Effect of Surface Hardness on Three-Segment Foot Kinematics during Barefoot Running

N. A. A. Abdul Yamin[1], K. S. Basaruddin*[1], W. M. R. Rusli[1], A. F. Salleh[1], N. A. Razak[1] and W.Z.A.W. Muhamad[2]
[1]Biomechanics & Biomaterials Research Group, School of Mechatronic Engineering, Universiti Malaysia Perlis, 02600, Pauh, Perlis, Malaysia
[2]Institute of Engineering Mathematics, Universiti Malaysia Perlis, 02600, Pauh, Perlis, Malaysia
*corresponding author
Email: khsalleh@unimap.edu.my, Phone: +604-9885280, Fax: +604-9885167

***Abstract*-- This paper presents an experimental study on the effect of different surface hardness to the foot kinematics during barefoot running using a multi segment foot model. Joint rotations during the stance phase of nine male subjects that running barefoot on three types of surface with different hardness level (concrete, artificial grass and rubber) was investigated experimentally. Screening process was conducted by evaluating the foot strike pattern among the subjects in order to eliminate the influence of the foot strike on kinematics response. Only heel strikers' data was analyzed since most of the subjects performed heel strike during the experiment. Differences on joint rotation due to surface effects were analyzed using visual 3D software. The result showed that the pattern of joint rotation was slightly different when barefoot running on artificial grass compared to rubber and concrete surface. The significant difference was found in further investigation of the joint rotation during the mid-stance of the stance phase. The joint rotation was varied in the varying of the surface conditions, meanwhile group-based analysis in the present study did not imply the significant difference among the surfaces. No significant correlation was found in the present experiment between various surface hardness and joint rotations of foot kinematics segments. This study could provide additional insight on relationships between foot segment kinematics in term of joint rotations when running on different surface hardness.**

***Index Term*-- running, surface hardness, foot kinematics, joint rotations.**

1. Introduction

Barefoot running can be regarded as the most natural form of exercise without any equipment or a specific environment. Barefoot running is believed will be able to prevent injury in term of kinetics adaptation by minimizing the impact peaks and increased the foot strength [1]. However, barefoot running in various environment either urban surroundings or a nature trail required adaptation and fast response to disturbances [2]. These adaptations and response are crucial in order to obtain a full benefit of health from running activity as well as to prevent any running-related injuries [3].

Running also necessitates interaction of all human body system. The interaction between parts and alteration part of the body during running can be monitored and analyzed using kinematic measurement as investigated by previous studies [4]–[6]. Different running surface conditions basically produced different kinematics response. For instance, posture of the leg during ground contact on the hard running surface required maximum extension while maximum knee flexion remained unchanged [7], [8]. In addition, decreased in ankle, knee, and hip flexion velocities occurred on softer running surfaces [9]. Therefore, running surface had been recognized as one of the major factors that contribute to an increase of injuries incident [10].

The foot kinematics response of running on different surfaces has been previously examined by several researchers [11]–[13]. Fu et al. [14] studied the effects of running on different surfaces on the characteristics of in-shoe plantar pressure and tibial acceleration. No significant differences were found in the plantar pressures, pressure–time integral and peak pressure distribution for the concrete, synthetic, grass, and normal treadmill surfaces. Similarly, Tessutti et al. [15] investigated the influence of running on asphalt, concrete, natural grass, and rubber on in-shoe pressure patterns in adult recreational runners. The results showed that among the more rigid surfaces (asphalt and concrete), there were no differences in the pressure patterns and similar behavior was observed on the rubber surface. Moreover, investigation on how mid sole hardness, surface stiffness, and running duration influence running kinematics has been conducted by Hardin et al. [9]. They found that an increase of midsole hardness resulted in greater peak ankle dorsiflexion velocity meanwhile increase of surface stiffness resulted in decreased hip and knee flexion on contact, reduced maximal hip flexion, and increased peak angular velocities of the hip, knee, and ankle. This study indicates that lower-extremity kinematics was adapted to increase midsole hardness, surface stiffness, and running duration.

