



**Changes in gait characteristics of a normal,
healthy population due to an unstable
shoe construction.**

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1.0 Introduction

This report was commissioned by Swiss Masai AG (Switzerland) and details research performed at and by The Centre for Sport and Exercise Science at Sheffield Hallam University. This independent research presents a scientific approach to the evaluation of gait, the results of which will inform the reader regarding the effect of MBT shoes on normal gait characteristics. The advanced biomechanics technology used will answer the following questions:

- Does wearing MBT shoes affect;
 - normal gait kinematics?
 - normal gait kinetics?
 - Muscle activation patterns?

Back pain is the third most common bodily symptom after headache and tiredness with 60 -80 % of people suffering low back pain at some time in their lives. Of these, 15-30% of people have some low back pain symptoms every day and 35-40 % report low back pain lasting 24 hours or more each month (*Croft et al.*). Back pain in itself is not becoming more common, rather there is an epidemic of absenteeism due to simple backache. In 1995-96, 117 560 500 days invalidity or sickness benefit were claimed as a result of back incapacity in the UK (Department of Social Security figures) and it is estimated that chronic back problems account for 85 % of absenteeism due to back pain (Barton *et al.* 2001).

Three systematic reviews (Evans and Richards, 1996; Van Tulder *et al.* 1997; Faas, 1996) have found that exercise improved pain ,disability and outcomes such as strength and flexibility.

One systematic review (Bigos *et al.* 1994) has found that in people with acute low back pain, advice to stay active speeds symptomatic recovery, reduces chronic disability and leads to less time off work compared to bed rest and traditional medical treatments such as analgesia, advice to rest and 'let pain be your guide' to activity.

Osteoarthritis is another common cause of pain and disability in older adults (Petersen, 1996; Felson, 1988). Radiographic changes are common in one or more joints in the majority of people aged 60 and above. Significant clinical disease affects up to 20% of these older populations; Osteoarthritis of the knee being twice as prevalent as osteoarthritis of the hip in people aged over 60 (Gaffney, 1995).

There have been three systematic reviews of the benefits of exercise and physical therapy in the management of osteoarthritis (Van Baar *et al.*, 1999; Puett and Griffin, 1994 and La Mantia and Marks, 1995). All showed the positive benefits of exercise programmes from the use of strength or aerobic activities or combinations of the two. These benefits being in the form of pain reduction and improved joint function.

The average UK citizen walks in the region of 183 miles (305 Km) per year ⁽¹⁾ (<http://www.statistics.gov.uk>). This equates to a huge number of footfalls every year, each of which results in a loading effect on the joints of the lower limb and back. As such, any reduction in the cyclic loading of the structures that comprise these joints will have a significant effect on the cumulative microdamage.

Anecdotal evidence has suggested that the frequent use of an unstable shoe construction can help in rehabilitation and even prevention of many of these complaints. The Masai Barefoot Technology (MBT) shoe (Figure 1.1) has a rounded sole in the anterior – posterior direction, by providing an uneven walking surface it challenges the muscles to maintain control over locomotion rather than relying on the stability provided by a conventional shoe.

MBT gait evaluation.

(1) On the public highway or other unrestricted areas which are paved or tarred and over 50 yards in length.



Figure 1.1: The Masai Barefoot Technology (MBT) shoe; note the rounded sole construction in the anterior – posterior direction.

If the MBT shoe can be shown to reduce loading through the hip, knee and ankle as suggested by a preliminary study from The Human Performance Laboratory University of Calgary (*Nigg B et al. 2004*) the potential beneficial impact the shoe could have as a tool for treating lower back disorders and Osteoarthritic hip and knee would be significant.

2.0 Methods

Participants

Twenty-two participants (11 male, 11 female) volunteered to take part in the study; Details of the participants' age, stature and body mass are given in Table 2.1 below. All participants were physically active and free from musculoskeletal injury at the time of testing. The University's Ethics Committee approved the procedures, and written informed consent was gained from each participant before data collection. On volunteering to take part in the study, each participant was provided with Masai Barefoot Technology (MBT) shoes (Figure 1.1) which were worn during the data collection sessions in the 'MBT condition' in addition to tight fitting shorts and vests. In the 'normal condition' participants were required to wear the same tight fitting clothing and their normal exercise shoes.

	Male	Female
Age (years)	28 (4)	26 (3)
Stature (m)	175 (7)	168 (5)
Body Mass (kg)	76 (11)	68 (11)

Table 2.1: Participant details (Mean \pm SD).

Experimental set-up

All kinematic data were collected using an eight-camera digital motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 200 Hz. The orientation of the right handed, orthogonal global coordinate system was such that the positive x axis pointed in the direction of forward progression, the positive y axis pointed vertically upward and the positive z axis pointed to the right.

The eight cameras of the motion capture system were placed in optimal positions around a calibrated measurement volume of dimensions 5.0 _ 2.0 _ 3.0 m in the x, y and z directions respectively. The measurement volume was made this size to incorporate a step both before and after the stance phase on the Kistler Type

9281CA force platform (Kistler Instrumente AG Winterthur, Switzerland) which was embedded in the laboratory floor and covered with a surface common to the entire laboratory. The force platform sampled data at 1000 Hz and was time-synchronised with the motion capture system. Wireless Infra-red timing gates (Brower Timing Systems, Utah, USA) were placed 5 m apart either side of the floor-mounted force platform to monitor walking speed during the trials. Muscle activation data were collected using an eight channel MT8 telemetred EMG system (MIE Ltd, Leeds, UK), feeding into the 3D motion capture system for synchronisation with the kinematic and force plate data. A schematic representation of the experimental setup is given in figure 2.1.

