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A foot joint and muscle force assessment of the running stance phase whilst wearing normal shoes and bionic shoes

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Purpose: The aim of this study was to investigate the differences in ankle joint parameters of biomechanics changes between the normal shoes (NS) and the bionic shoes (BS) during the running stance phases. *Methods*: A total of 40 Chinese male runners from Ningbo University were recruited for this study (age: 22.3 ± 3.01 years; height: 174.67 ± 7.11 cm; body weight (BW): 66.83 ± 9.91 kg). The participants were asked to perform a running task. Statistical parametric mapping (SPM) analysis was used to investigate any differences between NS and BS during the running stance phases. The principal component analysis (PCA) and support vector machine (SVM) were used to further explore the differences of the muscle force between the BS and NS. *Results*: Significant differences ($p < 0.05$) were found in the first metatarsophalangeal joint (MPJ1), ground reaction force (GRF), ankle joint and around muscle forces. Furthermore, the accuracy of SVM model in identifying the gait muscle force between BS and NS reached 100%, which proved that the BS had a very large impact on the gait muscle force compared with NS. *Conclusions*: We found that BS may be better suited to the human condition than other unstable shoes, or even NS. In addition, our results suggest that BS play an important role in reducing ankle injuries during running by increasing muscle participation in unstable conditions while better restoring the most primitive instability of foot condition that humans have.

Key words: *running gait, bionic shoes, muscle force, machine learning method*

1. Introduction

The fundamental function of shoes is to preserve human feet while also providing postural stability throughout daily tasks [43]. However, a part of researchers has demonstrated that the function of traditional shoes might contribute to overprotection in people [17], [33]. According to the consideration of early human evolution, there is strong evidence that barefoot walking occurred in the early stages of human life [3], [44]. Even now, various indigenous groups walk or run barefoot, without any footwear. This implies that, from a necessary standpoint, shoes are not required for human existence or regarded as a priority. Currently, research shows that conventional shoes have an effect on lower limb function, causing gradual degeneration, such as loss of lower limb muscle force, loss of lower limb balance and increasing lower limb injuries. For instance, this overprotection may impair muscular strength. Nigg and Sousa also showed that this overprotection may result in

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possible injuries while executing the activity [32], [34], [48].

With the evolution of human beings and the use of shoes over many years, the cuticle of the foot has not enabled humans to return to the state of barefoot walking. Therefore, Reflex control (RC) shoes and the Masai Barefoot Technology (MBT) have sparked significant research interest based on concerns of unstable shoes [54]. A previous study indicated that after six weeks of training in both NS and MBT, there were no significant gains found in balance while wearing the MBT shoes. The instability conditions of the MBT structure are close to the instability of ankle dorsiflexion and plantarflexion [67]. These unstable shoes did not actually reflect the real unstable condition that humans have. Based on this consideration, we created BS by combining real barefoot conditions and shoe protection features. The primary difference between BS and other unstable shoes is that their instability is exclusively developed based on the posture or motion condition of the human body when running or walking. Our BS, on the other hand, combine their benefits and incorporate the barefoot form, which more accurately reflects and restores the condition of barefoot running or walking.

Running has grown in popularity as a leisure activity in recent years. With up to 79% of runners suffering from musculoskeletal injuries each year, there are several variables that contribute to the etiology of the running-related ailments reported [56], [57]. Given the significant number of running-related injuries, medical professionals and academics are focusing their efforts on figuring out how to treat and prevent them [13], [20], [21], [49], [64]. Therefore, several scientific researches have examined the effect of shoe modifications on running style since most individuals wear running shoes when they go for a run. Unshod groups have different foot architecture (e.g., more sphaodilayed toes) than habitually shod ones, according to several sources of evidence (e.g., more evenly-distributed plantar pressure) [1], [2], [6], [16]. We have to admit that the biomechanics parameters can actually reflect the changes of lower limbs for researchers to better understand what types of potential injuries could impact athletes. However, the question is whether are those methods sufficient for the clinical judgments. Because of these considerations, the deep study methods have been caught researcher's visions, such as muscle force and multiple foot model which used to reflect the condition of changes in metatarsophalangeal.

