoming energy generating (H3) with high EMG activity in the two hip extensors analysed, gluteus maximus and especially biceps femoris. The early- and mid-swing moments and powers did not correlate with any key variables, but the late swing moment (H3) was correlated with both speed and cadence, and its associated power with cadence only (Table 2).

At initial contact, the hip angle was $170^{\circ} (\pm 3)$. A second, larger energy generating extensor moment (H4) occurred during early stance which extended the hip to approximately 180° . Both the early stance moment and power values were correlated with the knee angle at initial contact (Table 2), and power only with foot ahead distance ($r = .53, P = .016$). Additionally, this moment was negatively correlated with peak braking force ($r = -.45, P = .048$) and the duration of the braking phase ($r = -.47, P = .038$). Following this, a smaller extensor moment occurred as the hip hyperextended, and continued biceps femoris and gluteus maximus activity was observed. During midstance, a flexor moment began which absorbed energy. The hip reached a maximum hyperextension angle of $190^{\circ} (\pm 4)$ at which point the flexor moment generated energy reducing the hyperextension angle prior to toe-off and continuing the flexion movement into early swing. Rectus femoris was active during this flexor moment while the activity of the hip extensors diminished.

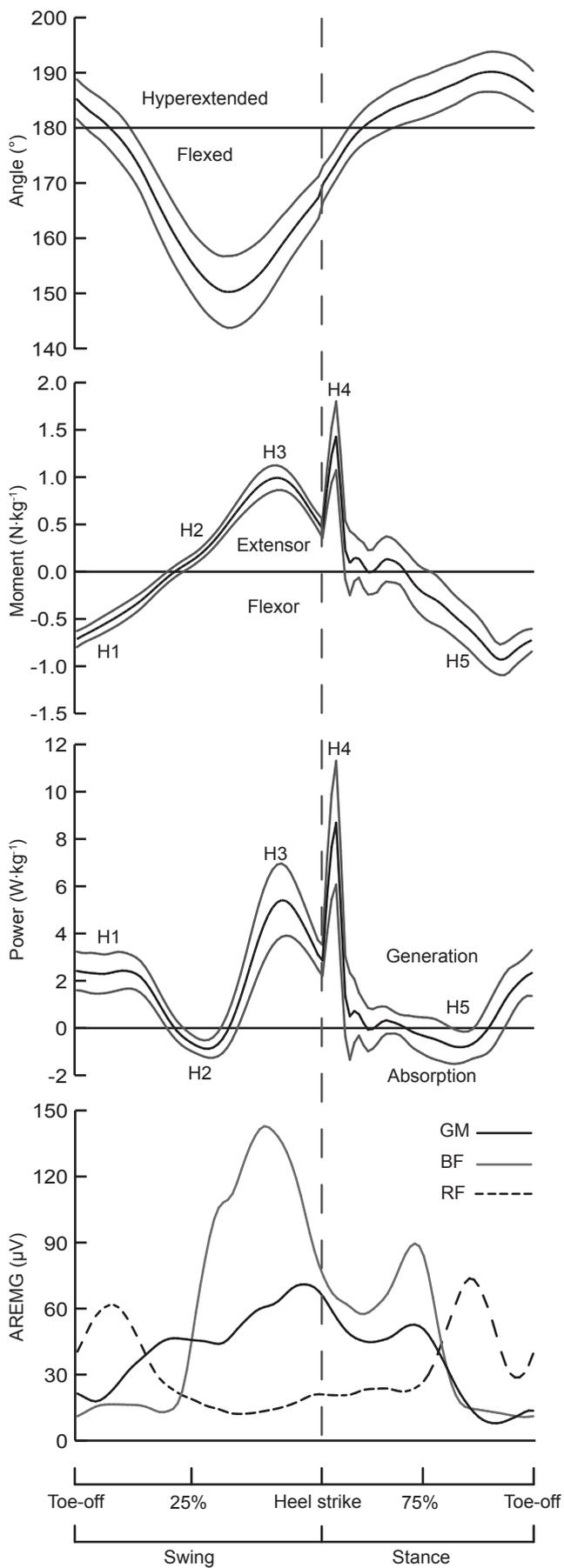


Figure 3. Hip joint angle, moment, and power means (\pm s) and AREMG during a race walking stride. The vertical dashed line represents the transition from swing to stance.

DISCUSSION

The aim of this study was to measure and analyse the lower limb joint moments, powers and EMG patterns in elite international race walkers. The differences between sexes for moment and power peaks were not present when normalised for anthropometric measurements, a finding replicated for performance variables such as stride length and peak braking force and so elite men and women had similar movement patterns. During early stance, the foot's rapid progress from heel strike to a flat foot position was controlled by a short energy absorbing dorsiflexor moment. The concurrent eccentric contraction of the tibialis anterior might be one cause of the high incidence of shin muscle pain found in race walkers (Sanzén, Forsberg & Westlin, 1986; Francis, Richman & Patterson, 1998). The brief energy generating plantarflexor moment prior to this appeared to be an attempt to 'unlock' the ankle from a dorsiflexed position to permit a smooth landing movement to occur and engage the ankle earlier in the stance phase. This active landing might be a feature of elite performers given its absence in studies of less skilled race walkers where a more passive landing occurred (White & Winter, 1985; Hoga et al., 2006). During late stance, the ankle plantarflexor moment generated the largest source of mechanical energy during the propulsive phase (4.5 ± 0.9 W/kg). Despite this, and in contrast to Hoga et al. (2006), the final plantarflexor moment was not correlated with race walking speed. The energy absorbing plantarflexor moment which occurred immediately prior to this might have been an important source of elastic energy storage in the gastrocnemius and soleus whose energy return contributed to the final plantarflexion movement (Zajac et al., 2003), as suggested by the EMG graphs which showed diminished activity in the triceps surae prior to toe-off.

