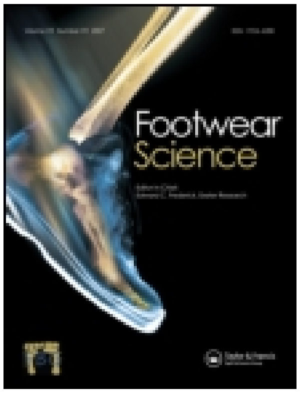

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Walking in an unstable Masai Barefoot Technology (MBT) shoe introduces kinematic and kinetic changes at the hip, knee and ankle before and after a 6-week accommodation period: a comprehensive analysis using principal component analysis (PCA)

Scott C. Landry [a] , Benno M. Nigg [b] & Karelia E. Tecante [b]

[a] School of Recreation Management and Kinesiology, Acadia University, Wolfville, Nova Scotia, Canada

[b] Faculty of Kinesiology, University of Calgary, Calgary, Alberta, Canada

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Walking in an unstable Masai Barefoot Technology (MBT) shoe introduces kinematic and kinetic changes at the hip, knee and ankle before and after a 6-week accommodation period: a comprehensive analysis using principal component analysis (PCA)

Scott C. Landry[a]*, Benno M. Nigg[b] and Karelia E. Tecante[b]

[a]*School of Recreation Management and Kinesiology, Acadia University, Wolfville, Nova Scotia, Canada;* [b]*Faculty of Kinesiology, University of Calgary, Calgary, Alberta, Canada*

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Background: Scientific and anecdotal evidence suggests that some individuals who wear the unstable Masai Barefoot Technology (MBT) shoe experience a reduction in back and joint pain. A more comprehensive biomechanical gait analysis is needed to better understand the mechanisms for symptom relief and what the long-term implications of wearing these shoes might be on the body.
Objective: The aim of this study was to determine the gait changes introduced at the hip, knee and ankle before and after wearing an unstable MBT shoe for a 6-week accommodation period.
Methods: Three-dimensional joint angles and moments were measured for 23 healthy individuals while walking in an unstable MBT shoe and a stable control shoe, both before and after a 6-week accommodation period of wearing the unstable shoe at their workplace. Principal component analysis (PCA) was used on the stance phase waveforms to identify differences between the two shoes and two testing sessions.
Results: Joint angle and moment differences between the two shoe conditions were identified both before and after wearing the unstable shoe for the accommodation period. Notable kinematic changes included reduced hip flexion–extension and ankle adduction–abduction range of motion, increased early stance dorsiflexion and increased knee internal rotation for unstable shoe walking. Ankle moments tended to be greater for the unstable shoe and at the hip and knee, both increases and decreases in moments were observed.
Conclusions: While many of the identified changes agree with previous research, this is the first study to report increases in some joint moments for the unstable MBT shoe. These increases along with other notable changes do, however, require further investigation to better understand the long-term implications of the unstable MBT shoe.

Keywords: Joint angles; joint moments; unstable MBT shoe; principal component analysis (PCA); lower extremity; musculoskeletal loading; walking

1. Introduction

Minimalist shoes and other shoes that have been designed to mimic a specific aspect of barefoot walking have become increasingly popular, particularly over the past 5 years. Originating in Switzerland in 1996, the unstable Masai Barefoot Technology (MBT) shoe can be considered the original 'barefoot' functional shoe, and of these functional shoes, the MBT has been the most extensively studied shoe in the peer-reviewed literature (Romkes *et al*. 2006, Nigg *et al*. 2006a, 2009, 2010, Stewart *et al*. 2007, Ramstrand *et al*. 2008, Boyer and Andriacchi 2009, Landry *et al*. 2010, Ramstrand *et al*. 2010, Stoggl *et al*. 2010, Buchecker *et al*. 2010). Scientific and anecdotal evidence, as well as personal testimonials, exist to support the notion that the unstable MBT shoe is effective at relieving joint pain and helping to reduce and/or manage lower limb or back injuries (Nigg *et al*. 2006b, 2009).

Four peer-reviewed studies have specifically addressed the kinematic and kinetic changes introduced by the unstable MBT shoe in comparison to a stable control shoe for walking (Romkes *et al*. 2006, Nigg *et al*. 2006a, Stoggl *et al*. 2010, Buchecker *et al*. 2010), with only one study making similar comparisons for running (Boyer and Andriacchi 2009). The walking comparative studies have all used relatively small sample sizes (8–12 subjects) and only two of these studies have addressed joint kinetic or moment differences (Nigg *et al*. 2006a, Buchecker *et al*. 2010). Nigg *et al*. (2006a) found no statistically significant differences in joint moment impulses for the hip, knee and ankle; however, trends related to a slight reduction in

*Corresponding author. Email: scott.landry@acadiau.ca

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hip and knee moments were reported. It was also noted that the MBT shoe showed a trend towards a more inverted ankle moment compared to the stable control shoe. Buchecker *et al.* (2010) compared lower extremity joint loadings in overweight males using MBT shoes and a stable control shoe and concluded that walking in the unstable MBT shoe reduced the first peak in the knee adduction moment. It was also reported that there were no statistical differences at the hip and ankle for other kinetic variables analysed.

Addressing the influence on joint kinematics, the unstable MBT shoe has its greatest effect at the ankle joint, with both Nigg *et al.* (2006a) and Romkes *et al.* (2006) reporting an increased dorsiflexion angle during the first half of stance. Nigg *et al.* (2006a) had their subjects walk at a preselected speed (5.0 ± 0.5 km/h) for both footwear conditions and found no other significant differences in lower extremity joint kinematics. Romkes *et al.* (2006), however, identified differences at the hip and knee, with walking in the unstable MBT shoe leading to smaller ranges of motion in the sagittal plane and smaller peak flexion angles at both the hip and knee, along with smaller peak hip extension angles in comparison to walking in the stable control shoe. A plausible explanation for the different findings at the hip and knee between the two studies could be because Romkes *et al.* (2006) had their subjects walk in the unstable MBT shoe and stable control shoe at self-selected speeds, leading to significantly reduced speeds for the unstable MBT shoe.

To better understand the long-term implications that unstable MBT footwear can have on the body and how these shoes may or may not help to alleviate different ailments, more comprehensive biomechanical analyses are required. The multivariate analysis technique of principal component analysis (PCA) has been used effectively to compare gait waveforms, particularly over the past decade (Deluzio *et al.* 1997, Landry *et al.* 2007a, b, McKean *et al.* 2007, Rutherford *et al.* 2008). The PCA technique was shown to have increased sensitivity over the more traditional parameter-based analysis method of capturing differences in biomechanical data (Wrigley *et al.* 2005). It is expected that the application of this highly sensitive PCA technique to a relatively large group of individuals walking in the unstable MBT shoe could lead to the identification of kinematic and kinetic features unique to the unstable shoe that may not have been previously reported in other studies using smaller sample sizes (Nigg *et al.* 2006a, Romkes *et al.* 2006, Stoggl *et al.* 2010, Buchecker *et al.* 2010).

The purpose of this study was to have a larger sample size of individuals (compared to the previous studies mentioned above) wear the unstable MBT shoe at their workplace for a 6-week accommodation period and determine whether immediate and more long-term differences in lower extremity joint angles (kinematics) and moments (kinetics) could be identified between the unstable MBT shoe and a more conventional stable control shoe. The following hypotheses were tested:

H1: The most pronounced kinematic and kinetic changes introduced by walking in the unstable MBT shoe will be at the ankle joint, with magnitudes being both reduced and increased for the unstable MBT shoe depending on the phase of stance and anatomical plane.