Ankle sprains are the most prevalent ailments, when it comes to sprinting and leaping sports like basketball and soccer [39]. Researches indicate that up to 40% of ankle sprain victims continue to experience residual impairment for up to seven years after the initial event [27], [46]. Freeman invented the phrase "Functional ankle instability" to describe the subjective sense of giving way or experiencing joint instability after a series of ankle sprains. Factors that contribute to functional ankle instability are diverse, but they include sensory, mechanical and muscular deficits [15], [24], [28]. According to previous studies, it is possible to analyze muscular force in order to understand the underlying mechanisms of damage in order to identify preventative measures and devise training strategies [51], [63], [67]. Rachel have summarized that one of the main functions of medial the gastrocnemius (MG), lateral gastrocnemius (LG) and soleus (S) are to assist the ankle joint during the performance of dorsiflexion and plantarflexion [30]. As for tibialis anterior (TA), peroneus longus (PL) and peroneus brevis (PB), these three muscles are essential for ankle stability or range of motion [45]. Therefore, these six muscles are ones of the essential parameters for assessing the function of ankle stability. Recently, machine learning and multivariate analysis have been shown to be useful methods for discovering mechanical trends in running performance [40]. For example, PCA and SVM have been used to recognize the gait patterns differences [35], [42], [50], [61]. Thus, by using machine learning and multivariate analysis, there could be differences identified in gait muscle forces generated while wearing the NS and BS. Through this approach, we can understand its internal mechanism more effectively for helping athletes avoiding injuries and improving lower limbs function as much as possible.

It is possible to better understand the differences between the NS and BS by using simple, fundamental running concepts to examine the changes occurring within the biomechanics parameters. To our knowledge, there are no studies that have investigated the differences in ankle joint between the NS and the BS during the running stance phases. The question is that, whether this unstable and bionic condition can improve the stability of ankle joint during the running stance phases. Therefore, the purpose of this study was to investigate the differences in ankle joint parameters of biomechanics changes between the NS and the BS during the running stance phases. We hypothesized that the muscle forces are recognized between BS and NS, and for the BS on PB and PL will be bigger than that of the NS during the running stance phases. We further hypothesized the BS might be increasing ankle stability during the running stance phase.

2. Materials and methods

Participants

A total of 40 Chinese male runners from Ningbo University were recruited for this study (age: 22.3 ± 3.01 years; height: 174.67 ± 7.11 cm; body weight (BW): 66.83 ± 9.91 kg). In the six months leading up to this trial, no surgical injuries had been discovered and none of the subjects had any lower limb medical conditions that may have affected the study's outcomes. A written informed consent form was signed by all participants after they were told of the study's objectives and the circumstances under which it would be conducted. The Ethics Committee of Ningbo University approved this study (protocol code: RAGH 20210608).

Shoes

There were two types of shoes utilized in the experiment, as can be seen in Fig. 1a. A foot-scanning machine was used (VAS-39, Orthobaltic, Lithuania) to scan individual foot shape, then using a 3D print (Dragon(L) 3D Printer, Winbo, China) [19]. Based on the data from the foot scanner, a plastic foot model was developed. This scanned data was then given to the shoe factory (Ningbo Jiangbei Feibu Sports Goods Co., Ltd., Ningbo, China), which developed the shoe tree and then manufactured the shoe. The materials and stiffness of BS and NS were totally the same [22], [23], [62], [65], [67]. Table 1 contains information on the shoes.

Table 1. A comparison between bionic shoes (BS) versus normal shoes (NS)

	BS	NS
Weight [g]	271.0 (2.0)	294.5 (2.3)
Heel height [mm]	23.0 (1.0)	27.0 (1.0)
Bending stiffness [N/mm]	14.2 (0.5)	13.6 (0.4)
Sole hardness [Asker C]	50.0 (0.9)	49.6 (0.6)
Shoe upper material	PVC	PVC
Shoe sole material	EVA	EVA

PVC – nylon (polyamide) polyvinyl chloride; EVA – ethylene-vinyl acetate.

Experimental protocol and equipment

Sports biomechanics research at Ningbo University Research Academy of Grand Health is well-equipped. The biomechanics laboratory was used to conduct all of the testing. The kinematics and dynamics data were gathered using a Vicon motion capture system (Oxford Metrics, Ltd., Oxford, UK) and a force platform (Kistler, Switzerland). Kinematic and dynamic data were recorded at 200 and 1000 Hz, respectively, for the purposes of this study. Surface muscle activations and forces were recorded using EMG equipment (Delsys, Boston, MA, USA) set to a frequency of 1000 Hz, including tibialis anterior, peroneus longus, peroneus brevis, gastrocnemius medial, gastrocnemius lateral, and soleus (Fig. 1d). The data from all of the devices was collected in a synchronized manner. All participants were requested to wear tight shorts and trousers for each test. Each participant was marked with thirty-

