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Rollover footwear affects lower limb biomechanics during walking

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ABSTRACT

Aim: To investigate the effect of rollover footwear on walking speed, metabolic cost of gait, lower limb kinematics, kinetics, EMG muscle activity and plantar pressure.
Methods: Twenty subjects (mean age-33.1 years, height-1.71 m, body mass-68.9 kg, BMI 23.6, 12 male) walked in: a flat control footwear; a flat control footwear weighted to match the mass of a rollover shoe; a rollover shoe; MBT footwear. Data relating to metabolic energy and temporal aspects of gait were collected during 6 min of continuous walking, all other data in a gait laboratory.
Results: The rollover footwear moved the contact point under the shoe anteriorly during early stance, increasing midfoot pressures. This changed internal ankle dorsiflexion moments to plantarflexion moments earlier, reducing ankle plantarflexion and tibialis anterior activity after initial contact, and increasing calf EMG activity. In mid stance the rollover footwear resulted in a more dorsiflexed ankle position but less ankle movement. During propulsion, the rollover footwear reduced peak ankle dorsiflexion, peak internal plantarflexor ankle moments and the range of ankle plantarflexion. Vertical ground reaction loading rates were increased by the rollover footwear. There were no effects on temporal or energy cost of gait and no effect of elevated shoe weight.
Conclusion: Investigating all proposed effects of this footwear concurrently has enabled a more valid investigation of how the footwear effects are interrelated. There were concurrent changes in several aspects of lower limb function, with greatest effects at the foot and ankle, but no change in the metabolic cost of walking.

1. Introduction

Rollover footwear is popular due to the proposed benefits to gait, posture and altered muscle activity. The underlying hypothesis is that the sole influences how load is distributed under the footwear, thus affecting load applied to the foot, external lower limb moments, the response of muscles and resultant joint motion. It is thought that these changes affect the metabolic cost of walking and there are thus varied and interrelated effects. To date, however, the literature has generally focussed on how separate aspects of gait are affected e.g. kinematics and kinetics [1–4], or plantar pressure [5,6], or EMG [7,8], or energy cost [9,10], but not how changes in one data might relate to changes in other data. Thus, it is not clear what kinematic or EMG changes occurred when changes in metabolic cost of walking were observed [9,10]. Also, shoe design has not been adequately controlled, with differences in shoe weight, last geometry, and upper material meaning that the curved sole profile has not been the independent variable in previous studies.

Investigating all proposed effects in the same subjects and shoes would enable more valid investigation of whether the proposed footwear effects are interrelated. For example, Demura et al. [1], Myers et al. [2], Nigg et al. [3], Taniguchi et al. [4] and Romkes et al. [7] reported loss of normal ankle plantar flexion in early stance with MBT footwear (Masai Barefoot Technology, characterised by soft heel and a rounded shoe sole in the anterior–posterior direction). Perhaps this is due to the curved rearfoot profile shifting the centre of pressure (COP) and ground reaction force anterior to the ankle. Furthermore, Myers et al. [2], Romkes et al. [7] and Wu et al. [11] all reported reduced ankle and forefoot dorsiflexion during mid stance, perhaps a further consequence of the sole changing the COP position and thus midfoot and ankle moments. However, despite it being fundamental to any effect on gait and directly related to sole shape, changes in the contact point under the footwear have not been reported. Likewise changes in plantar loading are too inconsistent [5,6] to explain how sole profile affects loading of the foot. Furthermore, ankle plantarflexion is part of the lower limb shock absorption mechanism, and we hypothesise that reducing plantarflexion might lead to the

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elevated loading rates reported [8]. Elevated loading rates and altered ankle position might affect external knee and hip moments and kinematics, and flexion of either could negate the loss of shock absorption at the ankle. Greater knee flexion has been reported [4,7] but not hip flexion [2,7]. The issue of how changes in loading under the footwear and foot due to the sole curvature affect joint moments and motions and loading rates at initial contact is thus unresolved.

We further hypothesise that from around 50% of stance the upward curvature of the footwear sole should enable the body centre of mass to "rollover" the base of support, explaining the reduced foot and ankle dorsiflexion reported [3,7,11]. The associated calf muscle activity, which we assume arrests ankle dorsiflexion, might be decreased when it is approaching its peak. This might change vertical and posterior ground reaction forces, the efficiency of walking and potentially walking speed. Decreases in peak calf muscle activity would be contrary to manufacturer claims that this footwear exercises leg muscles.

There are both positive [10] and negative [9] reports that rollover footwear increase the energy cost of ambulation. However, previous studies have controlled walking speed, a strong determinant of metabolic cost, and perhaps removed the real world effects of the footwear. Furthermore, the elevated footwear mass (e.g. twice normal) and changes in kinematics and ground reaction forces might themselves change walking speed. Investigating any change in walking speed due to this rollover style footwear seems fundamental to understanding its real life effect. Based on the belief that the changes in kinematics and small increases in foot mass are not sufficiently large perturbations to affect walking speed and energy cost, we hypothesise that there is no change in walking speed nor energy cost in rollover footwear.

The aim was therefore to compare kinematic, kinetic, plantar pressure, contact point under the shoe, EMG and metabolic cost of walking data between rollover and flat footwear.

Fig. 1. Control footwear (top) (300 g), weighted control footwear (453 g), rollover shoe (453 g), and MBT (bottom) (526 g).