In general, most of the kinematic variables that were measured in the previous studies regarding to the effect of running surface characteristics are the angle of hip [9], knee [7], [16], [17] and ankle [9], [16], [17]. Kinetic variables such as the impact force [16] and ground reaction force [18] were also included in combination with other variables to evaluate the angular velocities [9], initial and peak joint angle [16]. However, the foot model was tested as a single rigid body segment. Investigation on alteration of foot kinematics during running can be precisely predicted using a three dimensional multi segment foot model. The lack of a single rigid segment of foot model is inability to provide adequate information on deformation of the foot that may cause injury to runner [19]. To date, a number of studies have used a multi segment foot model to investigate the kinematics response influenced by surface inclines [20], slopes [21], [22]. However, there is no known research that has directly measured the effect of surface hardness on three dimensional multi segment foot kinematic during running. Thus, the aim of this study is to consider the effect of surfaces with different hardness on the foot kinematics during stance phase of barefoot running based on multi-segment foot model.

2. Materials and Methods

2.1 Subject

Nine healthy male recreational runners at the age of 24 ± 1.2 years old and having the normal body mass index (BMI) category were participated in this study. Their height and weight are 172 ± 2.7 cm and 67 ± 6.7 kg respectively. Individual with recently musculoskeletal injury or orthopedic abnormality were not part of this study due to dissimilarity in the movement and potential difficulty in performing the task. Each subject that filled the survey form was voluntarily consented to participate in the study and signed the consent form prior to participation.

2.2 Instruments and equipment

Experimental work was held in Biomechanics Laboratory at Universiti Malaysia Perlis. Three-dimensional foot segments kinematics data were obtained using five Oqus cameras in Motion Captured System at the frequency of 200 Hz with two Bertex force plates. Ground reaction force (GRF) is not discussed in the present study because it was only used to establish the stance phase, which is between foot strikes to toe-off. The cameras were conducted in a position that could captured all the twelve markers during the stance phase of running gait. The distance between the cameras and markers will not affecting the measured data. The markers were plastic sphere with 20 mm and 15 mm diameter covered by reflective tape. The track dimension used was 10 m long and 1 m wide and placed over the force plate. The arrangement of instruments and equipment involved is given in figure 1. The foot segment kinematics responses in this study needed to be measured using Qualysis Track Motion (QTM), Visual 3D and Statistical Package for the Social Sciences (SPSS) software.

Three different running surfaces employed in this study were concrete, artificial grass, and rubber. The cushioning property of each surface was obtained by conducting a simple experiment according to American Society of Testing and Materials (ASTM) F2117-10, that was also been used by Fu et al. [14]. Based on the ASTM standard method, a basketball was vertically dropped from a height of 2 m on each surface to obtain the vertical rebound characteristic that represents the cushioning property. Three trials were recorded and the average rebound height was calculated to find out the surface hardness as listed in Table 1. A higher rebound height indicates a harder surface. The concrete was found the hardest surface whereas artificial grass was harder than rubber surface.

2.3 Experimental protocols

Foot segments kinematics response was analyzed using multi segment foot modelling, based on the marker placement that was invented by Leardini et al. [19]. Twelve reflective markers were placed on the right foot of each subject as shown in Figure 2. The anatomical landmark is divided into three segments; metatarsus, mid-foot and calcaneus. The metatarsus segment considered the base of the first metatarsal, the head of the first metatarsal, the base of the second metatarsal, the head of the second metatarsal, the base of the fifth metatarsal and the head of the fifth metatarsal. As for mid-foot segment landmark involved the most medial apex of the tuberosity of the navicular, whereas the upper central ridge of the calcaneus posterior surface, medial malleolus and lateral malleolus were the regions for calcaneus segment.

The markers were mounted on the anatomical landmark using double-sided adhesive tape. Subjects were first ran on the runway prior data collection to familiarize with each condition of the experiment. Subjects then were asked to run over 10 m indoor running surfaces (rubber, concrete and artificial grass) at their comfortable speed that reflects to recreational run. A static trial was also recorded with the subject stand upright in double-leg support posture in order to determine the neutral position of the joint as referred in [23]. Data collection of subject running on all surfaces were then carried out. Trials were accepted if all markers position were well captured and the right foot contacted with the force plate without obvious alterations to the run stride. Ground reaction force was utilized to determine the stance phase, i.e. the touch-down and the toe-off samples. While the mid-stance phase was identified using visual analysis based on the centre of mass arrow displayed in the QTM software.