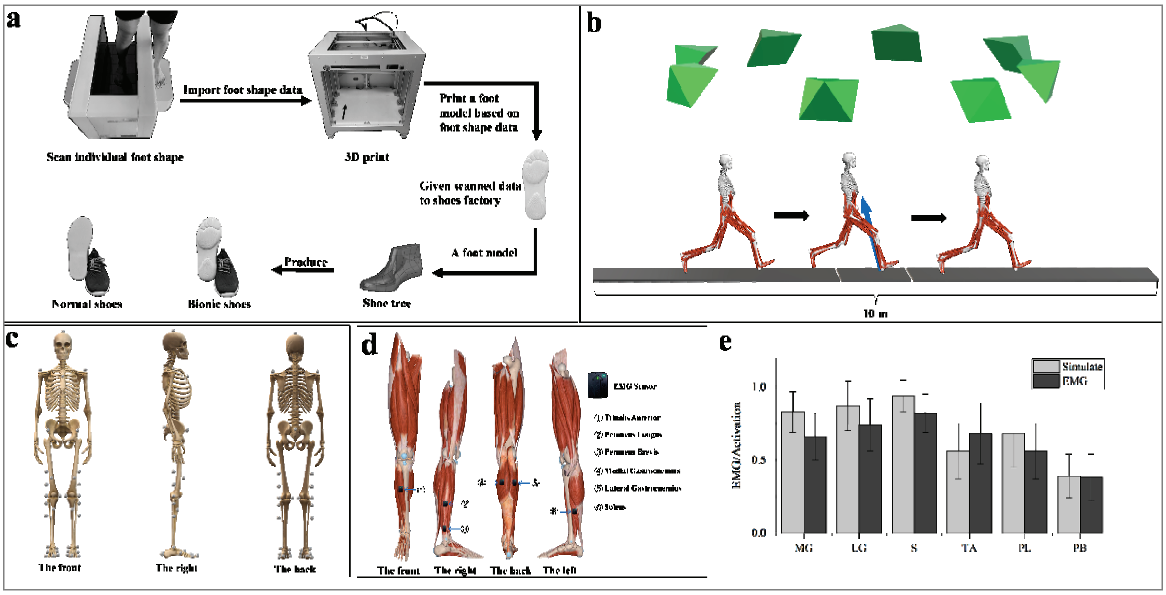

Fig. 1. (a) Procedures for producing shoes; (b) experiment design for gathering kinematics and dynamics data between the NS and BS; (c) skeleton representation showing marker positions on the joints and segments of the lower limb; d) placements of the EMG sensor on six muscles; e) illustration of EMG/activation of each muscle. The scale on the left of the image shows that 0 (no activity) ~1 (full activity). MG – Medial Gastrocnemius; LG – Lateral Gastrocnemius; S – Soleus; TA – Tibialis Anterior; PL – Peroneus Longus; PB – Peroneus Brevis

nine (12.5 mm in diameter) reflective markers. Each marker is shown in Fig. 1c.

Procedure

During the warm-up, participants were asked to accomplish the following: (a) jogging on a treadmill at a pace of 8 km/h for 10 min, then (b) lower limb stretching exercises. Prior to conducting official experiments, all participants were given three trials to familiarize themselves with the test motions. Hair was removed from the testing region and skin abrasion and alcohol washing was used to minimize the impedance of the skin–electrode contact. As soon as all of the participants had a thorough understanding of the methods and experimental settings, they were fitted with markers and sensors to record their maximum voluntary contraction (MVC) of each target muscle similar to [55]. Participants were asked to stand on a force platform to acquire static coordinates after recording the MVC.

Static coordinates were obtained by asking to stand on the *Y*-axis of the force platform having participants cross their arms over their shoulders and keeping their eyes focused ahead until it had been captured. Running exercises were done by all participants at a self--selected pace along a 10-meter pathway for the gathering of dynamic data (Fig. 1b). Only one leg was used for data collection, which was designated as the preferred leg to kick a ball. At the end of each test, there was a one-minute interval between the data collection. With respect to GRF, the first contact was defined as surpassing 10 N [60], [66]. For the sake of data collection, the BS and NS running tasks were split into two separate days and completed at the same time on each day. This was critical in order to eliminate fatigue-related potential errors during the data gathering process.

Data Collection and Processing

C3D files were created using the Vicon Nexus software, which detected the kinematics and GRF data. Low-pass filtering and data extraction for kinematics and GRF data were performed using MATLAB R2019a (The MathWorks in Natick, MA, USA). To sum it up, the following procedures were followed: (1) according to kinematic data, the resulting coordinate system was converted to the coordinate system utilized in OpenSim (Stanford University, Stanford, CA, USA). That is, the forward direction of the human body was the positive direction of the *X*-axis, and the upward direction perpendicular to the ground was the *Y*-axis. The positive direction and the direction of the human body to the right were the positive directions of the *Z*-axis; (2) the biomechanical data for the trajectory of the marker and the GRF were filtered using low-pass Butterworth filters at 6 and 30 Hz; (3) to use the OpenSim simulation software, the kinematics and GRF data were gathered during running stance phases and translated to the trc. (marker trajectory) and mot. (force plate data) formats.

EMG data was utilized to demonstrate the validity of muscle force and activation. The EMG data was first filtered using a 4th-order band-pass filter between 10 and 500 Hz. It was smoothed out by using a low-pass filter with a frequency of 10 Hz [67]. In Figure 1e, the comparison of the EMG data and musculoskeletal models in terms of muscle activation is shown.