2. Methods

To test our hypotheses and address issues related to control of footwear design not considered previously, walking in a rollover shoe (the curved sole was from Scholl STARLIT shoes) (Fig. 1) was compared to walking in (1) a flat control shoe made from the same leather upper and last as the rollover footwear, and (2) the flat control footwear weighted to the equivalent of the rollover footwear. To relate our results to prior literature we also investigated MBT footwear (Chapa, textile upper). All insoles were removed and replaced with a 1.2 mm poron insole to eliminate any effect variation in the insole might have on shoe fitting. Data describing the loading of the footwear against the floor (force plate) and the foot against the shoe sole (plantar pressure), was combined with lower limb kinematics, kinetics, EMG muscle activity and measures of walking speed and the metabolic cost of walking in each of the four footwear conditions.

Subsequent to ethical approval/consent 20 healthy subjects (12 male) were recruited through convenience sampling from the University population (age 33.1 years $\pm$ 8.4, height 1.71 m $\pm$ 0.04 m, body mass 68.9 kg $\pm$ 12.1, BMI 23.6 $\pm$ 4.1). All were self reported free from diabetes, systematic musculoskeletal disease and injury (last 6 months).

Metabolic energy data and temporal aspects of gait were measured during 6 min of continuous walking at a self selected speed in a 82.9 m corridor around a quadrangle. Heart rate and metabolic energy cost were measured using a Metamax portable gas analyser (Cortex Biophysik, Leipzig, Germany). Walking speed was the distance covered during 6 min. The number of steps was based on the number of steps taken over a 10 m section of the corridor on four occasions during the 6 min, factored to the total distance walked. Subjects were allowed 5 min rest between footwear conditions and the order of footwear tested randomised.

All other data were collected whilst walking a 10 m course at a preferred speed and 10 stance phases recorded. Kinematic data were collected using 12 cameras (100 Hz) (Qualisys, Sweden) and rigid clusters of four reflective technical markers positioned on the leg, thigh and posterior pelvis. The position of footwear markers (above first, second and fifth metatarso-phalangeal heads, and posterior calcaneus) was standardised across footwear conditions. A relaxed standing trial in the control footwear was used to identify anatomical markers (malleoli, femoral epicondyles, greater trochanters, anterior and posterior superior iliac spines) to establish a suitable model of the limb (CAST technique [12]) and define 0°. Kinetic data was collected using two AMTI force plates (1000 Hz) (AMTI, Watertown, MA, USA).

EMG was recorded (3000 Hz) using a wireless 8-channel system (Noraxon, USA, Inc.) filtering signals between 10 Hz (high-pass) and 500 Hz (low-pass). Electrodes were placed on soleus, medial gastrocnemius, tibialis anterior, lateral bicep femoris, rectus femoris, gluteus maximus and erector spinae (right side only) (reference: medial femoral condyle). Electrode placement and skin preparation was according to SENIAM [13] and tape/light bandaging reduced movement artefact. Plantar pressure data was collected at 60 Hz with a Medilogic wireless in shoe pressure measuring system (T&T Medilogic, Berlin, Germany). All data was collected within a $<$1 h data collection session.

2.1. Data processing and analysis

Heart rate (beats/min) and energy per unit of distance travelled (kcal/m) were derived in Metamax software and Excel. Joint angles and internal joint moments were derived using Visual 3D (C-Motion, USA). A 4th order Butterworth low-pass filter (6 Hz) was applied to marker trajectories. A four segment model (foot, shank, thigh and pelvis) was defined according to the position of anatomical markers during the static trial. For shank, thigh and pelvis segments, the vertical z-axis was the line joining midway between medial and lateral anatomical markers on the distal and proximal parts of each segment. The x-axis was orthogonal to z-axis and lay in the frontal plane of the segment defined by z-axis and medial and lateral anatomical markers on the distal part of each segment (y-axis was perpendicular to x- and z-). The foot longitudinal axis, y-axis, was a line joining midway between the malleoli and the second metatarsal head, projected onto the global transverse plane. The *x*-axis was in the transverse plane and orthogonal to *y*-axis, and *z* perpendicular to both. Joint rotations were calculated using the joint co-ordinate system convention and Euler sequence: sagittal; frontal and; transverse plane. Like others [3,4] we focused on sagittal plane parameters because this is the primary plane in which the sole curvature is intended to influence gait. Based on prior reports, and because they are characteristic of sagittal plane function of each joint, the angle at initial contact, and peaks in dorsiflexion/plantarflexion were derived for the ankle, and peaks in flexion/extension for the knee and hip.

Joint moment data were calculated in Visual 3D, normalised to body mass and the peak and integral of +ve and −ve internal moments derived. Moment integrals might better reflect total muscle effort than peak moment values. Both peaks of the vertical and anterior–posterior ground reaction force were derived to investigate changes in impact loading due to expected changes in joint motion, and altered propulsive forces due to potential changes in muscle activity. Average vertical loading rate was calculated by taking the average of *first derivative* of the force trace between initial contact and the first peak in vertical force. The anterior/posterior position of the COP under the footwear (derived via the force plate) relative to the anterior/posterior position of the heel marker (in the anterior/posterior axis of the global system) was also derived. This quantifies change in the contact between footwear and floor due to the curved sole profile.