At the hip, the gluteus maximus contracted first during midswing to halt flexion and then contracted together with biceps femoris to generate energy and reverse the hip's movement

into extension. The magnitude of this extensor moment was correlated with race walking speed and cadence, similar to previous findings (Hoga et al., 2003), confirming the critical role of the hip extensors even before the contact phase. During early stance, a large extensor moment continued extension of the hip. While it was not correlated with speed, this normalised hip moment and its associated power (1.4 ± 0.4 N/kg and 8.7 ± 2.6 W/kg) were the largest observed across all joints during the gait cycle and showed the hip extensors' crucial role in generating mechanical energy. Furthermore, the negative correlation between this extensor moment and the peak braking force and duration of the braking phase suggest that it served to reduce the amount of braking experienced during early stance, a key characteristic of efficient race walking. Race walkers are thus recommended to develop a forceful extension of the hip during these gait phases with exercises that position the hip and knee in a manner which replicates race walking movements as normal athletic exercises might lack the specificity required for this unique form of gait.

The defined role of the knee in race walking had a profound effect on all lower limb joints. Because the knee was straightened from initial contact to midstance (and tended to be hyperextended), it could not extend during late stance to propel the centre of mass forwards (as in running) but instead underwent flexion. The rate of knee flexion prior to toe-off was controlled by an energy absorbing knee extensor moment which allowed the foot to maintain ground contact for longer. This not only increased the foot behind distance but also provided the swing leg with more time to advance so that the foot ahead distance was increased before any flight occurred. The contribution of this knee extensor moment to stride length is noteworthy and may be one of the discrete factors that enable fast race walking while minimising flight time. The hyperextension of the knee resulted not only in little energy generation, but also in prolonged energy absorption by the ankle plantarflexor muscles during

midstance and early hyperextension and periods of energy absorption at the hip. As a result, the race walkers in this study employed the straightened leg as a long rigid lever for most of the stance phase prior to toe-off. This meant that the largest contributor to the generation of energy was the coupling of the hip extensors (during late swing and early stance) and the ankle plantarflexors (during late stance). This coupling resulted from a proximal-to-distal passive energy transfer from the hip to the ankle via the straightened knee, and aided the relatively small triceps surae muscles in producing the large final plantarflexor moment.

In Table 2, three of the joint moment peaks and two of the power peaks were correlated with speed, and all largely occurred during swing. The majority of these were at the knee, with one at the hip. There were no correlations between ankle moments and powers with any key performance variables. These five peaks were also correlated with cadence, but only one moment (annotated as K1 in Figure 2) was correlated with stride length as it began during late stance. Leg joint kinetics during swing were therefore important as they increased cadence through the rapid forward movement of the midswing leg which was a key component in maintaining whole body forward momentum given the limited contribution of the contralateral midstance leg to energy generation. While the contribution of the stance leg is still important, it appears from our data that the fast movement of the swing leg is a defining feature of better performances. In addition to the swing leg's role in performance, the restraint on knee flexion during midswing could be a crucial intervention in facilitating full knee extension prior to initial contact. However, the large magnitude of power (6.3 ± 1.0 W/kg) required for deceleration of the lower leg which occurred later in swing was by far the largest energy absorbing power recorded. Similar to fast running (Chumanov, Heiderscheit & Thelen, 2011), this may be a cause of the frequent injuries to the hamstrings in race walkers (Francis et al., 1998) given that at the same time this biarticular muscle group contributed to a

large hip extensor moment. Correct stance leg technique is obviously important, but race walkers should be attentive to the development of swing leg technique, and to the risk of injury inherent in its rapid movements.

The results from our study provide the most in-depth description of EMG activity patterns in elite race walking. Of the muscles analysed, the gluteus maximus and biceps femoris were active from midswing to prevent too great a hip flexion angle prior to heel strike. These synergist muscles continued as powerful hip extensors from late swing until midstance. The prominence of the biarticular hamstrings was also notable as they were involved in decelerating knee extension during late swing. On the anterior thigh, vastus lateralis activity was evident before, during and after initial contact which suggested that suitably trained quadriceps muscles are required for achieving and maintaining knee extension. However, the biarticular rectus femoris was less involved in knee extension during this phase, and more as a hip flexor. In the lower leg, the triceps surae muscles were almost silent during swing but contracted during contact causing the large plantarflexor moment in late stance which coupled with the early stance contraction of the hip extensors as the primary sources of energy generation. In building on our results, future studies which combine analysis of EMG with frontal and transverse plane movements are recommended.