2.4 Data analysis

The 3D foot segment kinematic positions and orientations were processed and analyzed using Visual 3D. Joint rotations at each running surface acquisition were smoothed and filtered using low pass filter at frequency 12 Hz [24]. The 3D joint rotation was calculated based on International Society of Biomechanics (ISB) recommendation as referred to [19]. It was used to describe the joint rotations terminology such as dorsi/plantar-flexion, abduction/adduction and eversion/inversion as rotation in z-axis, y-axis and the axis orthogonal to both z-axis and y-axis respectively. The joint rotation angle of the foot during running was analyzed based on the relative angle between the segment of mid-foot with respect to the calcaneus, metatarsus with respect to not-adjacent to calcaneus and metatarsus with respect to mid-foot.

Screening process on foot strike pattern performed by subjects was done in order to get rid influence of other variables instead of surface hardness. Subjects' foot strike pattern was evaluated based on the angle of incidence (AOI) which is the angle between the horizontal plane and the line formed by the fifth metatarsal head and lateral malleolus. Determination of foot strike using AOI was measured according to the procedure used by Miller et al. [25]. If AOI equal to 0^{o}, it indicates the mid-foot strike (MFS), if more than 0^{o} means the forefoot strike (FFS) whereas if less than 0^{o} means the rear foot strike (RFS). Data of AOI in foot strike were normalized using the AOI in standing posture. The foot strike pattern was then further verified using visual analysis in the QTM software to ensure the consistency. Since there are three possible foot strike patterns could be performed by the participants, only the majority group was further analyzed in kinematic adaption of foot.

Moreover, in order to evaluate the statistical significance among the surfaces, the measured joint rotation of the foot segment for mid stance was analyzed using SPSS software. Normality of the data was checked in order to identify the suitable method for statistically analysis. Then, the significant difference of foot segment in joint rotation during running on each surface was accepted if p value is less 0.1. Statistical analysis using one way Kruskal-Wallis was also included in the surfaces influence on the stance time, plantar fascia strain, peak medial longitudinal angle (MLA) and MLA relative to range of motion (ROM). Plantar fascia strain was measured as the change of length between the first metatarsal and calcaneus markers at the peak of the stance phase divided by the original length [20]. The MLA was quantified as the angle between the lines from the calcaneus marker to the navicular tuberosity and from the first metatarsal to the navicular tuberosity [26].

3. RESULTS

3.1 Foot strike patterns

Table II provides the result of foot strike for each subject during running on three surfaces based on the calculated AOI. It is apparent from this table that most of the subjects performed heel strike for all surfaces. Only a few cases that the subjects performed forefoot strike whereas none performed mid foot strike in the stance phase. However, further visual analysis using QTM showed that a few data obtained from the calculation were contradicted to the actual visual analysis. Based on the visual analysis, subjects 1 and 5 were actually performed forefoot strike when running on the concrete surface meanwhile subject 3 was actually performed heel strike when running on rubber surface, as indicated by an asterisk (*). Results of visual analysis seem more reliable compared to the calculation because the position of foot during strike was clearly visualized in QTM. Hence, only the subjects with consistently performed the heel strike at the stance phase were further analyzed in the joint rotation of foot segment kinematics.

3.2 Effect of running surface on the joint rotation during stance phase

Analysis on the joint rotations was restricted to the data of subjects who performed the heel strike during the stance phase. Figure 3 displays the mean value and standard deviation for three segments of joint rotations during the stance phase for three surfaces (ie. concrete, rubber, artificial grass). The standard deviation zone that indicated by the dotted line around the mean value (highlighted solid line), represents the dispersion of measurement and subject variability.

At the glance, the pattern of 3D rotation for the Calcaneus-Metatarsal joint was found almost similar for each surface. In Figure 3(a), there is a clear trend that Calcaneus-Metatarsal joint segment in the coronal plane was the least inverted on the concrete surface compared to the rubber and artificial grass surface. There were no significant differences between rubber and artificial grass surface on eversion/inversion motion of Calcaneus-Metatarsal joint segment. Moreover, there were no differences in rotation of Calcaneus-Metatarsal joint segment in transverse and sagittal plane among all surfaces as seen in Figure 3(b) and 3(c), respectively. Response for all surfaces showed Calcaneus-Metatarsal joint segment was more adducted in transverse plane meanwhile more plantarflexed in the sagittal plane.