EMG signals were rectified and low-pass filtered (10 Hz) to derive the linear envelope of the signal. Peak EMG and integral of the signal were derived. The integral might better reflect total muscle activity since changes in ankle kinematics might lead to altered activity in periods other than those when peak EMG values occur.

Plantar pressure data was divided into nine areas using custom software (Matlab 7.03). Anterior-posteriorly, the rearfoot was 31%, the midfoot 19% and forefoot 50% of foot length [14]. The forefoot was divided into metatarsal and toe areas by visual inspection. The medial–lateral division of rearfoot and midfoot was based on a line between the middle of the heel and second metatarsal. The metatarsal area was divided into medial (35%), central (30%) and lateral regions (35% of forefoot width) [15]. The toe area was divided manually into hallux and lesser toes. Peak and average pressure were calculated for each region.

Parameters of interest were derived from individual trials and averaged for each limb. Shapiro–Wilk was employed which confirmed normal distribution in all cases ($p > 0.05$) except for EMG data ($p < 0.05$). Repeated-measure ANOVA were used to evaluate differences ($p < 0.05$) between footwear conditions with Bonferroni adjustment for multiple comparisons. Nonparametric equivalents (Friedman and Wilcoxon respectively) were employed for EMG data (SPSS version 16.0). To evaluate the reliability of the new parameter describing COP position in relation to the heel marker, three points of interest were identified for each single stance phase (initial contact, 50% of stance and at toe off) and intra-class correlation coefficients (ICCs) calculated.

3. Results

For the rollover shoe the COP under the shoe was more anterior compared to the control footwear for the first 40% of stance, but posterior in the last 30%. For the MBT the COP was almost 5 cm anterior of the heel marker at initial contact, compared to 2.5 cm for the control footwear (Fig. 2). The ICC for the distance between the COP and the heel marker, ICCs were between 0.52 and 0.74 at initial contact and between 0.77 and 0.95 at 50% of stance phase and at toe off in four footwear conditions. The rollover and MBT reduced peak pressure under the heel and forefoot compared to the control footwear (Fig. 3) and increased pressure under the mid foot ($p < 0.05$).

Both the rollover shoe and MBT reduced the range of ankle plantarflexion after initial contact compared to the control and weighted footwear ($p < 0.001$) (Table 1 and Fig. 4). The MBT placed the ankle in a more dorsiflexed position (position was $>0^{\circ}$) earlier, $\sim$17% in the mean data (compared to rollover shoe = 31% and control footwear = 34%). The range of ankle dorsiflexion in mid