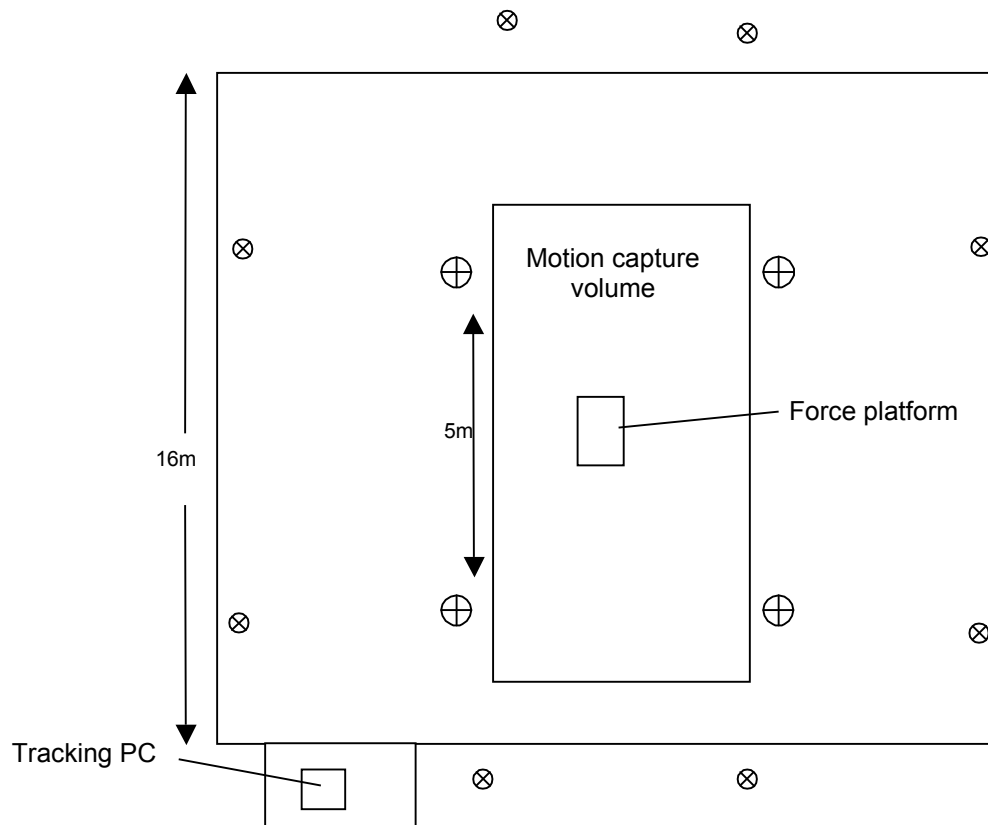


Figure 2.1: The experimental setup. ⊕ - Timing gate, ⊗ - Digital motion capture camera.

Twenty two retro-reflective markers (12.5 mm diameter) were attached to each participant in locations consistent with the Helen Hayes Marker Set (Kadaba *et al.*, 1990). Markers were attached to the participants' left and right 2nd metatarsal head (closest approximation on the shoe), posterior aspect of the calcaneus (closest approximation on the shoe), lateral malleolus, lateral epicondyle, anterior superior

iliac spine, acromion process, lateral humeral epicondyle and midpoint of the radius and ulnar styloid processes. Further markers were attached to the sacrum the seventh cervical vertebrae and the right scapula. In addition to these markers, four supplementary markers, mounted on 100 mm posts were attached to the lateral aspect of the participants' left and right shank and thigh (see figure 2.2).



Figure 2.2: The Helen Hayes Marker Set

Following suitable skin preparation in accordance with European recommendations for surface electromyography (Hermens *et al.*, SENIAM, 1999), surface EMG electrodes were positioned on Gastrocnemius, Gluteus Maximus, Biceps femoris and Multifidus on the right side, in line with European recommendations for surface electromyography (Hermens *et al.*, SENIAM, 1999).

Procedures

Tutorial session

Prior to testing, participants were required to attend a tutorial session in which they were provided with their MBT shoes and instruction conforming to manufacturer's specifications. Instruction was included exercises based on a theory of Dynamic Stability Training (DST) designed to maximize proprioceptive input to stabilizing muscles, and activate kinaesthetic learning processes using MBT footwear.

Exercises included:

- ◆ Relaxed Walking;
- ◆ Walking quickly with shorter and faster steps;
- ◆ Standing and rocking on pivot area of MBT;
- ◆ Somatosensory Balancing Drill* (eyes closed, balancing on pivot area of MBT);
- ◆ Dynatwist* standing and balancing;
- ◆ Dynatwist* walking and sensing (slow) with mid-foot loading on the pivot area of MBT;
- ◆ Relaxed walking;
- ◆ Walking quickly with shorter and faster steps;
- ◆ Postural Pull* exercise;
- ◆ walking Alternate Glutes* exercise;
- ◆ walking slowly;
- ◆ Relaxed walking;
- ◆ Walking quickly with shorter and faster steps Backwards;
- ◆ Walking with Postural Pull*;
- ◆ Crankshaft Hip Recruitment Exercise*;
- ◆ Fast Ski Walking* into jogging;
- ◆ Relaxed walking;

Following training, participants were instructed not to use the shoes until after the first testing session. Following this time they were to use them as frequently as possible.

Data collection session

After preparation and attachment of the Helen Hayes Marker Set, and surface EMG electrodes, each participant was required to traverse the laboratory, approximately 16 m in length, at their preferred speed while making contact with the force platform with the appropriate foot. A right foot contact was required in the direction of the positive x axis, whereas, a left foot contact was required in the opposite direction. The participants completed five 'good' trials for each foot in both the MBT and normal shoe conditions – the order in which the participants completed the conditions was randomised. Trials were accepted when the whole of the participant's appropriate foot contacted the force platform, without any obvious alterations to their gait. Participants were permitted as many practice trials as they required to become able to consistently achieve this prior to the onset of data collection. Five seconds of kinematic data were collected using the motion capture system along with the kinetic data from the force platform and EMG system. Walking speed was measured during every trial using the infrared timing gates; repeated measures t -tests were performed on these data and indicated that walking speed was consistent between conditions ($p>0.05$).

Static calibration trial

At the end of the data collection for each condition a further, static calibration trial was collected to allow for correct anatomical reference frame alignment. Additional markers were attached to the participants' left and right medial malleolus and medial femoral epicondyle. Kinematic data were collected for 5 seconds with the participant in the calibration standing position (see figure 2.2).

Data analysis

Kinematics

The three-dimensional coordinate data were filtered using a second order low-pass Butterworth filter; a cut-off frequency of 6 Hz was used and was selected through visual inspection of the fit. Hip, knee and ankle Joint Coordinate System (JCS) angles (Grood and Suntay, 1983) were then calculated using Orthotrak software (Motion Analysis Corporation, Santa Rosa, CA, USA). The three-dimensional angles of the pelvis and trunk were relative to the relative to the global coordinate system were also calculated using the orthotrak software.

The resulting angular displacement profiles were then cropped to the length of one foot contact. The vertical component of the ground reaction force was used to determine foot contact events - thresholds of 20 N and 10 N were used to determine foot-strike and toe-off respectively. Twenty-two dependent variables were taken from the kinematic data. These included the degree of angular displacement at the ankle, knee and hip joints in the sagittal plane at foot-strike and toe-off. Sagittal plane trunk and pelvis angles, relative to the global coordinate system, were also recorded at the times of these gait events. In addition to the variables, further discrete kinematic data were recorded at the times of maximum and minimum angular displacement for each profile.

Ground reaction force

Six further dependent variables were taken from the ground reaction force data. These included; peak vertical force in the first half of stance (impact peak), peak vertical force in the second half of stance (propulsive peak), maximum propulsive force and maximum braking force.