H2: Walking in the unstable MBT shoe will introduce less pronounced kinematic and kinetic changes at the hip and knee, with magnitudes being primarily reduced for the unstable MBT shoe.

H3: For the majority of the kinematic and kinetic gait measures, the immediate changes introduced by the unstable MBT shoe at the pre-accommodation testing session will continue to exist after the 6-week accommodation period.

2. Method

2.1. *Subjects*

Twenty-three subjects (16 females and seven males) participated in this study, with the mean age, height and mass being 52.4 ± 7.8 (SD) years, 165.2 ± 7.2 cm and 80.0 ± 15.4 kg, respectively. The study was approved by the University of Calgary's Office of Medical Bioethics and all subjects provided informed consent. Each of the 23 subjects had to meet the following inclusion criteria: (i) no previous experience wearing unstable shoes, (ii) able to wear the unstable shoe for a minimum of 30 h/week at their workplace, which involved mostly walking or standing, (iii) between 40 and 70 years of age, (iv) no lower extremity injury or major pain in the previous 6 months, (v) no previous major surgery to lower extremity or back, (vi) no evidence of arthritis, diabetes or neuromuscular condition, and (vii) no regular exercise routine for the lower legs.

2.2. *Testing protocol and analysis*

A comprehensive three-dimensional gait analysis was performed on all subjects, for two footwear conditions and for two visits spaced 6 weeks apart, at the Human Performance Laboratory (HPL) in the Faculty of Kinesiology at the University of Calgary (Calgary, Canada). For both testing visits, subjects had their gait measured while walking in an unstable MBT shoe and in a more conventional stable control shoe. The stable control shoe was considered to be the shoe that the subject normally wore at work prior to the study and the unstable MBT shoe (M. Walk model, Masai Barefoot Technologies, Switzerland) had a rounded sole in the anterior–posterior direction that created anterior–posterior instabilities and a soft cushioning heel pad that provided a degree of medial–lateral instability. During the initial visit to the laboratory and before making any measurements, subjects were instructed on how to properly stand and walk in the unstable MBT shoe and were then given approximately 10 min to get accustomed to the unstable nature of the shoe. After gait measurements were made for the first visit (pre-accommodation testing), subjects were asked to wear the unstable shoe for 1 h on the first day and gradually increase usage to 8 h/day by the end of the first week. After wearing the unstable MBT shoe for a 6-week accommodation period, which included standing/walking at their workplace for a minimum of 30 h/week (verified by a daily journal completed by the subjects), subjects returned for a second visit (post-accommodation testing) to have their gait measured in both footwear conditions.

The gait testing protocol was similar to that performed by Nigg *et al.* (2006a) on a smaller group of individuals ($n = 8$) wearing the unstable MBT shoe. The pelvis and lower limb segments on the left side of the body were tracked by placing 13 reflective markers on the pelvis, thigh, shank and foot. A standing neutral trial was collected with five additional reflective markers placed on the medial and lateral malleoli, medial and lateral epicondyles and greater trochanter to establish anatomical segment coordinate systems along with hip, knee and ankle joint centres.

Subjects performed five walking trials in the unstable MBT shoe and five trials in the stable control shoe, with the order being randomized between the two footwear conditions for both visits. Kinetic data were collected at 2400 Hz with a Kistler force platform (Kistler Instrumente AG, Winterthur, Switzerland) and kinematic data were collected at 240 Hz with an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA). Infrared timing gates were used to guide the subjects to walk at 1.5 ± 0.5 m/s and only the stance phase for a single stride for each trial was used for the analyses. The collected data were exported into Kintrak software (HPL, Calgary, Canada), with the kinematic and kinetic data being filtered using a zero-lag fourth-order low-pass Butterworth filter with cut-off frequencies of 12 and 50 Hz, respectively (Nigg *et al.* 2006a, 2010). For the hip, knee and ankle, three-dimensional joint angles were calculated using the joint coordinate system convention described by Grood and Suntay (1983) and internal joint moments, normalized to body mass, were calculated with inverse dynamics using the anthropometric, ground reaction force and motion data. Joint angle and moment waveforms were time normalized to percentage stance and the subject's five trials for each footwear condition were then ensemble averaged to obtain a representative waveform for each measure and for both visits.

PCA was applied to the ensemble average joint angle and moment waveform data. For each of the specific waveform measures (e.g. knee abduction angle), a separate PCA was performed by first arranging all the ensemble average waveforms for that particular gait measure [e.g. knee abduction angle waveforms from all subjects for both shoe conditions (stable and unstable MBT shoe) and both visits (pre- and post-accommodation testing)] into a data matrix (n waveforms $\times$ 101 waveform data points representing each percentage of stance). Described in detail by Deluzio *et al.* (1997), PCA takes the ensemble average waveforms for the specific analysis and subtracts the mean value at every instant of time for each of the included waveforms. The covariance matrix of the mean removed data set is calculated and an eignvector decomposition of the covariance matrix identifies principal patterns of variation that can be interpreted biomechanically. These patterns of variation or biomechanical waveform features generally include overall magnitudes, local magnitudes, phase shifts, amplitudes and other shape features. The biomechanical features are represented by the eigenvectors or PCs of the analysis and then each individual waveform receives a PC score based on how similar the individual waveforms are to the specific PC. The PC scores for each individual group (control shoe pre-testing, unstable MBT shoe pre-testing, control shoe post-testing and unstable MBT shoe post-testing) for a specific joint measure were compared statistically in SPSS version 18.0 (SPSS Inc., Chicago IL, USA) using a 2×2 two-way repeated-measures ANOVA to identify a shoe (stable control shoe versus unstable MBT shoe), visit (pre- versus post-accommodation testing) or interaction effect. An α level of $p < 0.05$ indicated a statistically significant difference and when an interaction effect was identified, Bonferroni-corrected pairwise *post-hoc* comparisons were made between the two footwear conditions and two visits.

3. Results

For the stance phase of the gait cycle, joint angle and net internal joint moment group mean ensemble average waveforms are shown for the hip (Figure 1), knee (Figure 2) and ankle (Figure 3) for both footwear