During this research, biomechanical parameters were processed and calculated using OpenSim (Stanford University, Stanford, CA, USA). OpenSim musculoskeletal model (gait 2392) with 10 rigid bodies, 23 degrees of freedom, and 92 musculotendon actuators was employed in this study [8], [9]. In order to gather data on muscle force and activation output, the modeling procedures were followed a previous study [67]: 1) use OpenSim 4.2, import the model of a static object. Then use the scale tool to get the anthropometric model for each participant; 2) use the inverse kinematics (IK) and inverse dynamics (ID) tool to calculate kinematics and dynamics data; 3) apply the residual reduction algorithm (RRA) and computed muscle control (CMC) tool with smoothed final test data, and calculate the muscle force and the activations.

Analysis

The dataset was subjected to Shapiro–Wilk normality tests prior to statistical analysis. Paired *t*-tests were utilized to compare running stance phases between two pairs of shoes, and no significant differences were discovered. All data from the walking and running phases were retrieved, and the data of the stance phase was stretched into a time series curve of 101 data points using a custom script of MATLAB for SPM). Open source SPM1d paired-samples *t*-tests software was utilized for the statistical analysis. The significance threshold *p*-value was set at 0.05.

For the muscle force data waveforms, the PCA was used to extract the main features, and then the SVM was used to recognize and classify the features. First, combining the muscle force data into a 400 × 606 matrix, which represents 40 subjects × 5 trials × 2 shoes and 101 data points × 6 muscles. The data matrix underwent a PCA because of the huge number of dependent variables and the possibility for data redundancy. PCA is a multivariate statistical analysis approach that transforms several indexes into a few comprehensive indices by orthogonal rotation transformation with the notion of dimensionality reduction and the assumption of retaining less information. The main purpose of PCA is to produce a collection of non-redundant variables for compactly describing a certain phenomenon or method (data dimension reduction) [7], [59]. In our study, the number of principal components (PC) was determined using the cumulative contribution rate [18]. Finally, the cumulative contribution rate of the early PC reached 90% was selected, then using these several PC scores as the predictor variables [18], [42].

In order to recognize the muscle force differences between the NS and BS, SVM models were built using the predictor variables as inputs. The SVM technique was selected because of its ability to overcome the challenge of high dimensionality with strong discriminative strength for group classification, even in circumstances when the sample size is quite small [36], [50], [58]. The SVM algorithm creates a maximum margin of separation between binary classes in a dataset by defining an ideal separating hyperplane [58]. The SVM uses kernel functions to turn the input feature's data into a higher-dimensional space and then creates a linear hyperplane in this transformed space, which may be projected back to the original data space. SVM employs the soft margin (*c* parameter was set as 1) idea to cope with the possibility of misclassifications (data points on the incorrect side of the separating hyperplane), which does not influence the final result in any way [4], [12].

In order to examine the generalization ability of the classifier in detecting the label of unknown data and to prevent data overfitting, 10-fold cross-validation was done, wherein the data were randomly assigned to 10 equally subsets [12]. An individual dataset was kept for testing purposes while the other nine datasets were utilized as training datasets for the model. Finally, a single classification rate was calculated by averaging all 10 outcomes of cross-validation. At the same time, the standardized effect size, Cohen's *d* was employed to discover significant changes in the waveforms' biomechanical meaning [26]. Cohen's standard definition of a significant difference between groups identified a big effect size as a meaningful difference ($d \geq 0.8$) [5].

3. Results

In Figure 2a, it is shown that there were significant differences detected between the NS and the BS during the running stance phase in ankle angle dorsiflex-