Kinetics

The Orthotrak software was used to calculate internal resultant joint moments using an inverse dynamics technique. Dependent variables for the kinetics data included the resultant joint moment at the ankle, knee and hip joints in the sagittal plane at foot-strike and toe-off. Furthermore, maximum and minimum sagittal plane joint moments were recorded at the ankle, knee and hip joints.

Muscle Activation

EMG data were analysed using Visual 3D software (C-Motion Inc, Rockville, MD, USA). A data smoothing technique was performed using a low pass Butterworth filter (400Hz) to filter out any high frequency component not associated with muscle electrical activity, and a high pass Butterworth filter (15Hz) to remove any movement artefact. A second low pass Butterworth filter was used to smooth the data to highlight changes in amplitude of the electrical signal.

Statistical analysis

Firstly, the data for each dependent variable were screened to ensure that it did not violate the assumptions of repeated measures analysis of variance (ANOVA) or *t*-test - normality, sphericity (for ANOVA only) and homogeneity of variance. A series of three-factor (condition, foot, gender) ANOVA, with repeated measures on the factors condition and foot, were performed for each dependent variable to assess differences between normal shoe and MBT walking. Ground reaction force graphs were visually inspected for the presence of a transient peak, identified by a short spike of force immediately following initial contact of the foot with the ground. A Wilcoxon matched pairs test was performed on these data to assess differences between the MBT and normal conditions. The alpha level of significance was set at 0.05 for all statistical tests. Muscle activation graphs were visually inspected and the stance phase broken down into four equal parts, the mean and percentage difference in amplitude of the electrical signal for each of the four phases of stance were calculated for both the MBT and normal shoe conditions.

3.0 Results and Discussion

Kinematics

Comparison of walking in normal and MBT shoes suggested that lower limb kinematics were largely unchanged (Table A1-4). Both hip, knee and pelvis kinematics were not different between the normal and MBT conditions. There were however two significant differences in kinematics; the use of MBTs elicited a significant decrease ($p < 0.05$) in the forward lean of the trunk (Figure 3.1) suggesting subjects adopted a more upright walking posture when using the MBT shoes.

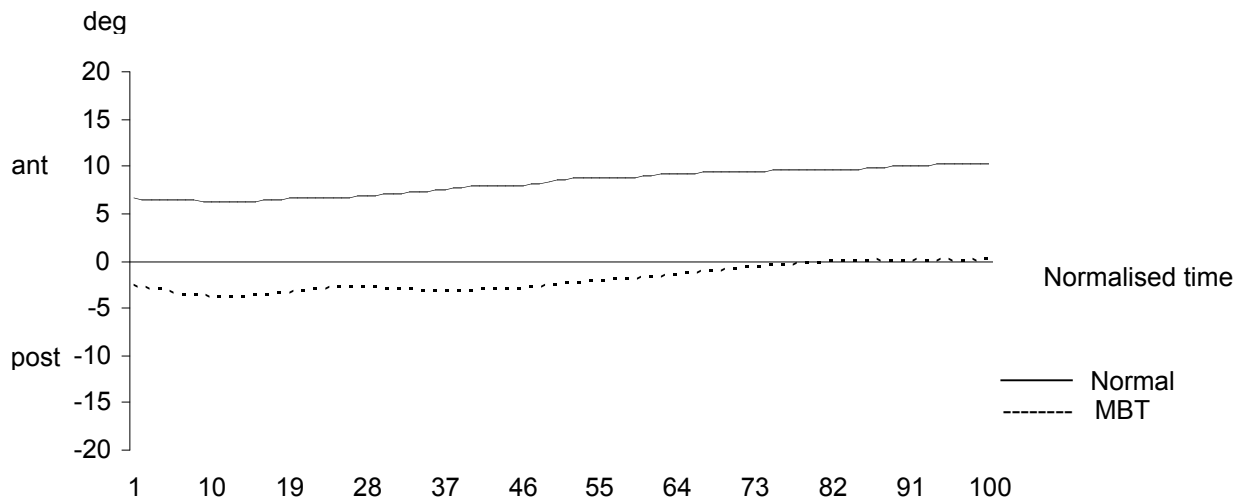


Figure 3.1: Trunk angle during the stance phase of gait for the MBT and normal condition.

At the ankle (Figure 3.2) subjects exhibited significantly higher maximum dorsiflexion angles when wearing MBT shoes ($p < 0.05$). Also evident was a significant decrease in the amount of plantar flexion following initial contact when wearing MBTs ($p < 0.05$). This is primarily due to the rolling action over the pivot point of the MBT shoe rather than the dropping of the forefoot following initial contact, as seen during normal walking.

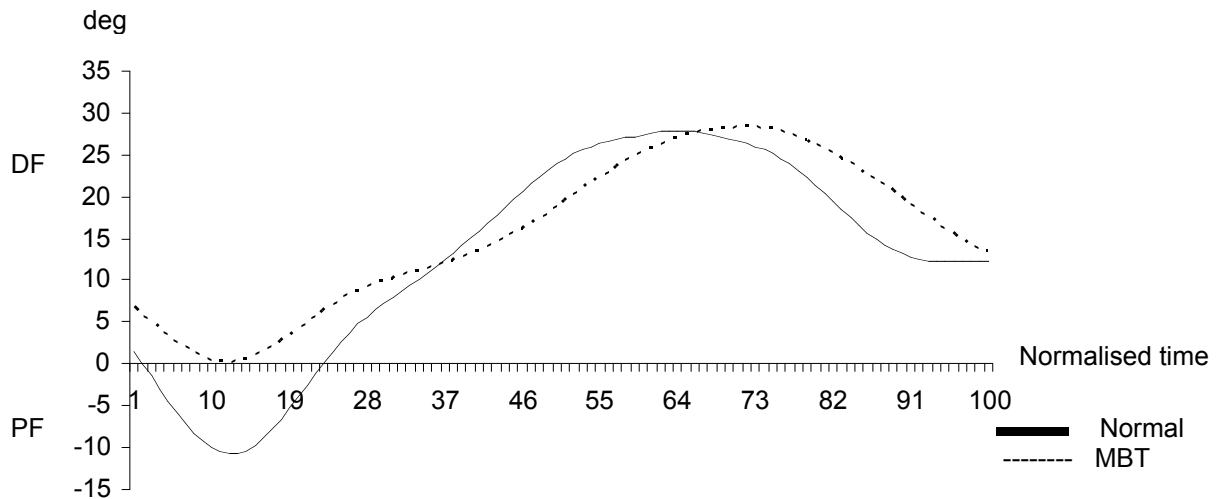


Figure 3.2: Ankle angle during the stance phase of gait for the MBT and normal condition.