Similarly, the trend of joint rotation for Calcaneus-Mid foot in the coronal plane during stance phase was also found not sensitive to the surface type as shown in Fig. 3(d). All surfaces contributed to the inversion motion for joint rotation of Calcaneus-Mid foot segment. Interestingly, in Fig. 3(e), the artificial grass was observed contributed to the abduction motion in the transverse plane, whereas both rubber and concrete showed adduction in foot motion. It seems that the surface hardness did not affect the joint rotation angle since in term of hardness magnitude, the artificial grass is in between concrete and rubber surfaces. In contrary, pattern of joint rotation for Calcaneus-Mid foot in the sagittal plane was found similar among all surfaces as implied in Fig. 3(f). Result indicated slight reduced dorsiflexion during the stance phase.

In Mid foot-Metatarsal joint segment, trend of angle of rotations in the coronal and sagittal plane was found similar among all surfaces as shown in Figs. 3(g) and 3(i), respectively. In the coronal plane, the Mid foot-Metatarsal joint demonstrated an inversion motion whereas plantarflexion motion was found in the sagittal plane. Surprisingly, there is an obvious difference was noticed in Fig. 3(h) where the artificial grass contributed to adduction motion compared to the rubber and concrete that contributed to abduction motion. Again, no correlation between surface hardness and joint rotation angle was noticed from this result.

3.3 Comparison on joint angle rotation at mid-stance phase for each running surface

Table 3 presents the difference in joint rotations during running on each surface at mid-stance of stance phase. Since the data are non-parametric, the statistical analysis using Kruskall Wallis was employed. As highlighted in the Table 3, analysis on the comparison of joint rotation at mid-stance of stance phase between the different running surfaces revealed that there were significant difference in the transverse plane for joint segments of Calcaneus-Mid-foot and Mid foot-Metatarsal respectively. Obviously, the differences are due to artificial grass having abduction and adduction motion for joint segments of Calcaneus-Mid-foot and Mid foot-Metatarsal respectively, whereas rubber and concrete were vice versa. On the other hand, no significant difference was found for other communal rotation motion for all surfaces.

3.4 Effect of running surface on temporal parameters, plantar fascia strain and MLA

Table IV lists the effect of running surface types on stance time, plantar fascia strain, peak MLA and MLA relative ROM. Slightly decreasing in stance time was found with the increase in surface hardness. Plantar fascia strain was highest when barefoot running on artificial grass. Meanwhile, peak MLA was marginally lower in rubber compared to artificial grass and concrete. However, no significant different ($p > 0.1$) was observed among surfaces in stance time, plantar fascia, peak MLA and MLA relative to ROM under present statistical analysis.

4. DISCUSSION

This study is particularly focusing on the kinematics of three foot segment during the stance phase of barefoot running when the foot is in contact with the running surface. The current study provides evidence that no correlation between varying surface hardness (rubber, artificial grass and concrete) and multi-segment foot kinematics during barefoot running. Although significant differences in inversion angle of Calcaneus-Metatarsal segment and abduction/adduction angle of Calcaneus-Mid-foot segment were observed, the difference might be due to the different types of running surface rather than surface hardness. A possible explanation for this might be due to the difference in the level of surface hardness based on the rebound height as listed in Table 1 is not so obvious. Hence, the kinematics adaption is not sensitive to the slight difference on surface hardness.

The Kruskal-Wallis (one way) showed that most of the kinematic variables obtained in the present study were not statistically significant. These results seem to be consistent with other research [16] which found no significant different in kinematic variables when running on the conventional asphalt, rubber-modified asphalt and acrylic sport surface based on a single foot segment. Dixon et al. [16] highlights a varied individual response was observed in initial joint angles, peak joint angles, and peak joint angular velocities with respect to the running surface but not in group data analysis. Surface effect on group analysis usually is hardly obtained owing to the unique adaptation strategies applied by each subject and the missing of sufficient statistical power [3]. The running different strategy applied by each subject can be observed during foot strike of running gait. Foot strikes patterns employed by subjects in the present study varied according to the AOI calculation. The findings agree with Dufek et al. [27] which reported that various performance of individual subjects in group analysis can affect each other result despite the fact that each subject results is valid. Therefore, this study had eliminated data of subject who performed fore-foot and mid-foot strike, and data remaining to be analysed were only heel strikers' data. This approach minimizes the existence of uncertainty factors other than surface effect that could influence the result of the present study.