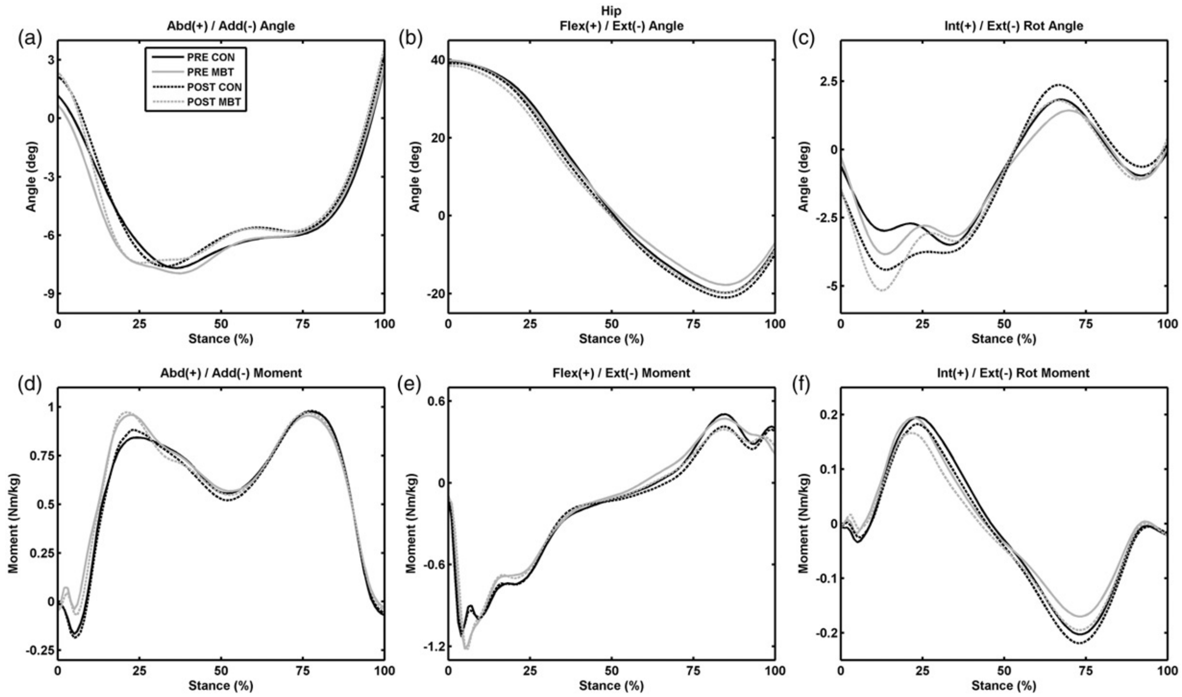

Figure 1. Mean hip joint angle and moment waveforms for the stance phase while walking in an unstable MBT shoe (MBT) and stable control shoe (CON) for both the pre- (PRE) and post-accommodation (POST) testing sessions. Joint angles are in the first row (A, B and C) and joint moments are in the second row (D, E and F). Frontal plane measures are in the first column (A and D), sagittal plane measures are in the second column (B and E) and transverse plane measures are in the third column (C and F).

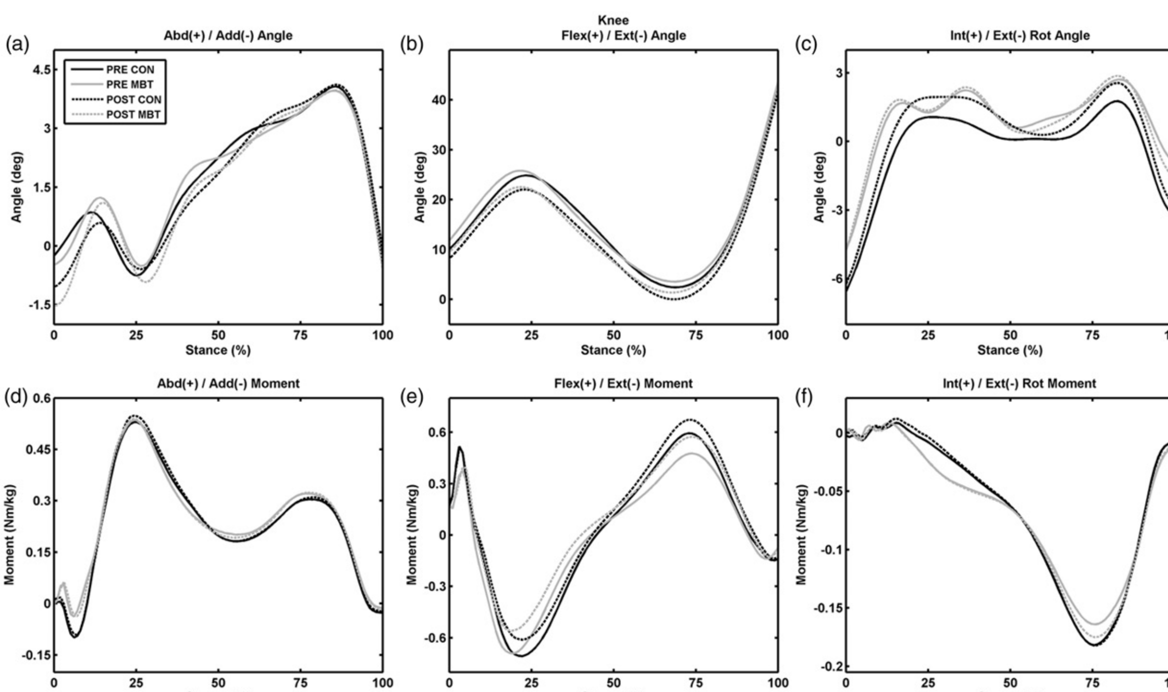

Figure 2. Mean knee joint angle and moment waveforms for the stance phase while walking in an unstable MBT shoe (MBT) and stable control shoe (CON) for both the pre- (PRE) and post-accommodation (POST) testing sessions. Joint angles are in the first row (A, B and C) and joint moments are in the second row (D, E and F). Frontal plane measures are in the first column (A and D), sagittal plane measures are in the second column (B and E) and transverse plane measures are in the third column (C and F).