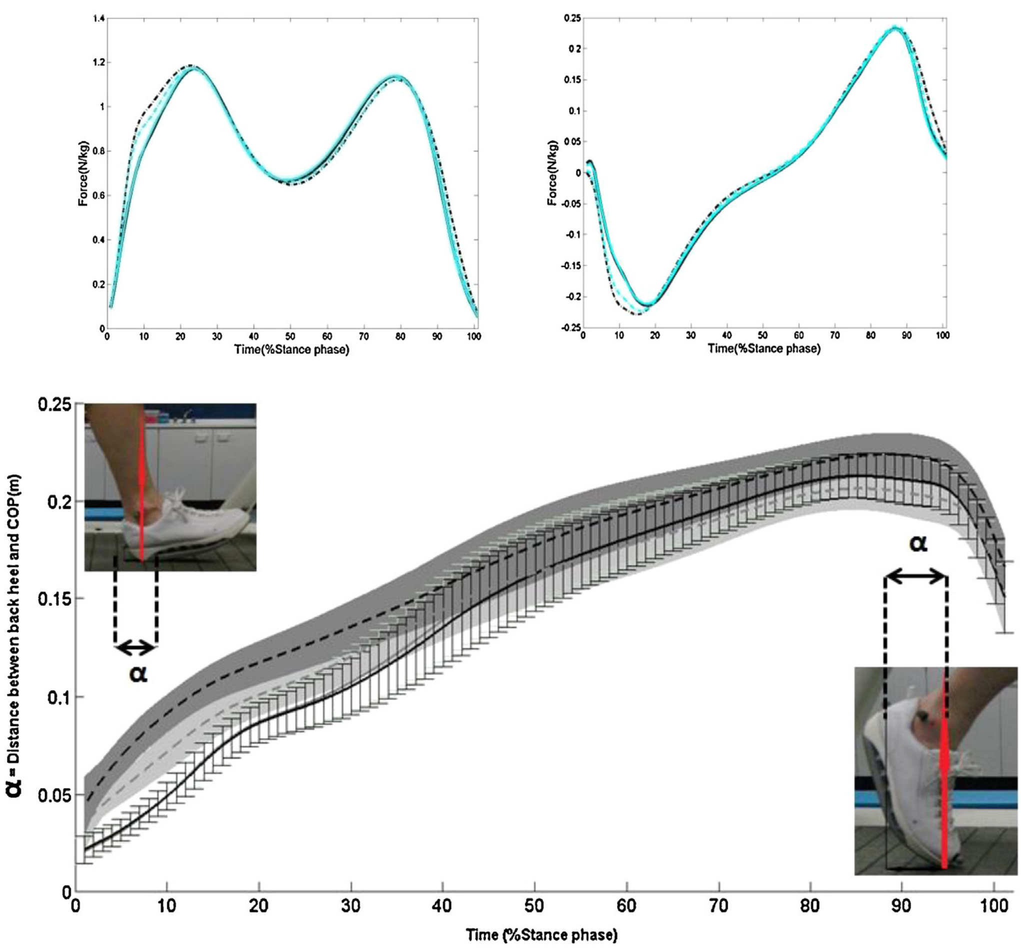

Fig. 2. Vertical (top-left) and anterior/posterior (top-right) ground reaction force data. Bottom: the anterior/posterior distance (α) between the marker on the heel and the anterior/posterior centre of pressure location (location of centre of pressure marked by vertical arrow for illustration purposes). Black dashed = mean MBT ± 1SD (dark grey band); grey dashed = Mean rollover shoe ± 1SD (light grey band), black solid = mean control ± 1SD (black error bar), grey solid = mean weighted footwear ± 1SD (grey error bar).

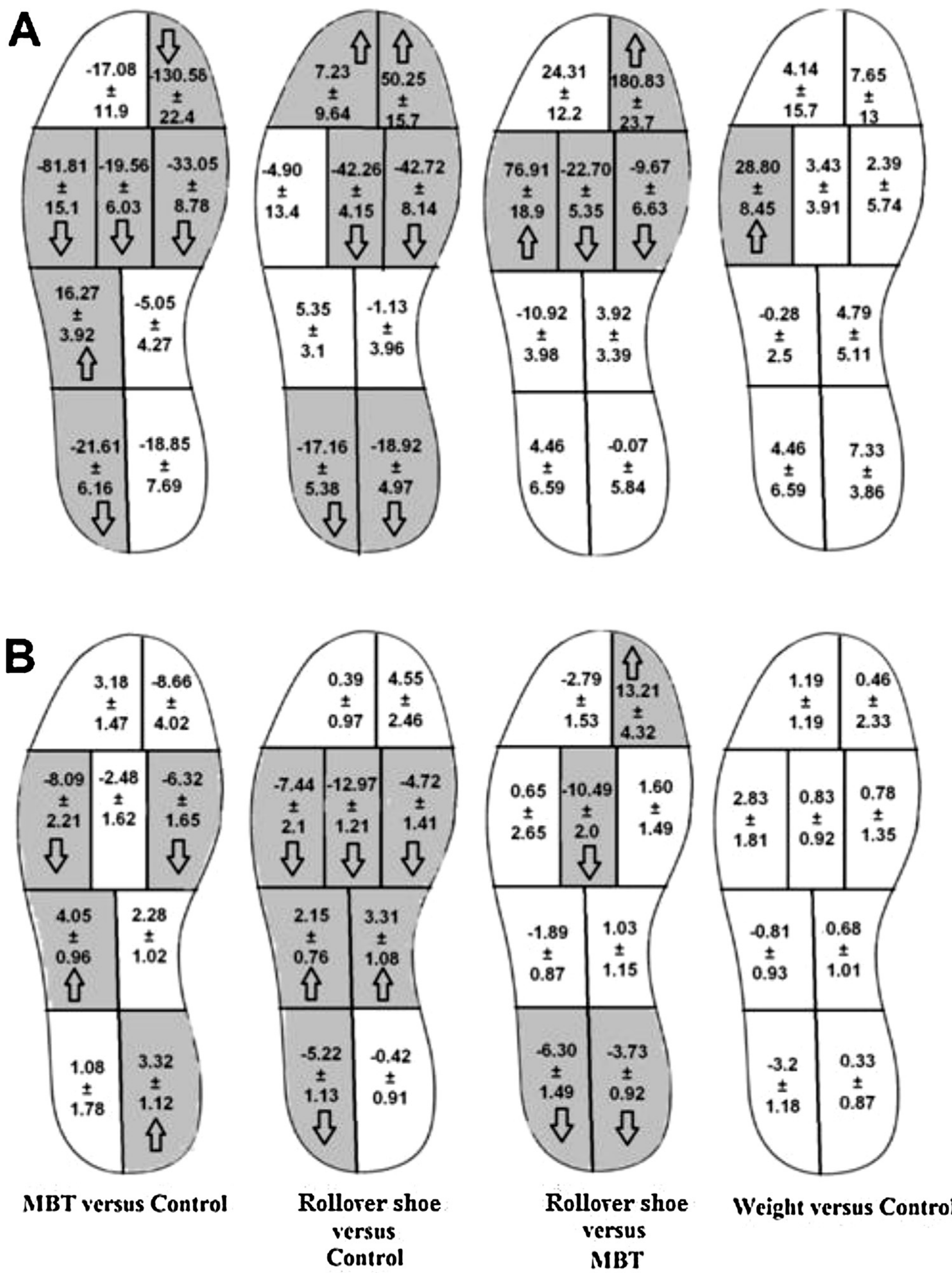

Fig. 3. Mean change (and SD. Err) in peak (A) and average (B) plantar pressure between the footwear conditions (KPa). Grey areas denote differences are at $p < 0.05$.

stance was significantly less than for the control and weighted footwear ($p < 0.001$). The ankle plantarflexion range of motion was statistically significantly decreased during propulsive phase by MBT and rollover footwear ($p < 0.001$), as was the total range of ankle motion.