Main finding: MBT shoes promote a more upright trunk posture during locomotion and a more dorsiflexed ankle position throughout stance.

Kinetics

Analysis of joint kinetics during walking showed a number of significant differences in the joint moments experienced during the MBT and normal conditions. At the hip (Figure 3.3) and ankle (Figure 3.5) there were significant reductions ($p < 0.05$) in the maximum extension and plantar flexion moments experienced during the MBT condition compared to the normal condition. Although this was not the case at the knee (Figure 3.4), there was a trend toward lower extension moments when wearing MBT shoes.

The ankle (Figure 3.5) also experienced significantly lower dorsiflexion moments during the MBT condition compared to the normal condition. This can be explained by the differences in the initial contact position during the MBT condition; the more dorsiflexed contact position and the reduced plantar flexion during the initial phase of stance. The knee (Figure 3.4) also exhibited a significantly lower flexion moment, flexion moments at the hip were found to be non-significant between conditions but showed a trend toward being lower during the MBT condition.

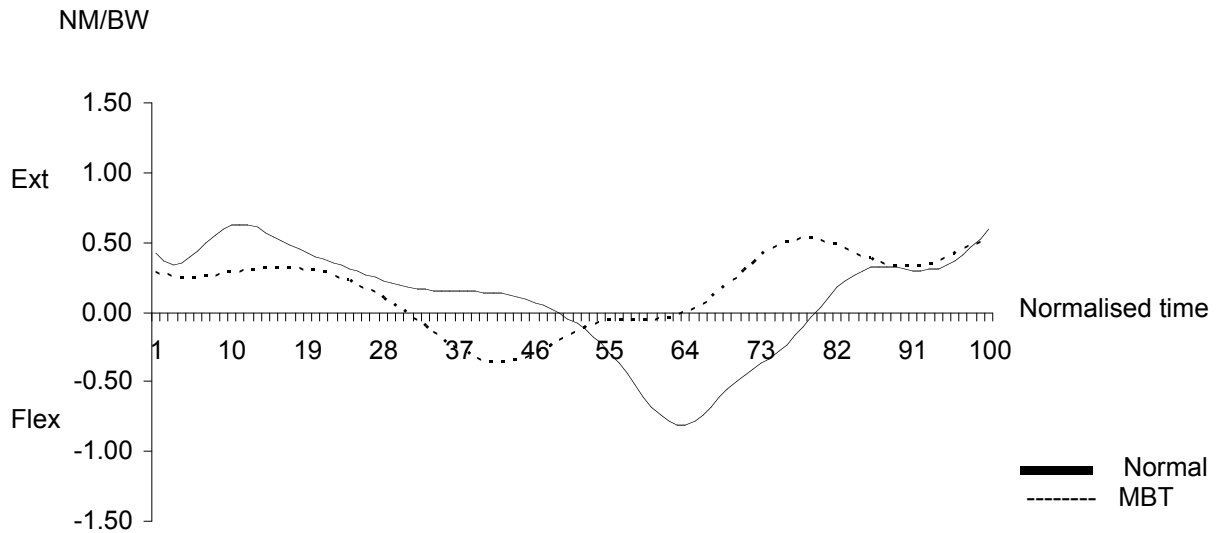


Figure 3.2: Hip flexion/extension moment during the stance phase of gait for the MBT and normal condition.

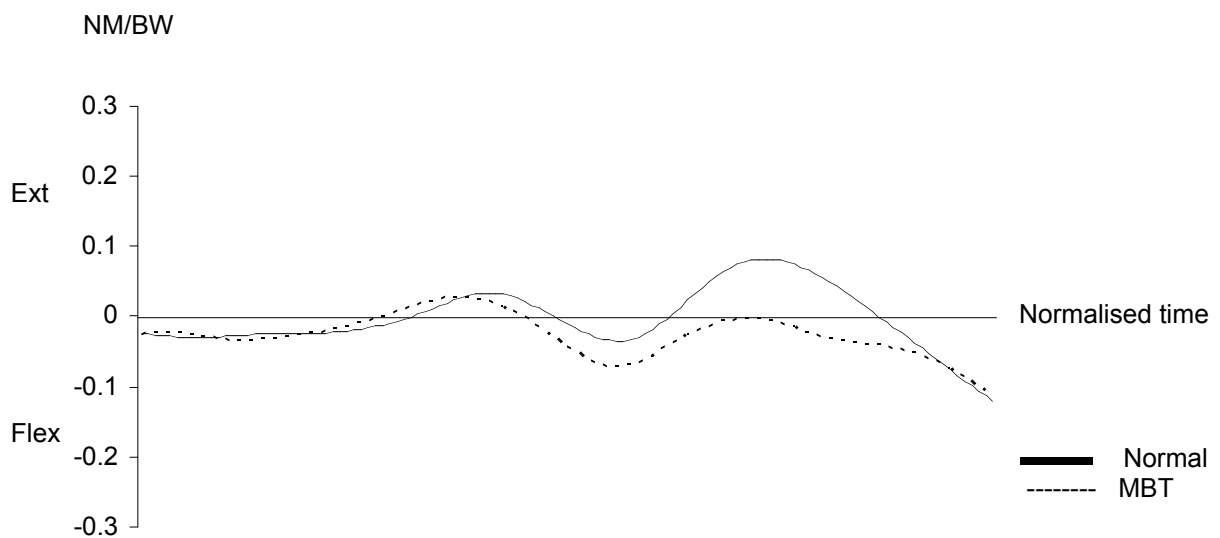


Figure 3.3: Knee flexion/extension moment during the stance phase of gait for the MBT and normal condition.