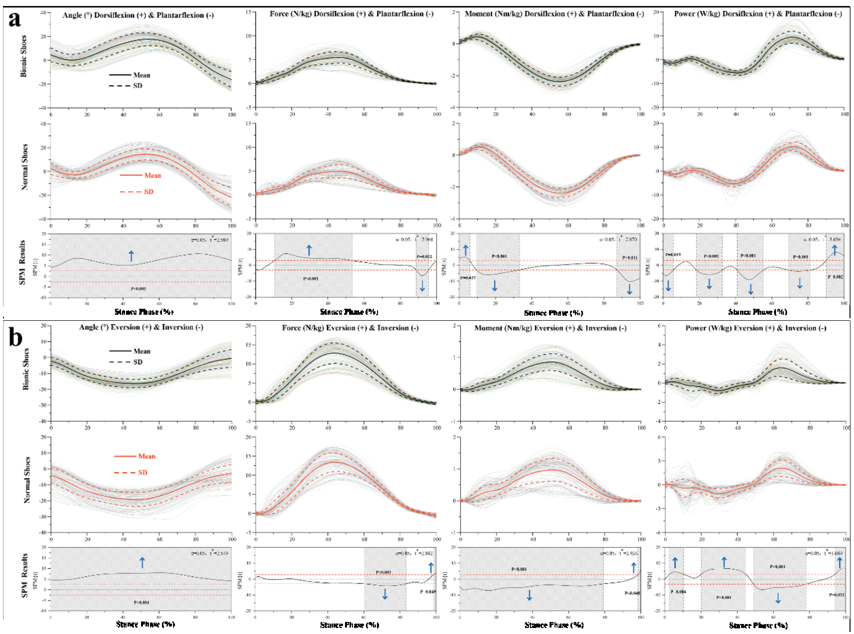

Fig. 2. a) The findings of SPM for the ankle angle, force, moment and power dorsiflexion and plantarflexion measurements during the running stance phase between the NS and BS lower extremities; b) The findings of SPM for the ankle angle, force, moment and power eversion and inversion measurements during the running stance phase between the NS and BS lower extremities. The coloured lines represent the all-test dataset during the BS running stance phase. The grey lines represent the all-test dataset during the NS running stance phase. The SPM findings for all participants (post hoc results; dashed red lines reflect the results at $p = 0.05$) are shown in grey shades. The upward arrow shows that the value of using BS is higher than NS during running stance phase. The downward arrow shows that the value of using BS is lower than NS during the running stance phase. Mean – Mean value, SD – standard deviation

ion and plantarflexion (0~100, $p < 0.001$). There were also significant differences detected between the NS and the BS during the running stance phase in ankle force dorsiflexion and plantarflexion (10.42~53.61, $p < 0.001$) (88.74~95.66, $p = 0.022$) as well as between the NS and the BS during the running stance phase in ankle moment dorsiflexion and plantarflexion (0~5.82, $p = 0.037$) (9.43~33.20, $p < 0.001$) (87.07 ~100, $p = 0.011$) and between the NS and the BS during the running stance phase in ankle power dorsiflexion and plantarflexion (0~5.52, $p = 0.019$), (18.9~32.61, $p < 0.001$), (40.92~55.26, $p < 0.001$), (69.34~81.23, $p = 0.001$), (90.09~100, $p = 0.002$).

In Figure 2b, significant differences detected between the NS and the BS during the running stance phase in ankle angle eversion and inversion (0~100, $p < 0.001$) were shown. There were also significant differences detected between the NS and the BS during the running stance phase in ankle force eversion and inversion (60.21~83.43, $p < 0.001$), (98.56~100, $p = 0.049$) as well as between the NS and the BS during the running stance phase in ankle moment eversion and inversion (0~79.28, $p < 0.001$), (98.34~100, $p = 0.048$) and between the NS and the BS during the running stance phase in ankle power eversion and inversion (2.52~10.38, $p = 0.004$), (20.09~44.77, $p < 0.001$), (48.99~78.02, $p < 0.001$), (94.06~100, $p = 0.012$).

In Figure 3a, significant differences detected between the NS and the BS during the running stance phase in medial and lateral GRF (1.75~7.93, $p = 0.010$), (11.28~22.70, $p < 0.001$), (43.82~49.29, $p = 0.014$), (57.60~89.33, $p < 0.001$) are shown. There were also significant differences detected between the NS and the BS during the running stance phase in anterior and posterior GRF (0~5.99, $p = 0.018$), (8.71~17.42, $p = 0.006$), (23.59~60.50, $p < 0.001$),