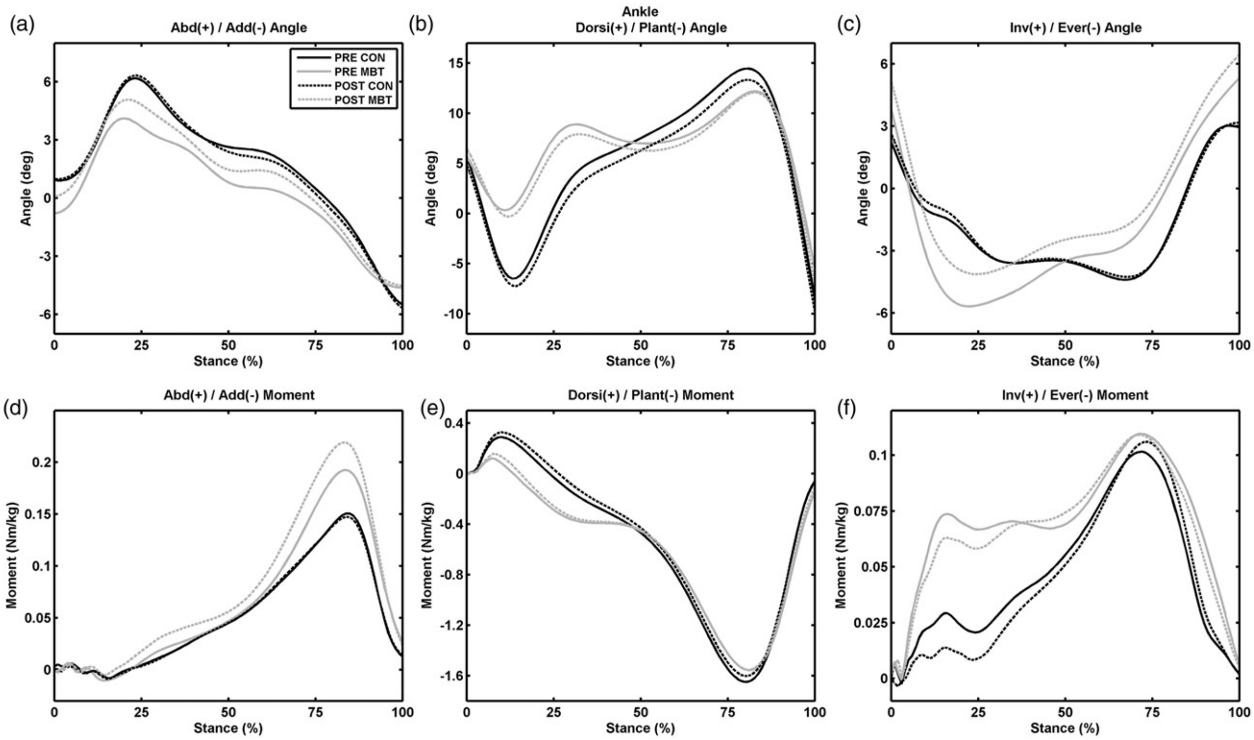

Figure 3. Mean ankle joint angle and moment waveforms for the stance phase while walking in an unstable MBT shoe (MBT) and stable control shoe (CON) for both the pre- (PRE) and post-accommodation (POST) testing sessions. Joint angles are in the first row (A, B and C) and joint moments are in the second row (D, E and F). Transverse plane measures are in the first column (A and D), sagittal plane measures are in the second column (B and E) and frontal plane measures are in the third column (C and F).

conditions (stable control shoe and unstable MBT shoe) and both visits (pre- and post-accommodation testing). The application of PCA was effective in describing the angle and moment waveforms, with the first three PCs for each waveform measure (18 in total) explaining a minimum of 92% of the total variance in the waveforms for all but the hip moment waveforms. Three PCs described 85.1, 84.9 and 77.3% of the waveform variability for the internal–external rotation, flexion–extension and abduction–adduction hip moments, respectively. Of the three PCs that were analysed for each of the 18 waveform measures (a total of 54 ANOVAs), there were 31 statistical differences captured between the stable control shoe and unstable MBT shoe (shoe effect), along with nine statistical differences between pre- and post-accommodation testing (Tables 1–3) and four interaction effects (Table 4). Eighteen statistical differences were identified at the ankle joint (Table 3) and 13 at both the knee (Table 2) and hip joint (Table 1). The biomechanical features identified by PCA by the first three PCs for the 18 waveform measures tended to be overall waveform magnitudes, local or overall phase shifts, and local or overall maximums/minimums. The first PC captured an overall magnitude that was generally across the entirety of stance for all the waveform measures. PC2 and PC3 captured local maximums, overall amplitudes and various phase shifts (local and overall). Figure 4 shows four examples of specific features that were captured by PCA. Figures 4A, 4E, 4I and 4M (aligned vertically in the first column) are all associated with capturing an overall magnitude difference for ankle abduction–adduction moment throughout stance. Figure 4A includes the ensemble average waveforms for the unstable MBT shoe (MBT) and the stable control shoe (CON) for both the pre- and post-accommodation testing sessions. Figure 4E is the first PC or loading vector that captures the overall abduction moment throughout the entirety of stance. Figure 4I includes an individual waveform with a high PC1 score and an individual waveform with a low PC1 score and these two waveforms were used along with the loading vector for interpreting the feature captured by PCA. Figure 4M is a plot of the mean PC1 group scores along with the standard error of the means for the unstable MBT shoe and the stable control shoe

Table 1. Description of hip waveform features captured by PCA and corresponding p values for shoe, visit and interaction comparisons.

Gait measure	PC	Shoe	Visit	Interaction	Feature captured
Angle					
Abduction–Adduction (Figure 1A)	1	0.31	0.34	0.61	Overall adduction angle magnitude throughout stance – no differences
	2	**0.016**	0.69	0.30	For both testing sessions, MBT had a greater adduction angle during early stance and reduced adduction angle during late stance compared to CON
	3	**0.032**	**0.028**	0.71	At foot strike, adduction angles comparable but MBT produced a more rapid rise towards peak adduction angle during first 40% of stance compared to CON More rapid rise towards peak adduction angle during the first 40% of stance observed for PRE compared to POST
Flexion–Extension (Figure 1B)	1	0.75	0.064	0.49	Overall flexion angle magnitude throughout majority of stance – no differences
	2	**<0.01**	0.64	0.99	CON had greater flexion–extension range of motion throughout stance compared to MBT
	3	0.085	0.28	0.91	Feature not interpreted biomechanically
Internal–External Rotation (Figure 1C)	1	0.61	0.37	0.93	Overall external rotation angle magnitude throughout stance – no differences
	2	0.82	0.20	0.78	Greater early stance external rotation angle and greater late stance internal rotation angle – no differences
	3	0.12	0.53	0.22	Feature not interpreted biomechanically
Moment					
Abduction–Adduction (Figure 1D)	1	**<0.01**	0.53	0.76	MBT had greater overall abduction moment throughout stance, particularly during the first 30% of stance compared to CON
	2	**<0.01**	0.31	**<0.01**	For PRE only, MBT had greater peak abduction moment during early stance and reduced peak abduction moment during late stance compared to CON
	3	**<0.01**	0.51	0.64	During first 20–30% of stance, CON produced a small peak adduction moment followed by larger peak abduction moment whereas MBT produced a small peak abduction moment followed by a much larger second peak abduction moment
Flexion–Extension (Figure 1E)	1	0.35	0.55	0.86	Overall extension moment magnitude throughout stance – no differences
	2	**<0.01**	0.14	0.85	MBT had greater peak extension moment during early stance and greater flexion moment during mid- to late stance compared to CON
	3	**<0.01**	**0.024**	0.72	MBT had a peak extension moment during the first 10% of stance that is delayed compared to CON POST had a peak extension moment during the first 10% of stance that is delayed compared to PRE
Internal–External Rotation (Figure 1F)	1	0.99	**<0.01**	0.20	Overall moment magnitude throughout stance shifted towards a more internally rotated moment for PRE compared to POST
	2	**<0.01**	0.39	0.82	CON had greater peak internal rotation moment during early stance and greater peak external rotation moment during late stance compared to MBT (amplitude difference)
	3	**<0.01**	0.42	0.64	Phase shift with MBT having peak internal rotation moment occurring earlier in stance compared to CON

2×2 repeated-measures ANOVA used on PC scores for each individual measure. Significant differences in bold ($p < 0.05$) with MBT being the unstable MBT shoe, CON the stable control shoe, PRE being the pre-accommodation testing session and POST the post-accommodation testing session. Supplementary Figure S1 (online) includes the group means and standard error of the means for each comparison.

Table 2. Description of knee waveform features captured by PCA and corresponding p values for shoe, visit and interaction comparisons.