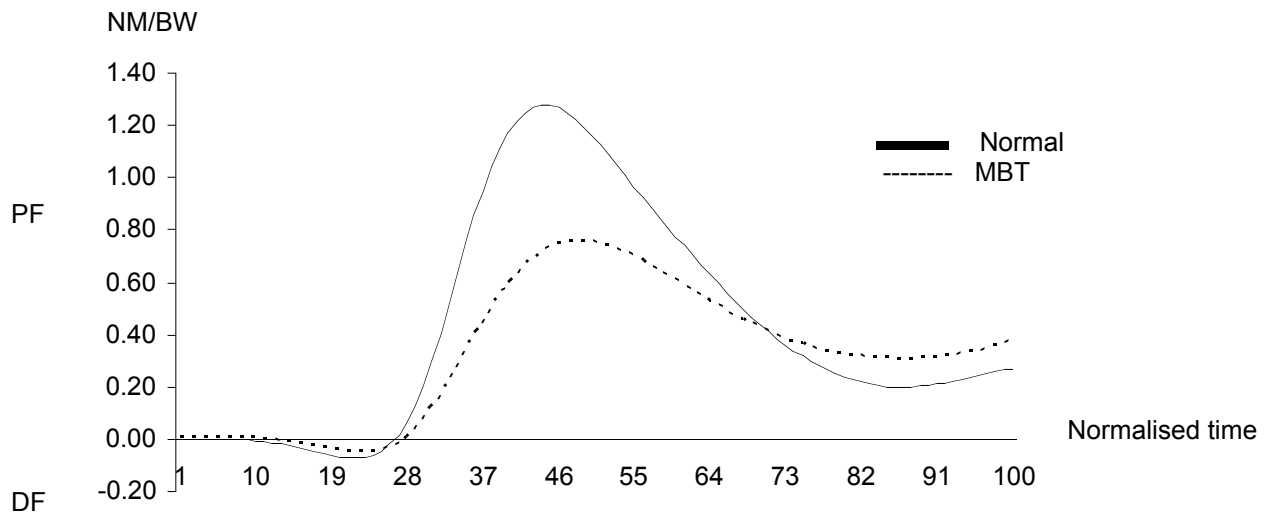


Figure 3.5: Ankle plantar flexion/dorsiflexion moment during the stance phase of gait for the MBT and normal condition.

Main finding: During the MBT shoe condition, subjects experienced reduced sagittal plane joint moments at the hip, knee and ankle.

Ground Reaction Force

Analysis of the ground reaction forces (Table A9) suggested that the MBT condition elicited a significant ($p < 0.05$) increase in the vertical impact peak. It is suggested that this is due to the increased mass of the MBT shoe when compared to the participant's normal shoes. No difference was found in the magnitude of the active peaks. Visual inspection of the vertical ground reaction force curves (Figure 3.6) suggested that the occurrence of a transient peak was more common ($p < 0.05$) in the normal condition than during the MBT condition with transient peaks occurring in 59% of the normal trials and only 27% of the MBT trials. This is probably as a direct result of the difference in foot contact with the ground when wearing MBT shoes or differences in shoe construction.

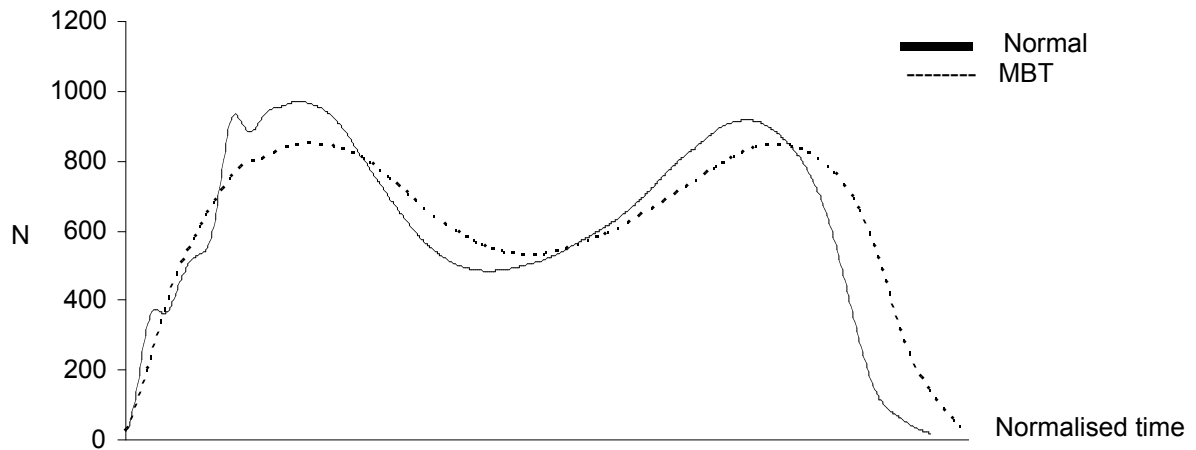


Figure 3.6: Vertical ground reaction force trace during the stance phase of gait for the MBT and normal condition.

I was also noted that where there was a transient peak present, the impact peak in the MBT condition was lower than that experienced during the normal condition. This, however, needs further investigation in order to be able to make any further assumptions.

In the anterior – posterior direction (Figure 3.7) the MBT condition showed significantly higher propulsive forces during the latter part of stance. This is probably due to the dragging motion promoted when walking in MBT shoes rather than the pushing motion used when walking in normal shoes.

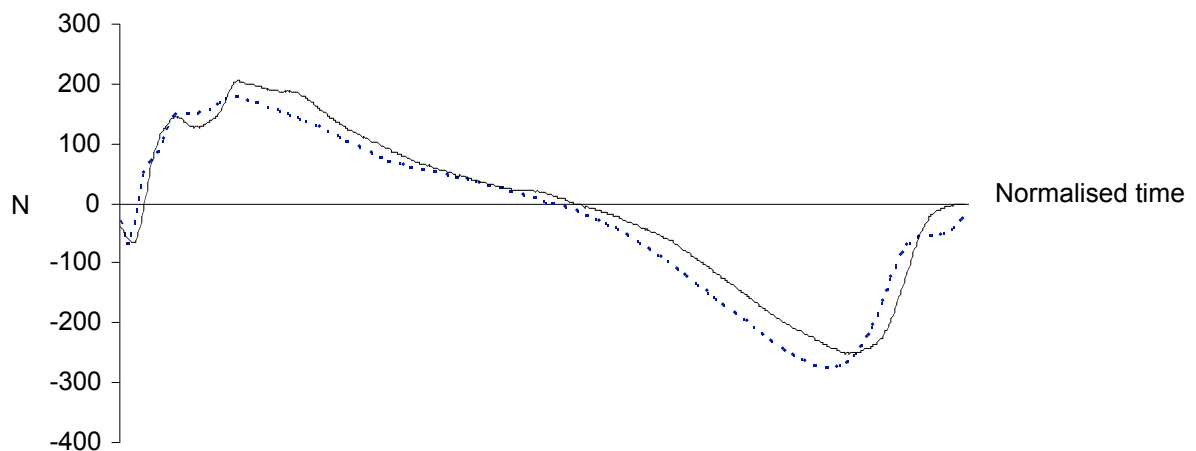


Figure 3.7: Anterior - posterior ground reaction force trace during the stance phase of gait for the MBT and normal condition.

Main finding: The incidence of transient peaks when wearing MBT shoes was lower than during the normal condition. MBT's also promote an increased propulsive (forward force).

EMG

MBTs elicit a heightened electrical activity (motor unit recruitment) at certain times during the stance phase of the gait cycle. These are related to the kinematics of MBT gait.