On the other hand, running on the artificial grass surface in the current study showed slightly different in joint rotation angle with respect to abduction-adduction motion compared to rubber and concrete surfaces. This discrepancy may be caused by kinematic adjustment of the lower extremity that took placed in adapting with surface condition during running. The results of present study also suggest that plantar fascia strain, temporal parameters and MLA not correlated to the surface hardness since the increased of hardness (rubber, artificial grass and concrete, in sequence) did not contribute to the obvious increased or decreased of those parameters. This relationship is contradictory to the influence of surface incline on the kinematic variables during running [20]–[22]. Sinclair et al. [20] reported that there are significant different between flat surface and surface with incline more than 10° in a plantar fascia strain, stance duration and ROM during treadmill running.

Rotation of joint angle was included as part of kinematics parameter which is the geometric description of motion that record the position and orientation of the body segments. Previous studies generally investigated the surface effect on foot kinematics using a two-dimensional system by measuring the ankle joint rotation which foot was treated as a single rigid body [9], [16]. In order to achieve an accurate and a detailed measurement, a calibrated 3D system is essential where measurement from more than one viewpoint. In the present study, a multi segment analysis was selected to the evaluation method for 3D foot kinematics. The multi segment analysis is another approach that can be utilized in dynamic modeling other than single rigid body analysis to define a segment of the body part using optical tracking equipment. This equipment that investigates gait in term of dynamic modeling is shown as an efficient method for measuring 3D kinematics and kinetics of human body [28]. Multi-segment analysis can be regarded as an extended method which was also selected to overcome single rigid foot segment assumption that applied by most of the previous studies [19], [29]–[31][17]. Moreover, the multi segment analysis performed an accurate evaluation as well by analyzing the relation of each appointed segment during motion and offer constructive awareness of segmental foot kinematics. Accurate foot kinematics measurement is substantial in order to determine foot function during running and treat foot dysfunction as it may contribute to injury or continuous pain.

Mid-stance where the period between heel rise and toe-off of the stance phase is considered as a crucial phase in the gait cycle. Among all phases in the running gait cycle, mid-stance was particularly investigated due to its contribution to the injury risk. The present study shows that there are significant differences on joints rotation in the transverse plane of Calcaneus-Mid foot and Mid foot-Metatarsal joint segment during mid-stance. In mid-stance phase, all body weight is borne by a single leg as the other foot is in the swing phase. Thus, a lower limb is especially exposed to injury. In addition, foot and leg enable the body weight to pass over and stop pronating. There are too much movement and instability that may lead to falling if the foot still pronating. Therefore, an adjustment is needed to prevent any excessive pronation that contributes to injury.

5. Conclusion

The present study was undertaken to determine the effect of different surface hardness on foot kinematic during barefoot running using a multi segment analysis. This study has found that generally in stance phase, both rubber and concrete surfaces produced almost similar pattern of joint rotation of the foot segment with certain range of angle but slightly different compared to the artificial grass surface. However, further investigation on mid-stance of stance phase shows significant difference of joint rotations among these surfaces in transverse plane for Calcaneus-Mid foot and Mid foot-Metatarsal joint segment. These data suggest that foot kinematic might be achieved through individual-based analysis as this analysis might provide different outcome instead of result obtained from group-based analysis. No correlation among running surface hardness on joint rotation angle, stance time, plantar fascia strain and MLA was noted in the present experiment. This study could provide additional insight on knowledge of relationships between the multi segment foot kinematics and running surfaces in order to predict injury risk for runners.

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FIGURES

Fig. 1. Layout of instrument and equipment for experiment.

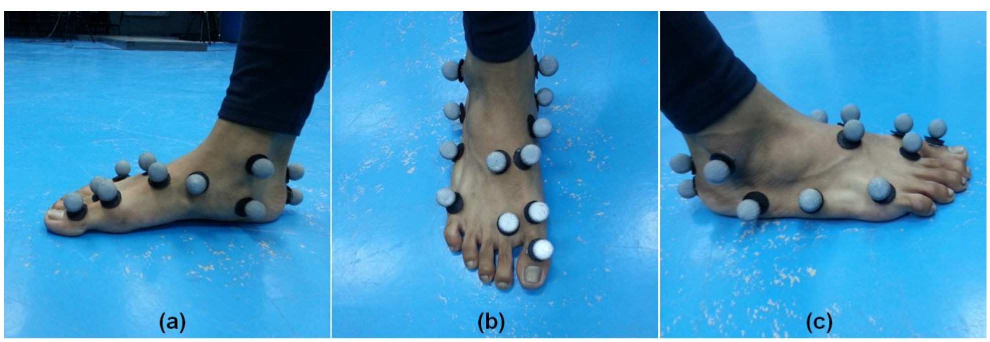

Fig. 2. Marker placement in; (a) medial view, (b) frontal view and (c) lateral view.