Gait measure	PC	Shoe	Visit	Interaction	Feature captured
Angle					
Abduction–Adduction (Figure 2A)	1	1.00	0.63	0.61	Overall abduction angle magnitude throughout stance – no differences
	2	0.73	0.25	0.12	Feature not interpreted biomechanically
	3	0.32	0.70	0.17	Feature not interpreted biomechanically
Flexion–Extension (Figure 2B)	1	**0.024**	**<0.01**	0.89	MBT had greater overall flexion angle throughout majority of stance compared to CON PRE had greater overall flexion angle throughout majority of stance compared to POST
	2	**<0.01**	0.93	0.16	Captured phase shift with PC3 MBT had greater flexion angle compared to CON, particularly during mid-stance
	3	**<0.01**	0.71	0.055	MBT had greater flexion angle during early stance and reduced flexion angle during mid-stance compared to CON
Internal–External Rotation (Figure 2C)	1	**0.025**	0.45	0.30	MBT had greater overall internal rotation angle throughout stance compared to CON
	2	0.084	0.87	0.10	Feature not interpreted biomechanically
	3	**0.014**	0.72	0.65	MBT had bimodal internal rotation angle peaks between 10% and 35% of stance whereas CON has a single peak during the same period
Moment					
Abduction–Adduction (Figure 2D)	1	0.37	0.79	0.47	Overall abduction moment magnitude throughout stance – no differences
	2	**0.030**	0.72	0.62	CON had greater adduction moment during the first 10% of stance, greater abduction moment during early stance and reduced abduction moment during late stance compared to MBT
	3	**<0.01**	0.94	0.81	CON had greater adduction moment during early stance followed by greater abduction moment from early to mid-stance compared to MBT
Flexion–Extension (Figure 2E)	1	0.34	**<0.01**	0.21	PRE had overall moment magnitude shifted towards extension compared to POST for duration of stance
	2	**<0.01**	**<0.01**	0.98	CON had greater flexion moment during last half of stance compared to MBT POST had greater flexion moment during last half of stance compared to PRE
	3	**<0.01**	0.42	0.49	MBT had peak extension moment occurring sooner in early stance compared to CON
Internal–External Rotation (Figure 2F)	1	0.17	0.37	0.07	Overall external rotation moment magnitude throughout stance – no differences
	2	**<0.01**	0.28	0.56	MBT had greater external rotation moment during first half of stance and reduced peak external rotation moment during late stance compared to CON
	3	0.16	0.61	0.84	Feature not interpreted biomechanically

2×2 repeated-measures ANOVA used on PC scores for each individual measure. Significant differences in bold ($p < 0.05$) with MBT being the unstable MBT shoe, CON the stable control shoe, PRE being the pre-accommodation testing session and POST the post-accommodation testing session. Supplementary Figure S2 (online) includes the group means and standard error of the means for each comparison.

groups for both the pre- and post-accommodation testing sessions. The second column of Figure 4 describes a feature captured by PC2 for knee internal–external rotation moment, with the MBT shoe having a greater external rotation moment during early stance and a reduced external rotation moment during late stance compared to the stable control shoe. The third column of Figure 4 describes a difference in the range of motion between the two types of footwear captured by PC2 for hip flexion–extension angle throughout stance. Finally, the fourth column of Figure 4 describes a phase shift between the unstable MBT shoe and the stable control shoe captured by PC3 for hip internal–external rotation moment, with the

Table 3. Description of ankle waveform features captured by PCA and corresponding p values for shoe, visit and interaction comparisons.

Gait measure	PC	Shoe	Visit	Interaction	Featured captured
Angle					
Abduction–Adduction (Figure 3A)	1	0.058	0.55	0.34	Overall abduction angle magnitude throughout stance – no differences but trend for greater abduction angle for CON compared to MBT
	2	**<0.01**	0.16	0.58	CON had greater abduction angle during early stance and greater adduction angle during late stance compared to MBT
	3	**<0.01**	0.64	**0.031**	CON had a greater abduction–adduction range of motion throughout stance compared to MBT, for both testing sessions
Dorsiflexion–Plantarflexion (Figure 3B)	1	**<0.01**	0.059	0.53	MBT had greater dorsiflexion angle magnitude compared to CON, particularly during the first half of stance
	2	**<0.01**	0.58	0.13	MBT had greater dorsiflexion angle during early stance and reduced dorsiflexion angle during late stance compared to CON
	3	**<0.01**	0.50	0.45	Phase shift with CON experiencing peak dorsiflexion sooner than MBT during late stance
Inversion–Eversion (Figure 3C)	1	0.082	0.087	0.15	Overall eversion angle magnitude throughout stance – no differences
	2	**<0.01**	**0.049**	0.24	MBT had a greater eversion angle during early to mid-stance and a reduced eversion angle during mid- to late stance compared to CON. PRE also had a greater eversion angle during early to mid-stance compared to POST
	3	**<0.01**	0.10	0.080	MBT had a greater inversion–eversion range of motion throughout stance compared to CON
Moment					
Abduction–Adduction (Figure 3D)	1	**<0.01**	0.20	0.14	MBT had greater overall abduction moment magnitude throughout stance compared to CON, particularly during mid- to late stance
	2	0.24	0.089	0.24	Greater abduction moment during early stance, reduced abduction moment magnitude during mid-stance and greater abduction moment during late stance – no differences
	3	**0.042**	**0.042**	**0.017**	For POST only, MBT had greater abduction moment between 10% and 40% of stance compared to CON For MBT only, POST had greater abduction moment between 10% and 40% of stance compared to PRE
Dorsiflexion–Plantarflexion (Figure 3E)	1	**<0.01**	**0.046**	0.33	MBT had greater overall plantarflexion moment compared to CON, particularly during first 50% of stance PRE had greater overall plantarflexion moment compared to POST, particularly during first 50% of stance
	2	**<0.01**	0.26	**0.028**	CON had greater overall plantarflexion moment compared to MBT, particularly during 50–90% of stance, with difference being greater during PRE
	3	**0.013**	0.38	0.14	During the last half of stance, MBT had peak plantarflexion moment occurring later in stance compared to CON
Inversion–Eversion (Figure 3F)	1	**0.024**	0.85	0.94	MBT had greater overall inversion moment throughout stance compared to CON
	2	**<0.01**	0.29	0.39	MBT had greater inversion moment during first half of stance compared to CON
	3	**0.033**	**0.027**	0.11	Feature not interpreted biomechanically

2×2 repeated-measures ANOVA used on PC scores for each individual measure. Significant differences in bold ($p < 0.05$) with MBT being the unstable MBT shoe, CON the stable control shoe, PRE being the pre-accommodation testing session and POST the post-accommodation testing session. Supplementary Figure S3 (online) includes the group means and standard error of the means for each comparison.

Table 4. Bonferroni-corrected *post-hoc* comparisons for four waveform measures showing significant interaction effects at hip and ankle.

Joint	Gait measure	Component	PC	Interaction	Pre-accommodation CON vs. MBT	Post-accommodation CON vs. MBT	CON PRE vs. POST	MBT PRE vs. POST
Hip	Moment	Abduction–Adduction	2	**<0.01**	**<0.01**	0.45	0.17	1.00
Ankle	Angle	Abduction–Adduction	3	**0.031**	**<0.01**	**<0.01**	1.00	0.60
Ankle	Moment	Abduction–Adduction	3	**0.017**	1.00	**<0.01**	1.00	**<0.01**
Ankle	Moment	Dorsiflexion–Plantarflexion	2	**0.028**	**<0.01**	**<0.01**	0.112	1.00

Significant differences for pairwise comparisons in bold ($p < 0.05$).

internal rotation moment peaking earlier in stance for the unstable MBT shoe. The angle and moment group mean PC scores and corresponding standard error of the means (interaction plots) for all comparisons in Tables 1–3 are presented in three supplementary figures (Figures S1–S3) accompanying the online version of this manuscript.