Heightened activity in the gastrocnemius (Fig 3.8a) occurs during initial stance as the lower leg has to work to control the instability of the MBT shoe during contact. In the hamstring activity (Fig 3.8b) is heightened during the initial stage but also during the propulsive part of mid stance as the hip extends and foot is 'pulled' backwards to move the centre of gravity forwards. This occurs slightly after an increase in gluteal electrical activity (Fig 3.8c) which occurs as a result of the stabilizing of the pelvis prior to hip extension. Reduced electrical activity in Multifidus (Fig 3.8d) is seen during initial stance possibly due to a reduction in the forward lean of the trunk and the subsequent reduction in the muscle activity required to remain upright.

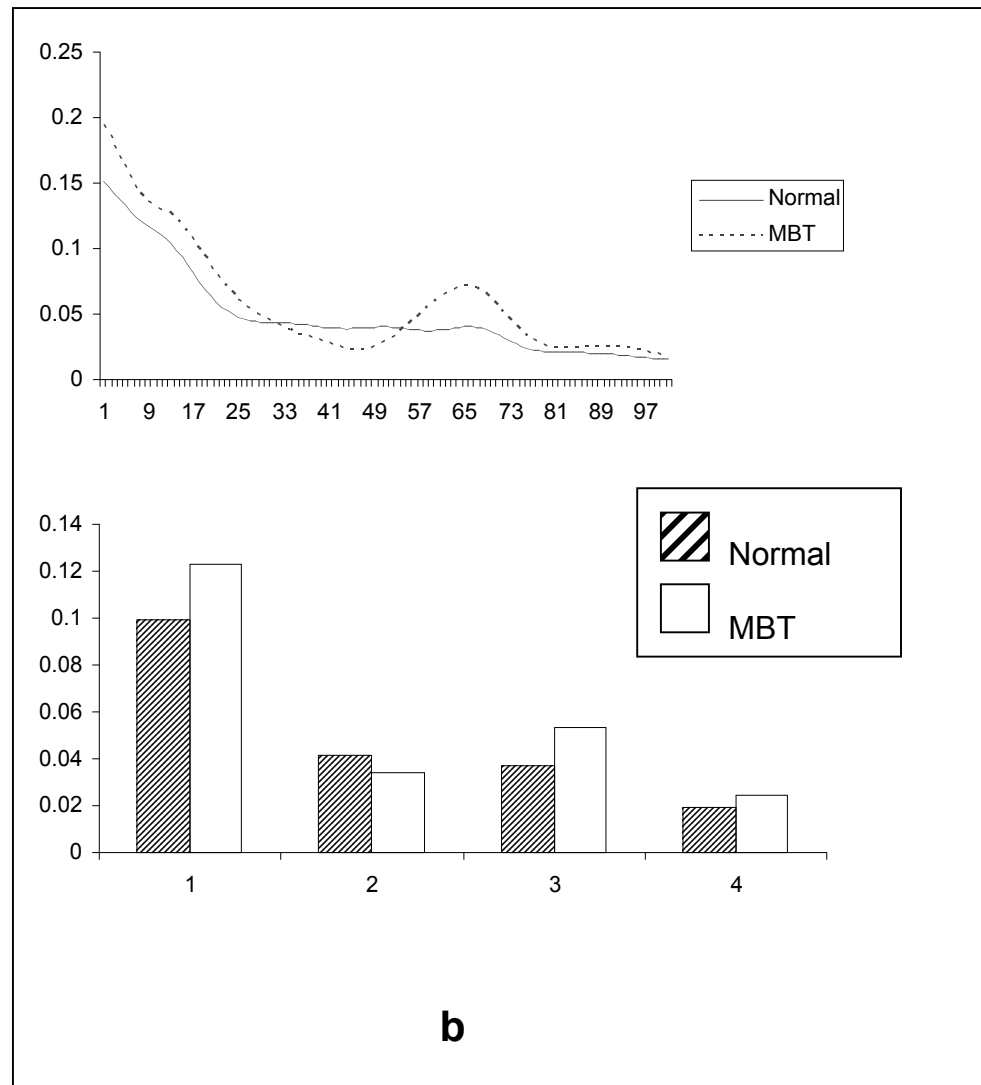
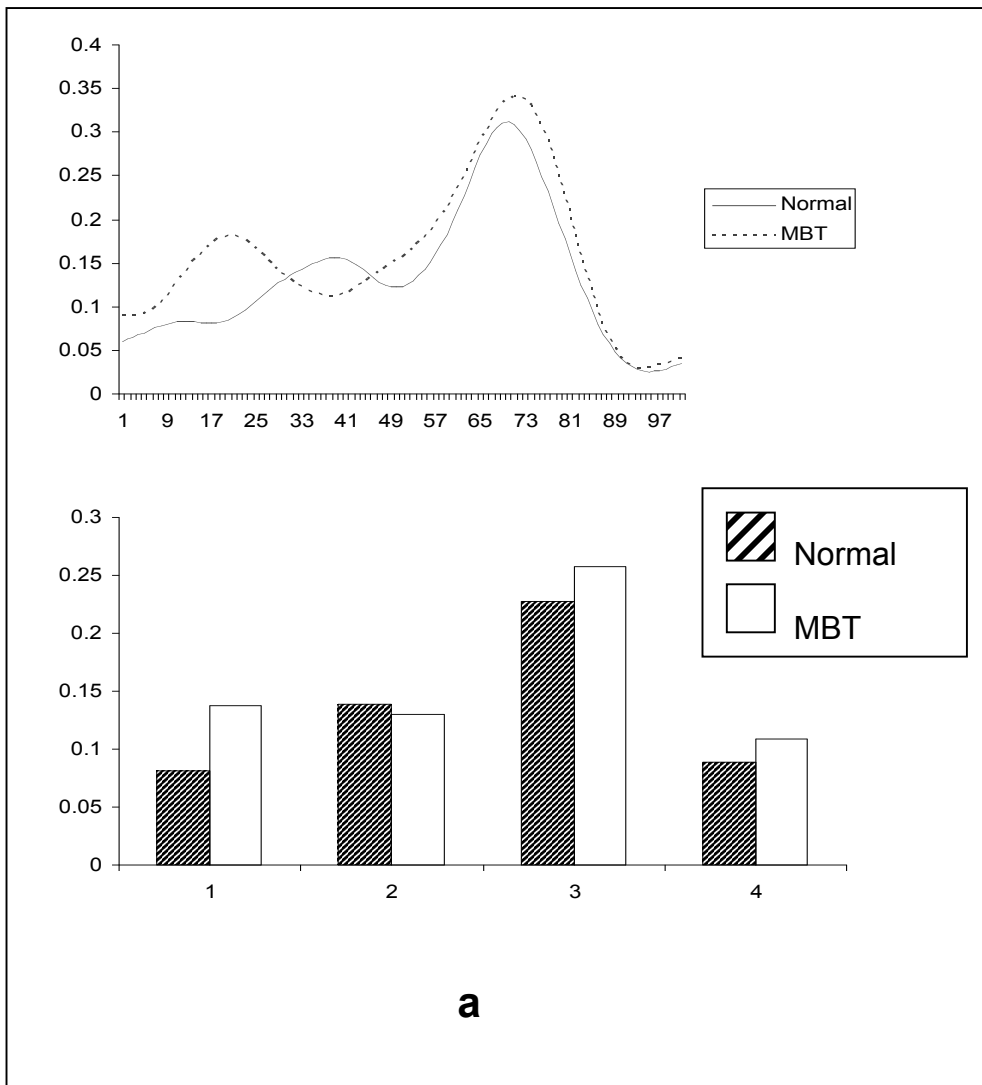
Percent change in electrical signal amplitude is shown in table 3.1 and suggests that overall for the three muscles of the lower limb, electrical activity increased when using MBTs for Multifidus, overall electrical activity decreased, again highlighting a reduction in muscle activity required to maintain upright posture. The result could be a reduction in the fatiguing effects of muscle activity and a subsequent reduction in lower back stress.