The ankle experienced reduced peak dorsiflexor muscle moment with MBT (−65%) and the rollover shoe (−38%) compared to the control and weighted footwear ($p < 0.001$) (Table 2 and Fig. 4). The integral of the dorsiflexor muscle moment was reduced significantly by 79% by the MBT and 52% by the rollover shoe footwear ($p < 0.001$). Both footwear generated earlier and greater plantarflexion muscle moments prior to 50% of stance and a plateau in the ankle plantar flexor muscle moment compared to the control footwear (30–50% of stance).

The knee was more flexed at initial contact ($p < 0.001$), at the first flexion peak and at the point of minimum flexion ($p < 0.002$) with the rollover and MBT footwear (Table 1 and Fig. 4). Total range of knee motion was reduced significantly by MBT and rollover footwear compared to the control footwear ($p < 0.001$). The maximum and integral of the internal knee flexor moment, and the max internal extensor moment, was reduced significantly by both rollover shoe and MBT footwear. The integral of the internal knee extensor moment was reduced significantly only by the MBT compared to the control footwear ($p < 0.02$) (Table 2 and Fig. 4).

At the hip, the total range of motion was reduced by the rollover shoe and MBT footwear compared to the control footwear ($p < 0.001$). MBT reduced maximum flexor muscle moment compared to control and weighted footwear ($p < 0.001$). The rollover shoe reduced the integral extensor moment compared to the weight footwear ($p < 0.03$) (Table 2 and Fig. 4).

The average loading rate was 14% higher for the MBT (8.61 N/kg s) compared to the control footwear (7.56 N/kg s) ($p < 0.01$).

Table 1
Sagittal plane kinematics (°) for the ankle, knee and hip. C = control, W = weighted footwear, ROS = rollover shoe, M = MBT.

	Control		Weight		ROS		MBT		$p < 0.05$
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Ankle									
Angle at initial contact	−0.1	3.6	−0.3	3.8	−0.4	3.9	1.2	4.5	ROS vs MBT, W vs MBT
1st peak in plantarflexion	−8.5	2.7	−8.9	3	−6.2	3.2	−2.1	3.6	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs ROS
Maximum dorsiflexion	9.7	2.8	9.3	3.2	6.1	3.4	7.6	4.1	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs ROS
2nd peak in plantarflexion	−12.7	4.6	−13.7	4.3	−14.1	5.1	−11.8	5.1	MBT vs ROS, W vs MBT, C vs ROS, W vs C
Total range of motion	22.6	2.6	23.1	2.6	20.3	2.5	19.4	2.3	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs ROS, W vs C
Knee									
Angle at initial contact	1.3	5.7	0.8	5.6	2.6	5.2	3.4	5.5	C vs MBT, W vs MBT, C vs ROS, W vs ROS
1st peak in flexion	17.9	6.3	18	6.1	18.5	5.8	18.6	5.9	
Maximum extension	2.7	3.8	2.7	4.0	3.7	3.8	4.4	4.1	C vs MBT, W vs MBT, C vs ROS, W vs ROS
2nd peak in flexion	43.7	4.8	44.1	5.0	42.0	4.5	40.8	4.9	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs ROS
Total range of motion	43.3	4.0	43.9	3.9	40.3	3.7	38.4	3.8	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs ROS
Hip									
Angle at initial contact	31.6	6.3	30.9	8.3	30.8	7.8	30.6	7.8	
Maximum extension	−8.4	5.1	−8.5	6.7	−8.0	6.9	−7.9	6.8	
Angle at toe off	1.3	5.4	1.3	6.8	1.0	6.8	0.4	6.9	
Total range of motion	40.3	3.8	39.7	3.7	39.4	3.4	38.8	3.6	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs C

The average loading rate for the rollover shoe (8.09 N/kg s) was also higher (+7%) though $p > 0.05$, and significantly lower than the MBT ($p < 0.008$).

The only statistically significant difference in maximum EMG values for the rollover shoe and MBT compared to the control footwear was a reduction in tibialis anterior activity at initial contact (Fig. 5) (29% for MBT, 22% for rollover shoe, $p < 0.05$). The integral of tibialis anterior EMG activity was reduced by 17% for both MBT and rollover footwear compared to the control footwear ($p < 0.05$). The integral of the EMG profile increased for soleus with both MBT and rollover footwear compared to the control footwear (+13% MBT, and +8%, respectively $p < 0.05$). The EMG integral was also increased for medial gastrocnemius (+8%) for MBT ($p < 0.05$) (rollover + 5.5%, though $p > 0.05$). There were small increases in EMG activity for medial gastrocnemius and soleus between 0 and 30% of stance, and for rectus femoris at 40–60% of stance in both MBT and rollover footwear.

There were no statistically significant differences between footwear conditions in the metabolic cost and temporal aspects of walking (Table 3). Walking in the laboratory was slightly faster than in the 6 min walk test but this was not statistically significant, and there were no differences between for any footwear condition in speed (Table 3). Across all data there were no notable differences between weighted and control footwear.