3.1. *Hip angles and moments*

For all three hip joint angle measures, there were no overall magnitude differences spanning the entirety of stance (PC1 for all three angles) detected between the stable control shoe and the unstable MBT shoe, for both the pre- and post-accommodation testing sessions (Table 1). Differences in hip adduction angle during early stance (shoe effect for PC2 and PC3, Figure 1A and Table 1) and hip flexion–extension range of motion (shoe effect for PC2, Figure 1B and Table 1) were identified between the two shoes for both visits. Figures 4C, 4G, 4K and 4O more clearly demonstrate the reduced hip flexion–extension amplitude or range of motion while walking in the unstable MBT shoe compared to the stable control shoe for both testing sessions.

Both large and more subtle moment differences were also captured at the hip joint. The abduction–adduction moment was the only moment at the hip that demonstrated a difference between the two shoes (shoe effect for PC1) throughout the majority of stance, with this difference being most evident during the first 30% of stance (Figure 1D and Table 1). More subtle differences (local differences or phase shifts) between the unstable MBT shoe and the stable control shoe were also captured by PC2 and PC3 for each of the three hip moments. These differences are clearly described in Table 1 and can be seen in Figures 1D, 1E and 1F. The only interaction effect at the hip was for the hip abduction–adduction moment, with Bonferroni corrected pairwise comparisons demonstrating that the unstable MBT shoe had a greater peak abduction moment during early stance and reduced peak abduction moment during late stance compared to the stable control shoe for the pre-accommodation testing session only (PC2, Figure 1D and Tables 1 and 4). Differences between the pre- and post-accommodation testing sessions were also identified for the hip flexion–extension moment (visit effect for PC3, Figure 1E and Table 1) and hip internal–external rotation moment (visit effect for PC1, Figure 1F and Table 1).

3.2. *Knee angles and moments*

No differences between footwear conditions or visits were identified for the knee abduction–adduction angle (Figure 2A and Table 2). Differences in knee flexion–extension (shoe effect for PC1, PC2 and PC3) and internal–external rotation angles (shoe effect for PC1 and PC3) were captured between the unstable MBT shoe and the stable control shoe for both visits (Figures 2B and 2C and Table 2). The only knee angle difference when comparing testing sessions was a greater overall knee flexion angle throughout the majority of stance for the pre- versus the post-accommodation testing session (visit effect for PC1, Figure 2B and Table 2).

PC1 captured an overall moment magnitude throughout the entirety of stance for each of the three knee moments; however, differences related to this feature were not identified between the two footwear conditions (no shoe effect, Figures 2D, 2E, 2F and Table 2). Related to PC1 or the overall moment magnitude feature, the only knee moment difference was between testing sessions with the knee moment being shifted more towards extension throughout the entirety of stance for the pre- versus post-accommodation testing session (visit effect for PC1, Figure 2E and Table 2). This was also seen as a reduced knee flexion moment during the last half of stance for the pre-accommodation testing session (visit effect for PC2,

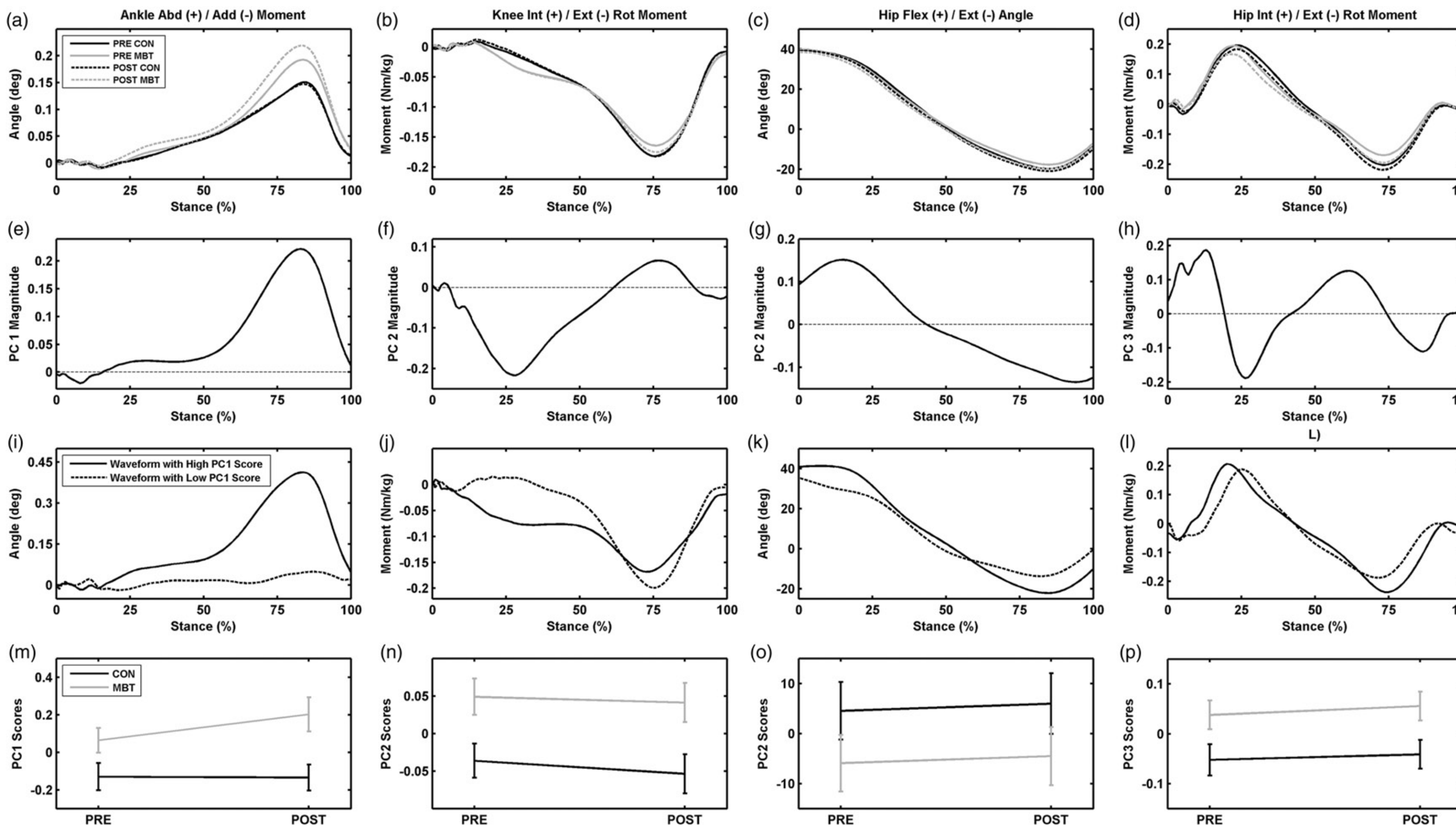

Figure 4. Mean waveforms for the stance phase of walking (first row: A, B, C and D), PC or loading vector waveforms (second row: E, F, G and H), high and low PC score waveforms (third row: I, J, K and L) and PC score means with standard error of the means (fourth row: M, N, O and P) for four representative examples. The first column represents waveforms and analysed data for ankle abduction–adduction angles, the second column represents knee internal–external rotation moments, the third column represents hip flexion–extension angles and the fourth column represents hip internal–external rotation moments.

Figure 2E and Table 2). The remaining identified knee moment differences were between the two footwear conditions at specific portions of the stance phase and these differences were present for all three knee joint moments (shoe effects, Figures 2D, 2E, 2F and Table 2).