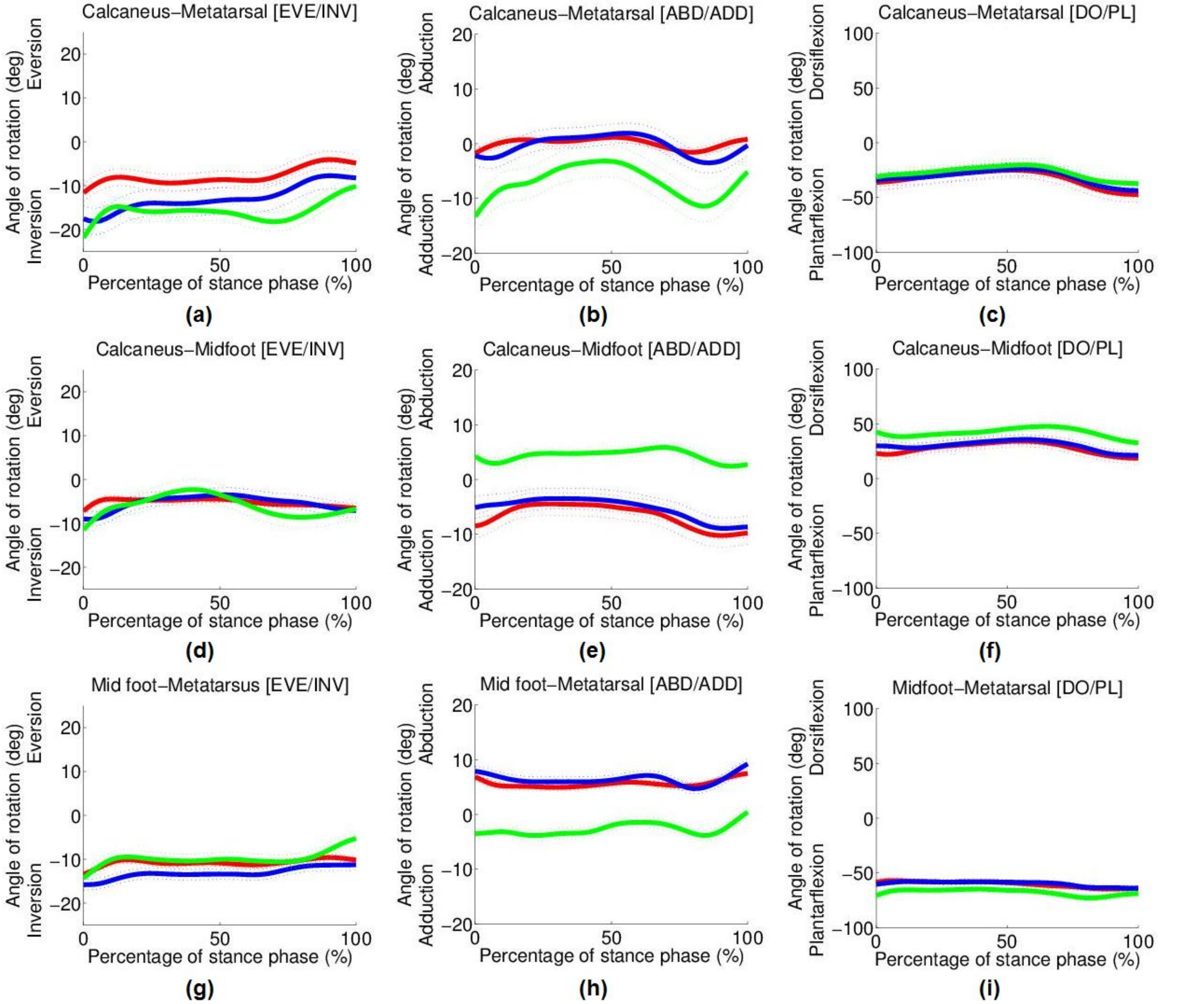

Fig. 3. Angle of joint rotation in foot segment: (a) Calcaneus-Metatarsal (Eversion/Inversion), (b) Calcaneus-Metatarsal (Abduction/Adduction), (c) Calcaneus-Metatarsal (Plantar/Dorsi Flexion), (d) Calcaneus-Mid foot (Eversion/Inversion), (e) Calcaneus-Mid foot (Abduction/Adduction), (f) Calcaneus-Mid foot (Plantar/Dorsi Flexion), (g)Mid foot-Metatarsal (Eversion/Inversion), (h) Mid foot-Metatarsal (Abduction/Adduction), and (i) Mid foot-Metatarsal (Plantar/Dorsi Flexion). *Note: Red line=concrete, Blue line=rubber, Green line=artificial grass.*

TABLES

Table I
Comparison of ball rebound height for three different surfaces.