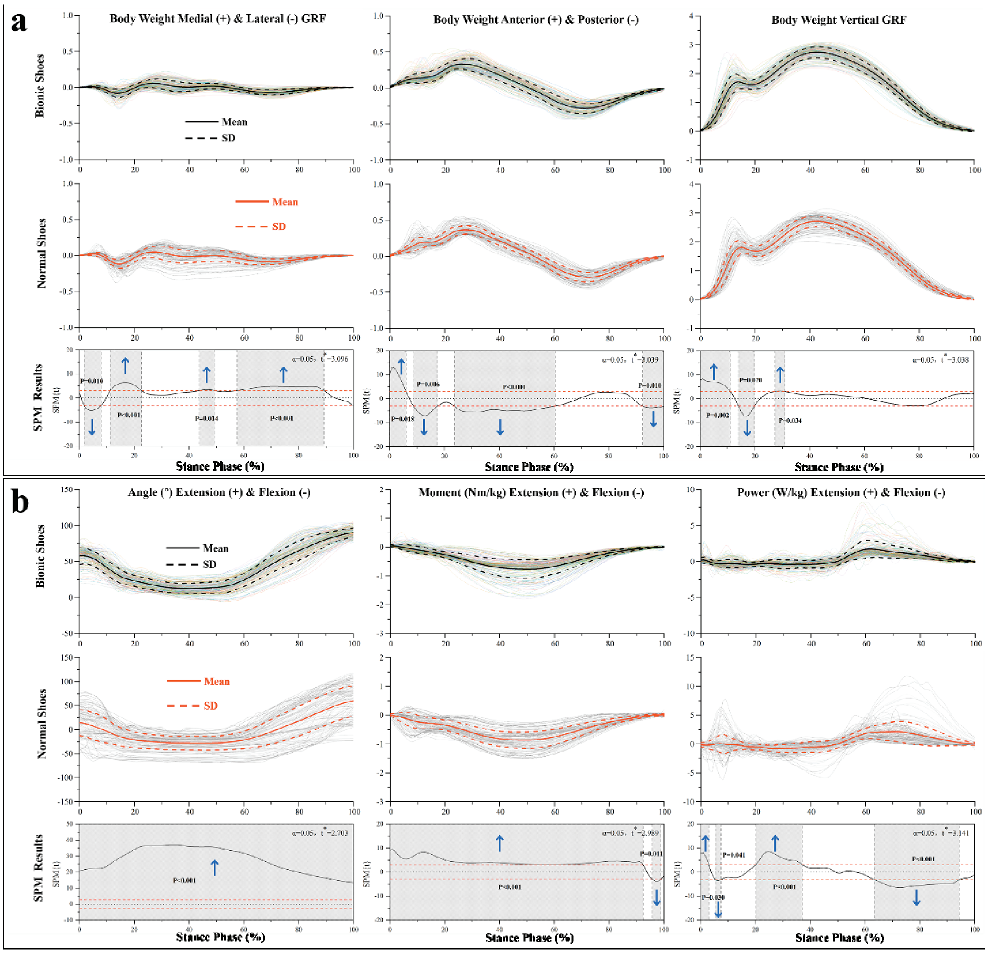

Fig. 3. a) The findings of SPM for the GRF on three anatomy planes during the running stance phase between the NS and BS lower extremities; b) the findings of SPM for the MPJ1 angle, moment and power extension and flexion measurements during the running stance phase between the NS and BS lower extremities. The coloured lines represent the all-test dataset during the BS running stance phase. The grey lines represent the all-test dataset during the NS running stance phase. The SPM findings for all participants (post hoc results; dashed red lines reflect the results at $p = 0.05$) are shown in grey shades. The upward arrow shows that the value of using BS is higher than NS during the running stance phase. The downward arrow shows that the value of using BS is lower than NS during the running stance phase. Mean – Mean value, SD – standard deviation

(92.34~100, $p = 0.010$) as well as between the NS and the BS during the running stance phase in vertical GRF (0~11.04, $p = 0.002$), (14.00~19.80, $p = 0.020$), (27.29~30.98, $p = 0.034$).

In Figure 3b, significant differences detected between the NS and the BS during the running stance phase in MPJ1 angle extension and flexion (0~100, $p < 0.001$) are shown. There were also significant differences detected between the NS and the BS during the running stance phase in MPJ1 moment extension and flexion (0~92.46, $p < 0.001$), (95.83~98.99, $p = 0.041$) as well as between the NS and the BS during the running stance phase in MPJ1 power extension and flexion (0~3.00, $p = 0.030$), (5.48~7.37, $p = 0.041$), (20.17~37.09, $p < 0.001$), (63.38~94.44, $p < 0.001$).

In Figure 4, significant differences detected between the NS and the BS during the running stance phase in medial gastrocnemius (0~43.51, $p < 0.001$), (53.17~89.94, $p < 0.001$) are shown. There were also significant differences detected between the NS and the BS during the running stance phase in lateral gastrocnemius (0~23.19, $p < 0.001$), (29.13~75.44, $p < 0.001$), (81.45~100, $p = 0.001$) as well as between the NS and the BS during the running stance phase in soleus (0~10.73, $p = 0.024$), (51.42~76.6, $p = 0.001$) between the NS and the BS during the running stance phase in tibialis anterior (0~7.51, $p = 0.038$), (12.53~100, $p < 0.001$) and between the NS and the BS during the running stance phase in peroneus brevis (0~35.23, $p < 0.001$), (37.61~61.93, $p < 0.001$), (63.71~90.80, $p < 0.001$), (97.52~100, $p = 0.046$) as