3.3. *Ankle angles and moments*

The ankle was the joint in which the greatest number of significant differences were identified, particularly between the two footwear conditions. From the three PCA models used on the ankle joint angles, a total of six shoe effects, three visit effects and one interaction effect were identified (Figure 3 and Tables 3 and 4). Dorsiflexion angle magnitudes were substantially greater while walking in the unstable MBT shoe compared to the stable control shoe, specifically during the first half of stance (shoe effect for PC1, Figure 3B and Table 3). Trends related to differences in the overall angle magnitudes throughout the entirety of stance (PC1) between footwear were also observed for ankle abduction and eversion angles ($p = 0.058$ and $p = 0.082$, respectively) and between visits for dorsiflexion angle ($p = 0.059$). Range of motion differences and/or phase shifts were also identified between the two types of footwear across the three ankle joint angles (shoe effects, Figure 3A, 3B and 3C and Table 3). The only difference between visits for ankle joint angles was a greater eversion angle during early to mid-stance for the pre- compared to the post-accommodation testing session (visit effect for PC2, Figure 3C and Table 3).

Walking in the unstable MBT shoe led to increases in the three ankle joint moments across different portions of the stance phase when compared to the stable control shoe (Figure 3D, 3E, 3F and Table 3). There were a total of six shoe effects, two visit effects and two interaction effects identified from the three PCA models used on the ankle moment gait data (Tables 3 and 4). Compared to the stable control shoe, walking in the unstable MBT shoe led to a greater abduction moment during mid- to late stance (shoe effect for PC1, Figure 3D and Table 3), a greater plantarflexion moment during the first half of stance (shoe effect for PC1, Figure 3E and Table 3), a greater overall inversion moment throughout stance, particularly during the first half of stance (shoe effect for PC1 and PC2, Figure 3F and Table 3), and a reduced plantarflexion moment during 50–90% of stance (shoe effect for PC2, Figure 3E and Table 3). The plantarflexion moment was also greater during the pre- versus post-accommodation testing session for both footwear conditions (visit effect for PC1, Figure 3F and Table 3). The interaction effects associated with the abduction–adduction ankle moment and dorsiflexion–plantarflexion ankle moment are described in Tables 3 and 4.

4. Discussion

This study is one of the first to identify increases in select lower extremity joint moments during unstable MBT walking compared to walking in a conventional stable control shoe. Specifically, the unstable shoe generated greater magnitudes during some portion of stance for ankle joint moments in all three planes (partially satisfying the first hypothesis, H1) and for hip abduction, hip flexion–extension and knee external rotation moments (partially satisfying the second hypothesis, H2). Other researchers have reported reductions in knee joint moments (Buchecker *et al.* 2010) and trends for reductions in hip and knee moments (Nigg *et al.* 2006a) for unstable MBT walking and the current study also identified reductions or temporal changes in select joint moments at the hip, knee and ankle during different portions of stance (further satisfying H1 and H2). Kinematic changes at all three joints were also captured using the PCA technique, with the most evident changes occurring at the ankle joint, and these findings are in agreement with previous studies (Nigg *et al.* 2006a, Romkes *et al.* 2006).

Focusing on the frontal plane, unstable MBT walking led to increased early stance hip adduction angles and decreased late stance adduction angles compared to walking in the stable control shoe, indicating that unstable MBT walking alters the frontal plane alignment beyond the ankle joint. Accompanying these frontal plane hip changes were ankle eversion angle changes, with unstable MBT walking having increased early stance eversion angles and decreased late stance eversion angles. The inversion–eversion ankle angle changes for unstable MBT walking are comparable to the frontal plane ankle changes that occur with the introduction of lateral wedge orthoses aimed at relieving pain in patients with medial knee osteoarthritis (Butler *et al.* 2009). All these changes, along with most others identified in this study, were evident during both the pre- and post-accommodation testing sessions, thereby satisfying the third hypothesis, H3. This hypothesis stated that most gait changes introduced by the unstable MBT shoe would be both immediate and more long-term in nature.

Walking in the unstable MBT shoe introduced a number of frontal plane moment changes across the three joints, with the most evident changes based on observations of the mean ensemble waveforms (Figures 1–3) occurring at the ankle followed by the hip. Ankle inversion moments and peak hip abduction moments, primarily during the first half of stance, were greater for unstable MBT walking. In contrast to these increased frontal plane moments at the hip and ankle, wearing the unstable MBT shoe reduced knee frontal plane moments. These reductions included a decreased peak knee adduction moment immediately after foot contact and a decreased peak knee abduction moment for unstable MBT walking during early stance. These knee moment findings are similar to those of Buchecker *et al.* (2010), who reported a reduction in the peak knee abduction moment in overweight males using the unstable MBT shoe, and to those of Butler *et al.* (2009), who reported a reduction in the peak knee abduction moment in medial knee OA patients when wearing lateral wedge orthoses in their shoe. Anecdotal evidence and peer-reviewed literature (Nigg *et al.* 2006b) indicate that the unstable MBT shoe provides pain relief for many individuals inflicted with such conditions as knee osteoarthritis (OA). Knee OA most commonly affects the medial compartment (Andriacchi and Mundermann 2006) and a greater knee internal abduction moment while walking has been linked to increased medial knee joint loading (Schipplein and Andriacchi 1991), and a high peak internal knee abduction moment can influence the occurrence (Baliunas *et al.* 2002), severity (Sharma *et al.* 1998) and progression rate (Miyazaki *et al.* 2002) of knee OA to the medial compartment. The reduced early stance peak knee abduction moment seen during unstable MBT walking may decrease the medial compartment load enough in individuals with knee OA to provide a degree of pain relief, particularly for those experiencing medial compartment knee OA. It therefore seems that unstable MBT walking leads to a walking strategy with greater frontal plane loadings at the hip and ankle and reduced loadings at the knee, particularly during early stance.

Walking in the unstable MBT shoe also introduced changes in sagittal plane joint angles at the hip, knee and ankle. Unstable MBT walking led to a decreased hip flexion–extension range of motion throughout stance compared to walking in a stable control shoe. These findings agree with those of Romkes *et al.* (2006), who attributed the decreased range of motion to a reduced walking speed, cadence and stride length for the unstable MBT shoe; however, in the current study the walking speeds were statistically similar. Walking in the MBT shoe also led to early stance peak knee flexion occurring closer to foot contact compared to the stable control shoe. The knee was also more flexed throughout the majority of stance when walking in the unstable MBT shoe (versus the conventional stable control shoe) and for the pre-accommodation testing session (versus the post-accommodation testing session). The most evident sagittal plane differences between both footwear types were identified at the ankle joint, with unstable MBT walking leading to a more dorsiflexed ankle throughout the first half of stance and a less dorsiflexed ankle during the second half. These sagittal plane findings agree with similar MBT walking studies on the knee (Romkes *et al.* 2006) and ankle (Nigg *et al.* 2006a, Romkes *et al.* 2006). Stewart *et al.* (2007) used in-shoe pressure distribution measurements to show that an individual's weight is displaced from the heel and towards the midfoot near foot strike while walking in the unstable MBT shoe. The shoe prevents individuals from 'striding out', which is manifested through a reduced hip range of motion and increased knee flexion angle during stance. The MBT's rounded heel in the anterior–posterior direction also seems to restrict the degree of plantarflexion just after foot contact by preventing foot drop, which is a normal biomechanical feature observed during walking in a more conventional stable control shoe.