CONCLUSIONS

Race walking is a highly technical event whose rules have a distinct effect on lower limb joint kinetics and EMG. By conducting a unique study combining muscle moments and powers with EMG in elite athletes, we have provided evidence for the role of particular leg muscles and associated joint movements. The straightened knee rule restricted the leg to the role of a rigid lever for much of the stance phase, and the event's rules influenced a swing

pattern which was crucial in generating forward momentum. The swing leg moments and powers were important contributors to race walking speed because they negated the limited contribution of the leg during midstance. In addition, hip extensor moments during early stance coupled with an ankle plantarflexor moment during late stance were key contributors to energy generation. Race walk coaches are advised in particular to employ training regimens which develop the strength of the key muscle groups, such as the hamstrings and hip flexors, to forcefully move the lower limbs and reduce the risk of injury during the swing phase.

REFERENCES

- Alkjaer, T., Simonsen, E.B. & Dyhre-Poulsen, P. (2001). Comparison of inverse dynamics calculated by two- and three-dimensional modules during walking. *Gait and Posture*, 13, 73-77.
- Bezodis, I.N., Kerwin, D.G. & Salo, A.I.T. (2008). Lower limb mechanics during the support phase of maximum-velocity sprint running. *Medicine and Science in Sports and Exercise*, 40, 707-715.
- Cairns, M.A., Burdette, R.G., Pisciotta, J.C. & Simon, S.R. (1986). A biomechanical analysis of racewalking gait. *Medicine and Science in Sports and Exercise*, 18, 446-453.
- Chumanov, E.S., Heiderscheit, B.C. & Thelen D.G. (2011). Hamstring musculotendon dynamics during stance and swing phases of high-speed running. *Medicine and Science in Sports and Exercise*, 43, 525-532.
- Clarys, J.P. & Cabri, J. (1993). Electromyography and the study of sports movements. *Journal of Sports Sciences*, 11, 379-448.
- de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*, 29, 1223-1230.
- Donà, G., Preatoni, E., Cobelli, C., Rodano, R. & Harrison, A.J. (2009). Application of functional principal component analysis in race walking: an emerging methodology. *Sports Biomechanics*, 8, 284-301.

Francis, P.R., Richman, N.M. & Patterson, P. (1998). Injuries in the sport of racewalking. *Journal of Athletic Training*, 33, 122-129.

Giakas, G. & Baltzopoulos, V. (1997a) A comparison of automatic filtering techniques applied to biomechanical walking data. *Journal of Biomechanics*, 30, 847-850.

Giakas, G. & Baltzopoulos, V. (1997b). Optimal digital filtering requires a different cut-off frequency strategy for the determination of the higher derivatives. *Journal of Biomechanics*, 30, 851-855.

Hanley, B., Bissas, A. & Drake, A. (2011). Kinematic characteristics of elite men's and women's 20 km race walking and their variation during the race. *Sports Biomechanics*, 10, 110-124.

Hoga, K., Ae, M., Enomoto, Y. & Fujii, N. (2003). Mechanical energy flow in the recovery leg of elite race walkers. *Sports Biomechanics*, 2, 1-13.

Hoga, K., Ae, M., Enomoto, Y., Yokozawa, T. & Fujii, N. (2006). Joint torque and mechanical energy flow in the support legs of skilled race walkers. *Sports Biomechanics*, 5, 167-182.

IAAF (2011). Competition rules 2012-13. Monte Carlo, Monaco: IAAF.

Leskinen, A., Häkkinen, K., Virravirta, M., Isolehto, J. & Kyröläinen, H. (2009). Comparison of running kinematics between elite and national-standard 1500-m runners. *Sports Biomechanics*, 8, 1-9.

Murray, M.P., Guten, G.N., Mollinger, L.A. & Gardner, G.M. (1983). Kinematic and electromyographic patterns of Olympic race walkers. *The American Journal of Sports Medicine*, 11, 68-74.

Okamoto, T., Tsutsumi, H., Goto, Y. & Andrew, P.D. (1987). A simple procedure to attenuate artifacts in surface electrode recordings by painlessly lowering skin impedance. *Electromyography and Clinical Neurophysiology*, 27, 173-176.

Sanzén, L., Forsberg, A. & Westlin, N. (1986). Anterior tibial compartment pressure during race walking. *The American Journal of Sports Medicine*, 14, 136-138.

Smith, G. (1989). Padding point extrapolation techniques for the Butterworth digital filter. *Journal of Biomechanics*, 22, 967-971.

White, S.C. & Winter, D. (1985). Mechanical power analysis of the lower limb musculature in race walking. *International Journal of Sport Biomechanics*, 1, 15-24.

Winter, D.A. (2005). *Biomechanics and motor control of human movement* (3rd ed.). Hoboken, NJ: John Wiley & Sons.

Zajac, F.E., Neptune, R.R. & Kautz, S.A. (2003). Biomechanics and muscle coordination of human walking: Part II: Lessons from dynamical situations and clinical implications. *Gait and Posture*, 17, 1-17.