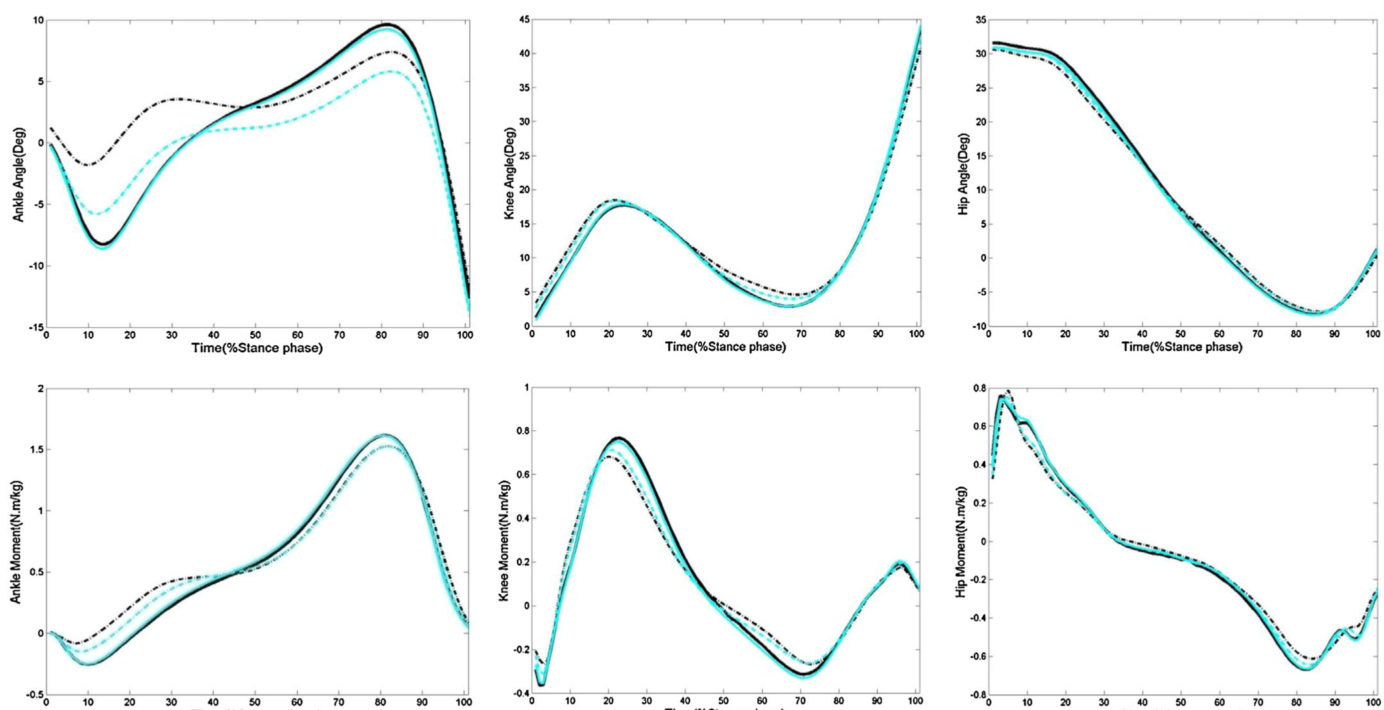

Fig. 4. Sagittal plane kinematics and kinetics (muscle moments) for ankle, knee and hip. Black dashed = MBT, grey dashed = rollover shoe, black solid = control, grey solid-weighted footwear. Positive angles are dorsiflexion, flexion and flexion for ankle, knee and hip, respectively. Positive muscle moments are plantarflexor, extensor and extensor for ankle, knee and hip, respectively.

Table 2
Sagittal plane joint muscle moments for the ankle, knee and hip. C = control, W = weighted footwear, ROS = rollover shoe, M = MBT. Moments = N m/kg, Integral of moment = N m %Stance/kg.

	Control		Weight		ROS		MBT		$p < 0.05$
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Ankle									
Max. dorsiflexor moment	−0.26	0.05	−0.26	0.04	−0.16	0.05	−0.09	0.05	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, w vs ROS
Max. plantarflexor moment	1.62	0.14	1.62	0.14	1.54	0.14	1.53	0.13	C vs MBT, W vs MBT, C vs ROS, W vs ROS
Integral dorsiflexor moment	−0.032	0.01	−0.031	0.01	−0.015	0.008	−0.007	0.004	C vs MBT, ROS vs MBT, W vs MBT, C vs ROS, W vs ROS
Integral plantarflexor moment	0.61	0.07	0.62	0.07	0.61	0.06	0.64	0.06	C vs MBT, ROS vs MBT
Knee									
Max. extensor moment	0.78	0.25	0.76	0.22	0.72	0.23	0.69	0.22	C vs MBT, W vs MBT, C vs ROS
Max. flexor moment	−0.32	0.13	−0.34	0.14	−0.27	0.15	−0.27	0.15	C vs MBT, W vs MBT, C vs ROS, W vs ROS
Integral extensor moment	0.20	0.07	0.193	0.07	0.189	0.08	0.184	0.08	C vs MBT
Integral flexor moment	−0.092	0.04	−0.098	0.05	−0.081	0.05	−0.076	0.05	C vs MBT, W vs MBT, W vs ROS
Hip									
Max. extensor moment	0.84	0.18	0.84	0.16	0.83	0.17	0.83	0.17	
Max. flexor moment	−0.69	0.16	−0.7	0.17	−0.67	0.15	−0.64	0.19	C vs MBT, W vs MBT, W vs C
Integral extensor moment	0.134	0.05	0.138	0.05	0.125	0.05	0.127	0.06	W vs ROS
Integral flexor moment	−0.222	0.07	−0.220	0.09	−0.214	0.08	−0.206	0.10	