Surface	Rebound height (cm)			
	Trial 1	Trial 2	Trial 3	Mean
Rubber	77.43	84.99	77.49	79.97±4.4
Artificial Grass	94.51	100.01	98.88	97.80±2.9
Concrete	104.59	99.08	105.44	103.04±3.5

Table II
Determination of foot strike.

Subject	**Concrete**			**Artificial grass**			**Rubber**		
	Static, Θ_S (^{O})	Running, Θ (^{O})	Foot strike	Static, Θ_S (^{O})	Running, Θ (^{O})	Foot strike	Static, Θ_S (^{O})	Running, Θ (^{O})	Foot strike
S1	16.439	15.831	Heel*	18.090	16.544	Heel	17.178	26.895	Forefoot
S2	22.395	18.066	Heel	16.036	21.002	Forefoot	18.921	2.847	Heel
S3	28.014	32.107	Forefoot	19.482	7.482	Heel	17.178	26.895	Forefoot *
S4	19.041	9.593	Heel	19.065	11.961	Heel	16.265	0.184	Heel
S5	17.965	16.885	Heel*	13.787	22.787	Forefoot	20.057	17.551	Forefoot
S6	17.109	15.945	Heel	13.058	6.998	Heel	15.431	3.166	Heel
S7	22.125	5.242	Heel	18.072	22.207	Forefoot	20.983	1.492	Heel
S8	17.311	4.640	Heel	19.526	7.328	Heel	18.159	5.280	Heel
S9	17.420	16.088	Heel	19.275	13.073	Heel	19.474	17.195	Heel

Note: $\Theta > \Theta_S$ forefoot strike, $\Theta < \Theta_S$ heel strike, $\Theta = \Theta_S$ mid-foot strike

Table III
Angle of joint rotation at mid stance of stance phase with respect to surface types.

	Angle of Rotation	**Rubber**		**Artificial Grass**		**Concrete**		***p*-value**
		Mean	SD	Mean	SD	Mean	SD	
Calcaneus - Metatarsal	Eversion/Inversion (^{O})	-14.028	3.615	-16.105	5.573	-8.967	5.651	**0.097**
	Abduction/Adduction (^{O})	1.123	5.828	-3.427	5.270	0.594	4.441	0.548
	Dorsiflexion/Plantarflexion (^{O})	-26.299	7.598	-21.269	12.701	-31.845	19.138	0.854
Calcaneus – Mid foot	Eversion/Inversion (^{O})	-4.241	5.560	-2.736	5.584	-4.677	3.167	0.692
	Abduction/Adduction (^{O})	-3.739	4.382	4.938	3.693	-4.745	5.833	**0.079**
	Dorsiflexion/Plantarflexion (^{O})	32.899	12.749	43.786	12.430	32.163	4.591	0.587
Mid foot - Metatarsal	Eversion/Inversion (^{O})	-13.350	5.531	-10.470	4.917	-10.809	3.543	0.751
	Abduction/Adduction (^{O})	6.296	9.153	-2.911	7.963	5.218	4.964	0.529
	Dorsiflexion/Plantarflexion (^{O})	-58.016	10.321	-64.756	3.145	-58.052	11.461	0.470

Table IV
Stance time, plantar fascia strain and MLA angle with respect to surface types.

	Rubber		**Artificial Grass**		**Concrete**		***p*-value**
	Mean	SD	Mean	SD	Mean	SD	
Stance time ($x10^{-3}$ s)	257.1	39.2	254.0	23.0	243.3	20.2	0.822
Plantar fascia strain ($x10^{-3}$)	114.023	36.0	132.121	58.5	103.734	27.3	0.738
Peak MLA (^{O})	167.140	5.66	170.406	2.88	170.412	6.40	0.414
MLA relative ROM (^{O})	5.034	3.51	4.876	3.19	7.505	3.98	0.331