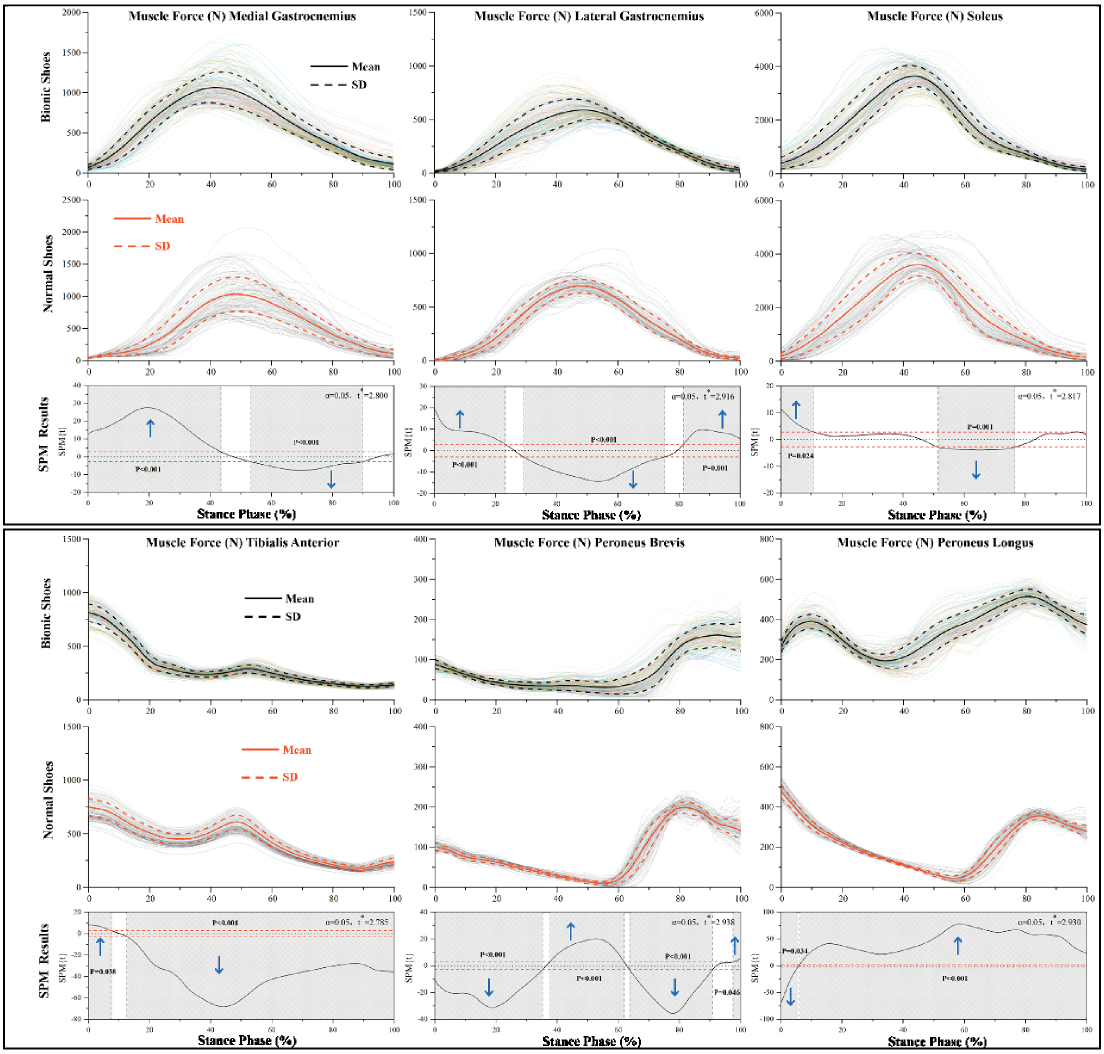

Fig. 4. The findings of SPM for the MG, LF, S, TA, PB and PL muscle force measurements during the running stance phase between the NS and BS lower extremities. The coloured lines represent the all-test dataset during the BS running stance phase. The grey lines represent the all-test dataset during the NS running stance phase. The SPM findings for all participants (post hoc results; dashed red lines reflect the results at $p = 0.05$) are shown in grey shades. The upward arrow shows that the value of using BS is higher than NS during the running stance phase. The downward arrow shows that the value of using BS is lower than NS during the running stance phase. Mean – Mean value, SD – standard deviation

well as between the NS and the BS during the running stance phase in peroneus longus (0~5.44, p = 0.034), (6.17~100, p < 0.001).

Additionally, a total of 18 PC scores were obtained as the predictor variables of the SVM model. The three-dimensional scatter diagram of the first three PC scores of muscle force of each muscle is shown in Fig. 5, and there is no discernible linear relationship between them. The accuracy of SVM model in identifying the gait muscle force between BS and NS reached 100%, which proved that the BS had a very large impact on the gait muscle force compared with NS. Among them, the muscle that contributes most to recognizing the BS and NS is mainly the peroneus longus muscle, which is related to PC1 (represents 44.90% of variance explained of the peroneus longus muscle, d = 7.57). The second major contributor muscle is the peroneus brevis, which is also related to PC1 (represents 82.31% of variance explained of the peroneus longus muscle, d = 3.89). The muscle that does not contribute to recognizing the BS and NS is soleus.

4. Discussion

The purpose of this study was to investigate the differences in ankle joint when using the NS and the BS during the running stance phases. We hypothesized that the muscle force of the BS on PB and PL will be bigger than that of the NS during the running stance phases. We further hypothesized that using the BS might increase ankle stability during the running stance phase. The results of our study are partly consistent with our hypothesis.