4. Discussion

During the first 12% of stance phase the curved sole of the rollover shoe and MBT footwear moved the point of contact forward under the foot and pressure was transferred from the heel to the midfoot [5,16]. In accordance with our hypothesis this altered the external (and therefore internal in our data) ankle moments resulting in reduced ankle plantarflexion and reduced ankle dorsiflexor muscle activity after initial contact. This was synchronous with elevations in calf EMG activity at 10–20% stance which concurs with the altered internal ankle moments observed in early stance [3,7]. The more dorsiflexed ankle position might perhaps account for the slightly more flexed knee position in early stance (Fig. 4). Coactivity between tibialis anterior and medial gastrocnemius and soleus increased when wearing MBT and rollover footwear (10–25% stance). This concurs with the only comparable previous study [7] and might increase ankle stability and joint loading during early stance phase.

As hypothesised and consistent with Sacco et al. [8], the rate of lower limb loading after initial contact was increased wearing the rollover shoe and MBT footwear, probably due to the reduced ankle plantarflexion absorbing less energy. An increased loading rate combined with increased dorsiflexor and plantarflexor co-contraction could increase the load on ankle joint and suggests care should be advised when using rollover footwear in patients with ankle problems. The slightly increased flexion of the knee in early stance, and reduced knee extensor muscle moment, could be strategies for reducing this, although other strategies, such as hip flexion, were absent.

Concurring with others [2,3,5,7], in mid stance the COP moved forward under the shoe and pressure increased under the mid foot with the rollover and MBT footwear. There was a greater external dorsiflexor moment resulting in a more dorsiflexed ankle position. We assume that the reduced range of ankle dorsiflexion and a more dorsiflexed position in mid stance would result in a more inclined position of the shank [17]. This would result in the reduced internal knee extensor moment and the more flexed knee we observed. We assume the increased internal plantarflexor ankle moments, and associated elevation in calf muscle EMG signal pre 40% of stance, restrained the ankle dorsiflexion.

As hypothesised the rollover footwear and MBT reduced peak ankle dorsiflexion [7] and the range of ankle plantarflexion, but also the peak internal plantarflexor ankle moment. We assume this occurs because the curved profile of the footwear allows the body centre of mass to move forwards more easily, and reduces the demand for internal ankle plantarflexion moment and ankle power generation [4,18]. This might also contribute to the reduced forefoot pressure observed [5,15,19]. However, contrary to our hypothesis, only MBT displaced the COP anteriorly under the shoe in propulsion (Fig. 2). Furthermore, consistent with Romkes et al. [7], changes in ankle movement and moments were not associated with corresponding changes in calf muscle EMG data (Fig. 5). This is contrary to our assumption that such data are strongly coupled and can collectively explain the effects of rollover footwear.

The effect of the elevated weight in the rollover shoe was negligible. Hansen and Wang [9] also showed no difference in oxygen consumption when using footwear up to 650 g in weight. Whilst others have shown weight and energy cost of walking to correlate strongly, this has only been proven when additional weight was in excess of 0.9 kg, triple the weight of some shoes tested in our study [20]. We assume the weight of the MBT footwear (which was greater still) also had no effect. It is only 73 g more than the rollover shoe, and the difference between control and weighted footwear was twice this (153 g). Effects at the level of the whole body (energy, walking speed) were also negligible. Others have reported some differences in these data with MBT [4,7,9,10]. However, we believe that 6 min of uninterrupted walking at a self selected speed is a superior protocol compared to treadmill walking at a fixed speed since it reflects real world use of footwear, and is thus a better approach to measurement of energy expenditure [9,10]. We interpret this nil effect to mean that the body adapts to local changes (e.g. ankle motion) without a change in the overall effort and speed of ambulation.

There are three primary limitations to this work. First, we isolated the effect of the sole in the rollover shoe by using the same upper and last in the control and weighted footwear, and identifying that the effect of footwear weight was negligible. However, we did not entirely isolate the effect of the sole curvature, since use of different material thickness means that sole stiffness was also different between footwear. Second, we report on the immediate effects of these footwear, whereas training programmes are advocated and adaptation reported [21]. Third, interrelationships between data types (motion, pressure and so on) were not quantified statistically. Linear correlation analyses between parameters were avoided because the parameters we focused on only describe specific points in time whereas