	Percentage of stance phase				
	25	50	75	100	0-100
gastroc norm	0.0809	0.1384	0.2278	0.0892	0.1336
gastroc mbt	0.1376	0.1299	0.2573	0.1086	0.1579
% change	70.182	-6.17	12.946	21.793	18.137
ham norm	0.0989	0.0414	0.0369	0.0195	0.0489
ham mbt	0.123	0.0344	0.0534	0.0242	0.0584
% change	24.369	-16.93	44.671	24.182	19.478
glute norm	0.0762	0.0154	0.0152	0.0177	0.031
glute mbt	0.066	0.0229	0.0271	0.0198	0.0338
% change	-13.31	49.349	78.558	12.02	9.261
mult norm	0.0531	0.012	0.0116	0.0238	0.0251
mult mbt	0.044	0.012	0.0115	0.0247	0.0231
% change	-17.16	-0.186	-1.105	3.9014	-8.178

Table 3.1: Percentage change in electrical activity across all trials for all participants.

Main finding: Recruitment of the tested muscles is altered when wearing MBTs. Increased electrical activity in Gastrocnemius, Biceps Femoris and Gluteus Maximus suggest an increase in motor unit recruitment during MBT walking. MBTs elicit a reduction in electrical activity in Multifidus as a result of the more upright posture associated with MBT walking and the subsequent decrease in need for muscular work.

MBT gait evaluation.



MBT gait evaluation.

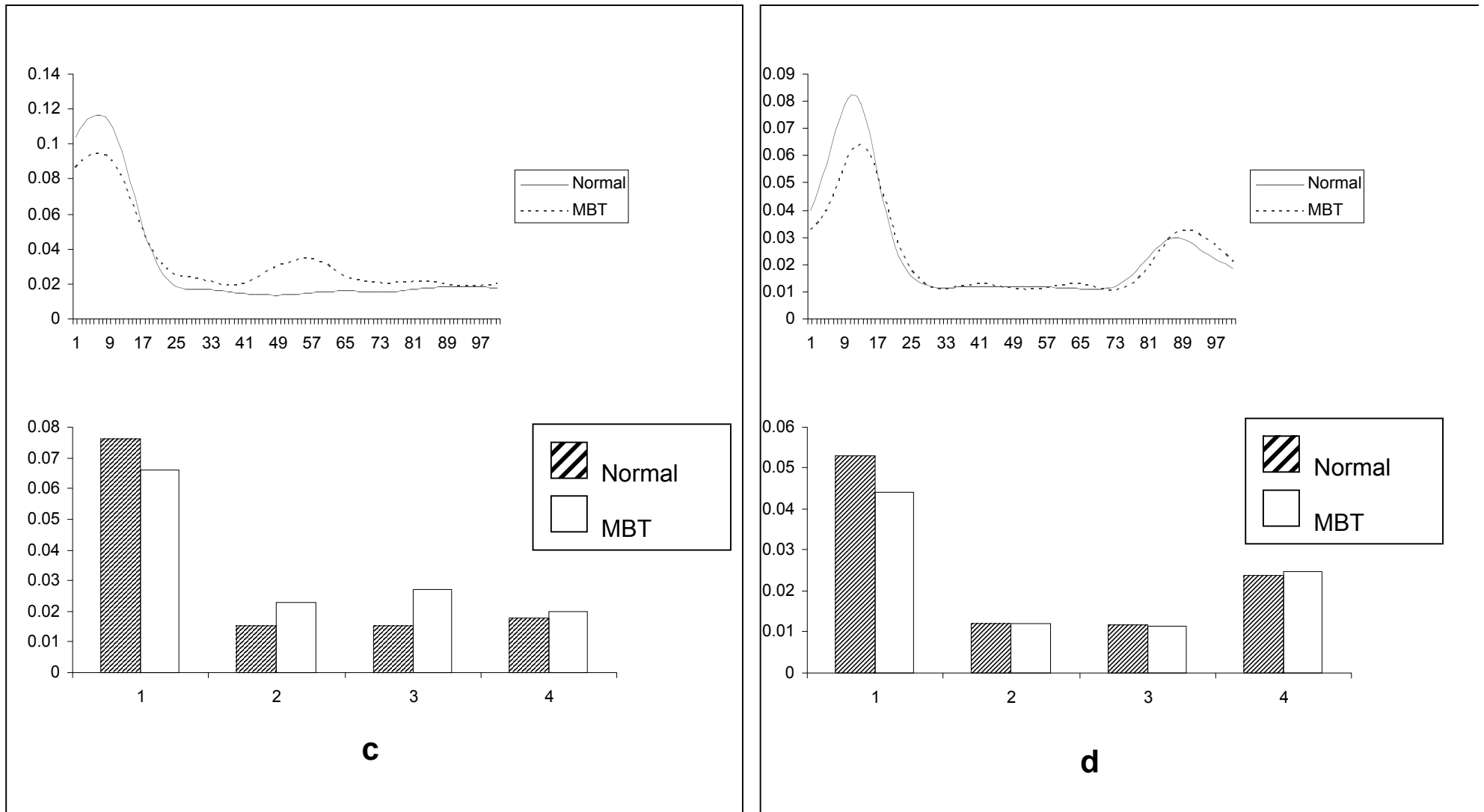


Figure 3.8: Representative and average over four phases of the stance phase, electrical activity at the gastrocnemius (a); hamstring (b); gluteus maximus (c); multifidus (d) of normal and MBT gait.

5.0 Summary

The findings of the current study suggest that certain gait characteristics are affected during walking in a shoe of unstable construction.