Unstable MBT walking also introduced several sagittal and transverse plane moment changes that have not been identified previously in the literature. Compared to the stable control shoe, unstable MBT walking produced greater peak hip extension moments during early stance, along with greater knee external rotation moments and plantarflexion moments during the first half of stance. The ankle also experienced greater ankle abduction (toe-out) moments during mid- to late stance and greater inversion moments throughout the entirety of stance for unstable MBT walking. These moment increases suggest that joint loadings may be larger during specific portions of stance, depending on the gait measure, and that the contribution of the muscles crossing the three joints may also be altered as a result of these moment changes. In contrast to the greater moments mentioned above, walking in the unstable MBT shoe also led to a reduction in peak hip internal rotation moments during early stance and a reduction in peak values for ankle plantarflexion, knee flexion, knee external rotation and hip external rotation moments during late stance. These moment reductions during late stance could be a result of the reduced dorsiflexion angle and the MBT's curved sole helping with push-off just prior to the foot leaving the ground.

For transverse plane joint angles, differences were captured at the knee and ankle but not at the hip. The most evident change at the knee was a greater internal rotation angle during stance for unstable MBT walking. At the ankle joint, walking in the unstable MBT shoe decreased stance adduction–abduction (toe-in–toe-out) range of motion and produced a strong trend ($p = 0.058$) for an overall reduced abduction (toe-out) angle across the entirety of stance. With Romkes *et al*. (2006) investigating only sagittal plane joint angles and Nigg *et al*. (2006a) capturing joint angle differences only in the sagittal plane, the present study serves as the first to identify transverse plane joint angle changes at both the ankle and knee during unstable MBT walking.

In conclusion, the application of PCA proved to be very effective at identifying kinematic and kinetic differences between walking in the unstable MBT shoe versus a more conventional stable control shoe. Many of the findings are similar to previous biomechanical studies on the unstable MBT shoe (Nigg *et al*. 2006a, Romkes *et al*. 2006, Buchecker *et al*. 2010, Roberts *et al*. 2011) but several additional changes were also identified that have not been previously reported in the literature. The unstable MBT shoe seems to have its greatest effect on joint angles and moments at the ankle joint followed by the hip and knee. More advanced forward dynamic and muscle modelling studies using these biomechanical findings could offer improved insight into (i) how these unstable MBT shoes are able to provide pain relief in some individuals and (ii) if there are any long-term implications from wearing such footwear.

Appendixes

Three supplementary figures, showing the group mean PC scores along with the standard error of the means, accompany the online version of this manuscript.

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References

Andriacchi, T.P. and Mundermann, A., 2006. The role of ambulatory mechanics in the initiation and progression of knee osteoarthritis. *Current Opinion in Rheumatology*, 18 (5), 514–518.

Baliunas, A.J., *et al*., 2002. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis and Cartilage*, 10 (7), 573–579.

Boyer, K.A. and Andriacchi, T.P., 2009. Changes in running kinematics and kinetics in response to a rockered shoe intervention. *Clinical Biomechanics*, 24 (10), 872–876.

Buchecker, M., *et al*., 2010. Lower extremity joint loading during level walking with Masai barefoot technology shoes in overweight males. *Scandinavian Journal of Medicine and Science in Sports*, Published online 30 August 2010. doi:10.1111/j.1600-0838.2010.01179.x.

Butler, R.J., *et al*., 2009. Effect of laterally wedged foot orthoses on rearfoot and hip mechanics in patients with medial knee osteoarthritis. *Prosthetics and Orthotics International*, 33 (2), 107–116.

Deluzio, K.J., *et al*., 1997. Principal component models of knee kinematics and kinetics: normal vs. pathological gait patterns. *Human Movement Science*, 16 (2–3), 201–217.

Grood, E.S. and Suntay, W.J., 1983. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *Journal of Biomechanical Engineering*, 105 (2), 136–144.

Landry, S.C., *et al*., 2007a. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated run and crosscut maneuver. *American Journal of Sports Medicine*, 35 (11), 1901–1911.

Landry, S.C., *et al*., 2007b. Knee biomechanics of moderate OA patients measured during gait at a self-selected and fast walking speed. *Journal of Biomechanics*, 40 (8), 1754–1761.

Landry, S.C., Nigg, B.M., and Tecante, K.E., 2010. Standing in an unstable shoe increases postural sway and muscle activity of selected smaller extrinsic foot muscles. *Gait and Posture*, 32 (2), 215–219.

McKean, K.A., *et al*., 2007. Gender differences exist in osteoarthritic gait. *Clinical Biomechanics*, 22 (4), 400–409.

Miyazaki, T., *et al*., 2002. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. *Annals of the Rheumatic Diseases*, 61 (7), 617–622.

Nigg, B., Hintzen, S., and Ferber, R., 2006a. Effect of an unstable shoe construction on lower extremity gait characteristics. *Clinical Biomechanics*, 21 (1), 82–88.

Nigg, B.M., Emery, C., and Hiemstra, L.A., 2006b. Unstable shoe construction and reduction of pain in osteoarthritis patients. *Medicine and Science in Sports and Exercise*, 38 (10), 1701–1708.

Nigg, B.M., *et al*., 2009. The effectiveness of an unstable sandal on low back pain and golf performance. *Clinical Journal of Sport Medicine*, 19 (6), 464–470.

Nigg, B.M., *et al*., 2010. Gender differences in lower extremity gait biomechanics during walking using an unstable shoe. *Clinical Biomechanics*, 25 (10), 1047–1052.

Ramstrand, N., Andersson, C.B., and Rusaw, D., 2008. Effects of an unstable shoe construction on standing balance in children with developmental disabilities: a pilot study. *Prosthetics and Orthotics International*, 32 (4), 422–433.

Ramstrand, N., *et al.*, 2010. Effects of an unstable shoe construction on balance in women aged over 50 years. *Clinical Biomechanics*, 25 (5), 455–460.

Roberts, S., Birch, I., and Otter, S., 2011. Comparison of ankle and subtalar joint complex range of motion during barefoot walking and walking in Masai Barefoot Technology sandals. *Journal of Foot and Ankle Research*, 4 (1), 1–4.

Romkes, J., Rudmann, C., and Brunner, R., 2006. Changes in gait and EMG when walking with the Masai Barefoot Technique. *Clinical Biomechanics*, 21 (1), 75–81.

Rutherford, D.J., *et al.*, 2008. Foot progression angle and the knee adduction moment: a cross-sectional investigation in knee osteoarthritis. *Osteoarthritis and Cartilage*, 16 (8), 883–889.

Schipplein, O.D. and Andriacchi, T.P., 1991. Interaction between active and passive knee stabilizers during level walking. *Journal of Orthopaedic Research*, 9 (1), 113–119.

Sharma, L., *et al.*, 1998. Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. *Arthritis and Rheumatism*, 41 (7), 1233–1240.

Stewart, L., Gibson, J.N., and Thomson, C.E., 2007. In-shoe pressure distribution in 'unstable' (MBT) shoes and flat-bottomed training shoes: a comparative study. *Gait and Posture*, 25 (4), 648–651.

Stoggl, T., *et al.*, 2010. Short and long term adaptation of variability during walking using unstable (MBT) shoes. *Clinical Biomechanics*, 25 (8), 816–822.

Wrigley, A.T., *et al.*, 2005. Differentiating lifting technique between those who develop low back pain and those who do not. *Clinical Biomechanics*, 20 (3), 254–263.