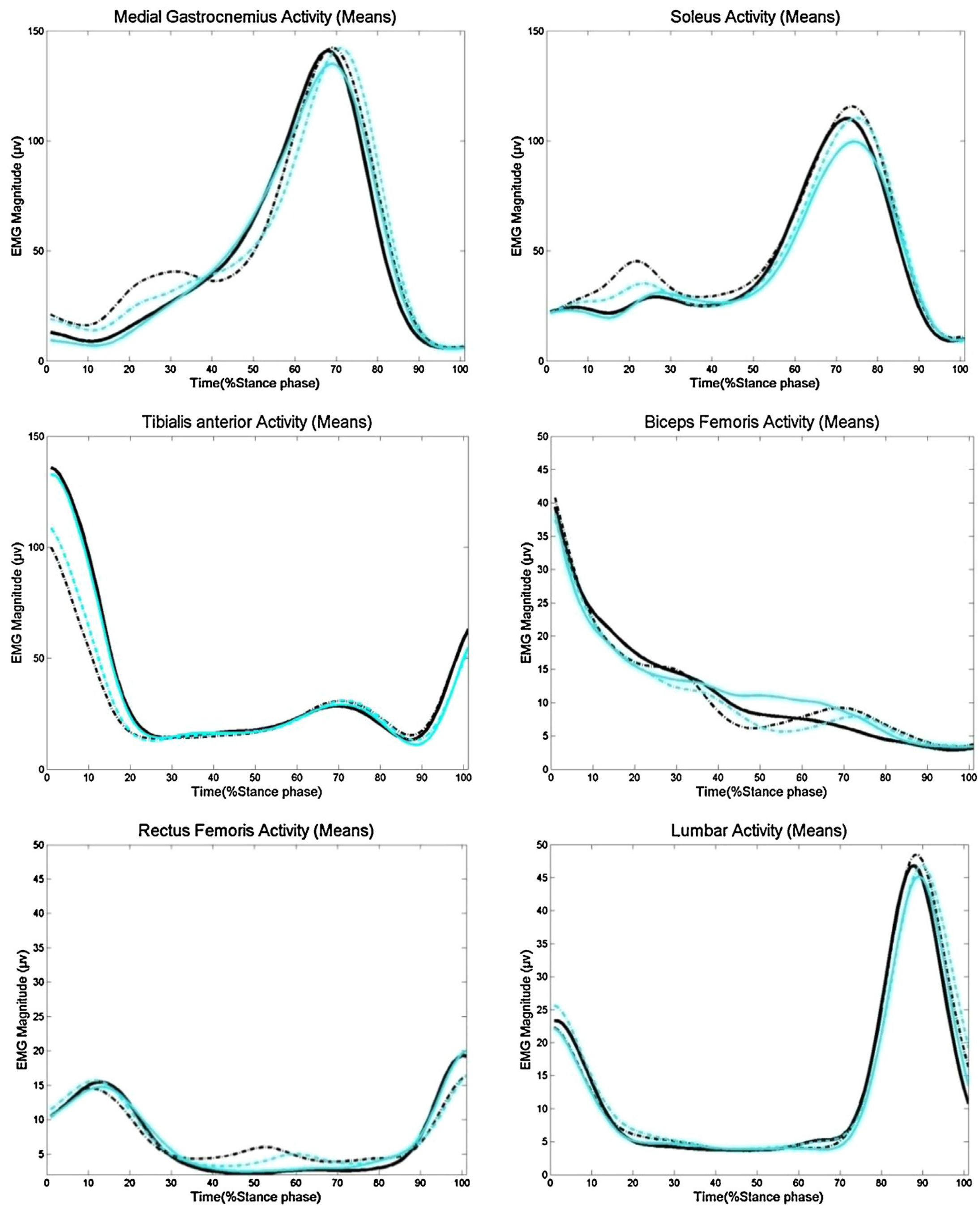

Fig. 5. Graphs show the level of muscle activity (milli volts) during the stance phases of gait for each of the four conditions. Black dashed = MBT, grey dashed = rollover shoe, black solid = control, grey solid-weighted footwear.

Table 3
walking speed during 10 m laboratory tests, and metabolic cost, walking speed, distance travelled and steps taken in each of the four footwear conditions during 6 min of continuous walking. All $p > 0.05$.

Condition	10 m test	6 min walk				
	Speed (m/s)	Speed (m/s)	Distance (m)	Steps taken (n)	Heart Rate (1/min)	Energy per metre (kcal/m)
Control	1.42 ± 0.10	1.37 ± 0.15	493.28 ± 54.87	664 ± 39.62	96 ± 13	0.0529 ± 0.011
Weighted	1.41 ± 0.08	1.38 ± 0.15	496.37 ± 52.10	670 ± 37.96	97 ± 13	0.0539 ± 0.012
Rollover shoe	1.42 ± 0.08	1.37 ± 0.14	492.35 ± 48.63	659 ± 37.30	97 ± 13	0.0540 ± 0.012
MBT	1.42 ± 0.08	1.40 ± 0.16	503.82 ± 58.14	661 ± 50.49	98 ± 13	0.0545 ± 0.011

interrelationships might exist between periods of time. Nonlinear analyses, such as artificial neural networks, may be more appropriate.

5. Summary

Investigating all proposed effects concurrently in the same subjects and using adequately controlled shoes has enabled a more valid investigation of how the effects of this footwear might be interrelated. In rollover footwear the COP was displayed anteriorly at initial contact with equivalent increases in midfoot plantar pressure. Changes in lower limb kinematics, moments and muscles activity were greatest at the ankle. Reduced contact phase plantarflexion was coupled with reduced tibialis anterior activity, and some evidence of elevated calf activity. Lower limb loading after initial contact was increased wearing the rollover footwear. Despite changes in kinematics and kinetics there was no change in the metabolic cost of walking and the effect of shoe weight was negligible.

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Conflict of interest

SSL International own the rights to the rollover footwear tested in this study (the sole unit is from the Scholl STARLIT range). Their staff worked with the authors to define the need for this study but took no part in study design, subject recruitment, data collection, analysis nor interpretation. SSL international made no contribution to the writing of this paper but have authorised its publication. None of the authors received financial benefit from the work in this paper.

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