Although many of the kinematic variables remained unchanged, there were significant differences in both the trunk and the ankle angles. MBT shoes promote less forward lean during locomotion suggesting a more upright posture. Although not confirmed by the present study, the probable effect is a shift in the centre of mass position closer to the centre of the base of support, possibly aligning the body more optimally for locomotion. This, in conjunction with the lower hip moments experienced in the MBT condition may suggest reduced loading of the lower back. The changes in the ankle plantar flexion / dorsiflexion angle at the ankle, was primarily due to the reduced plantar flexion following initial contact during the MBT condition when compared to the normal condition.

Kinetics at each of the joints of the lower limb were different between the MBT and normal conditions. The lower moments experienced at these joints suggests a resultant decrease in joint loading.

The major finding from analysis of the ground reaction force data is the suggestion that there is a higher incidence of transient peaks when wearing normal shoes compared to MBTs. There is some evidence suggesting that transient forces transmitted through the skeleton are the primary aetiological factor in the development of many musculoskeletal disorders. These include, osteoarthritis, stress fractures, plantar fasciitis and achilles tendonitis and low back pain (Whittle, 1999).

Muscular recruitment during MBT gait is also altered, eliciting an increase in motor unit recruitment for Gastrocnemius, Biceps Femoris and Gluteus Maximus and a decrease in Multifidus. The increased activity in the muscles of the lower limb can be attributed to differences in the mechanics of MBT gait, i.e. more upright posture and the production of greater propulsive forces during the dragging motion of the stance limb. Although not confirmed by this study, the decrease in multifidus

recruitment may suggest that MBTs reduce the possibility of fatigue in the lower back.

This initial study into the effects of an unstable shoe construction suggests that MBTs alter certain gait characteristics and that with frequent use they may reduce the incidence of some musculoskeletal problems. In those already suffering from such disorders, MBTs may allow patients to remain mobile by reducing cyclic loading of the already damaged joint.

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APPENDIX A

TABLES

MBT gait evaluation.

	Hip (deg)	Knee (deg)	Ankle (deg)	Pelvis (deg)	Trunk (deg)
MBT	38.19 (8.40)	12.42 (7.71)	11.43 (4.95)	11.14 (5.96)	-3.27 * (3.79)
Normal	38.79 (8.79)	9.83 (7.70)	9.77 (5.83)	11.51 (6.42)	-1.99 * (3.94)

Table A1: sagittal joint angles (mean \pm SD) during walking for the normal and MBT conditions at foot strike (* denotes statistically significant difference between conditions).

	Hip (deg)	Knee (deg)	Ankle (deg)	Pelvis (deg)	Trunk (deg)
MBT	-0.72 (7.55)	66.08 (9.55)	10.78 (6.87)	3.01 (5.66)	0.54 * (3.86)
Normal	-0.87 (8.21)	63.50 (8.97)	10.40 (6.66)	3.46 (5.86)	1.51 * (3.81)

Table A2: sagittal joint angles (mean \pm SD) during walking for the normal and MBT conditions at toe off (* denotes statistically significant difference between conditions).

MBT gait evaluation.

	Hip (deg)	Knee (deg)	Ankle (deg)	Pelvis (deg)	Trunk (deg)
MBT	38.91 (8.55)	45.45 (9.18)	28.51 * (5.65)	12.10 (5.97)	3.37 (3.46)
Normal	39.47 (8.81)	45.11 (9.41)	26.31 * (5.99)	12.20 (6.38)	3.83 (3.46)

Table A3: Maximum sagittal joint angles (mean \pm SD) during walking for the normal and MBT conditions (* denotes statistically significant difference between conditions).

	Hip (deg)	Knee (deg)	Ankle (deg)	Pelvis (deg)	Trunk (deg)
MBT	-8.84 (7.17)	7.38 (7.36)	6.74 * (5.45)	2.19 (5.42)	-4.09 * (3.76)
Normal	-8.90 (8.07)	4.44 (8.31)	4.20 * (4.72)	2.69 (5.78)	-2.93 * (4.06)

Table A4: Minimum sagittal joint angles (mean \pm SD) during walking for the normal and MBT conditions (* denotes statistically significant difference between conditions).

MBT gait evaluation.

	Hip (Nm/bw)	Knee (Nm/bw)	Ankle (Nm/bw)
MBT	0.78 (0.59)	-0.39 (0.33)	0.04 (0.08)
Normal	0.92 (0.78)	-0.41 (0.32)	0.04 (0.08)

Table A5: sagittal joint moments (mean \pm SD) during walking for the normal and MBT conditions at foot strike (* denotes statistically significant difference between conditions).

	Hip (Nm/bw)	Knee (Nm/bw)	Ankle (Nm/bw)
MBT	0.21 (0.89)	-0.22 (0.29)	-0.01 (0.10)
Normal	0.60 (0.93)	-0.32 (0.29)	-0.03 (0.10)

Table A6: sagittal joint moments (mean \pm SD) during walking for the normal and MBT conditions at toe off (* denotes statistically significant difference between conditions).

MBT gait evaluation.

	Hip (Nm/bw)	Knee (Nm/bw)	Ankle (Nm/bw)
MBT	8.06 * (4.66)	-0.06 (0.37)	0.96 * (0.57)
Normal	12.41 * (5.68)	0.32 (1.17)	1.26 * (0.50)

Table A7: Maximum (extension / plantar flexion) sagittal joint moments (mean \pm SD) during walking for the normal and MBT conditions (* denotes statistically significant difference between conditions).

	Hip (Nm/bw)	Knee (Nm/bw)	Ankle (Nm/bw)
MBT	-0.67 (1.27)	-2.18 * (1.33)	-0.34 * (0.35)
Normal	-1.97 (4.33)	-2.99 * (1.53)	-0.73 * (0.46)

Table A8: Minimum (flexion / dorsiflexion) sagittal joint moments (mean \pm SD) during walking for the normal and MBT conditions (* denotes statistically significant difference between conditions).

MBT gait evaluation.

	\bar{fz}_i (N)	\bar{fz}_a (N)	\bar{fy}_{max} (N)	\bar{fy}_{min} (N)
MBT	840.07 * (164.54)	789.32 (110.05)	42.58 (10.37)	-45.60 * (12.04)
Normal	812.09 * (180.07)	785.59 (129.27)	41.31 (10.79)	-42.60 * (11.90)

Table A9: Ground reaction force variables (mean \pm SD) during walking for the normal and MBT conditions, where \bar{fz}_i = impact peak; \bar{fz}_a = active peak; \bar{fy}_{max} = maximum anterior force; \bar{fy}_{min} = maximum posterior force. (* denotes statistically significant difference between conditions).