According to prior research, MBT shoes have been shown to have a larger flexion angle of the knee and an increased range of ankle mobility during the gait stance phase [11], [52]. The ankle's range of motion may be reduced by wearing shoes that are unstable from another viewpoint [29]. This is completely inconsistent with our results. Our results show that using BS does change the kinematics on ankle joint angle during the running stance phase, but no statistically significant differences were found in joint changes of range of motion. The reason for this is probably because as the unstable shoe approaches the plantarflexion and dorsiflexion of the ankle, it alters the lower extremity kinematics and dynamics. This forced increase in range of motion, in some ways, leads to a loss of proprioception, more like a mechanized motion, but we do not know if it is good for humans. In contrast, our result indicated that running with BS has higher ankle angle dorsiflexion than NS, with no forced changes in joint range of motions. We propose that BS is the real unstable condition shoes that human needs.

Interestingly, although the ankle angle dorsiflexion increased in BS running, both the joint moment and force were lower than that in NS running in most stages

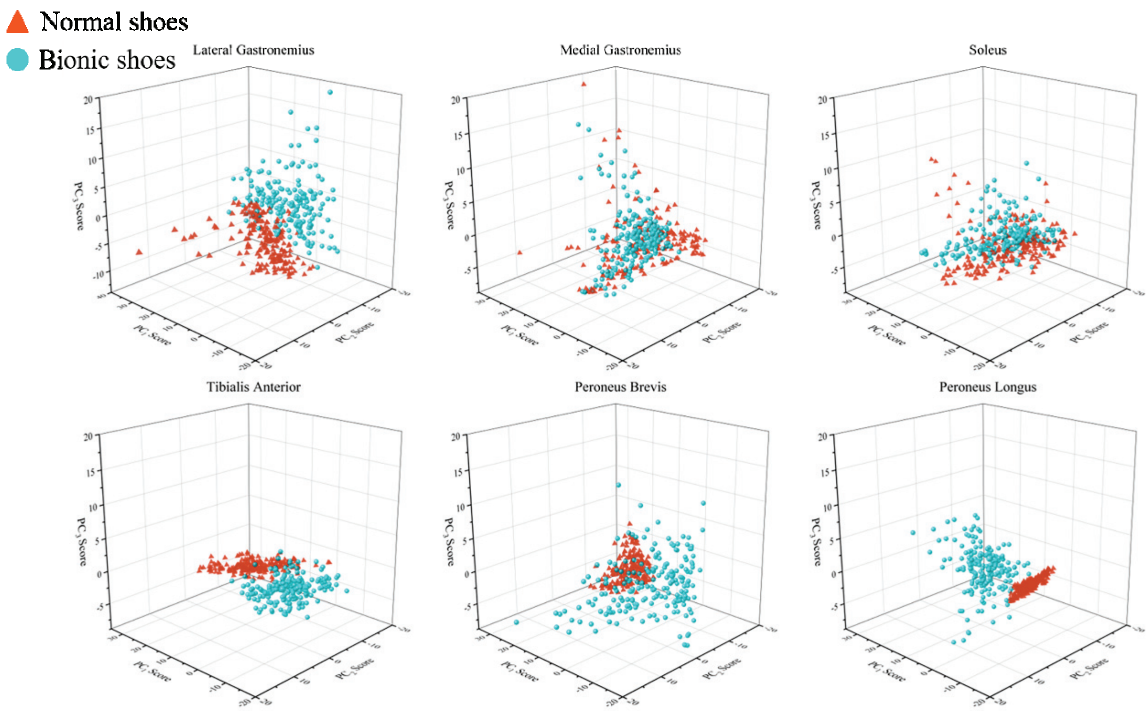

Fig. 5. The three-dimensional scatter diagram of the first three PC scores of muscle force of each muscle (X-axis – PC_1 Score; Y-axis – PC_2 Score; Z-axis – PC_3 Score)

of the stance phase, and there was a significant difference. To our knowledge, there have been no scientific investigations on the relationship between ankle kinematics and dynamics to running in BS, which makes it impossible to speculate about the particular causes. Previous research on unstable shoes suggests that there are many plausible explanations for this. By summarizing the results of muscle function, Rachel [30] has indicated that one of the functions of MG, LG and S assisted the ankle joint during dorsiflexion and plantarflexion. From this point of view, this seems to be a valid explanation for our results. When running in BS, the ankle joint is not an independent part, which may be caused by the dynamic chain. If some parameters change in the same motion or in the same object, it means that another part of the parameters will change with it, and causality is well presented here. Therefore, a large part of the reason for this result may be due to the change of MG, LG and S compensation. On the other hand, according to the analysis of SVM model, we can prove that the BS had a very large impact on the gait muscle force compared to NS. Among them, the muscle that contributes most to recognizing the BS and NS is mainly the peroneus longus muscle, and the second major contributor muscle is the peroneus brevis, which further evidences that BS may be more helpful to users than NS.

The instability conditions we mentioned earlier are medial and lateral, not dorsiflexion and plantarflexion. In this sense, the frontal plane may be a more important indicator parameter when evaluating the stability or function of BS. Our results show that running in BS have higher ankle angle eversion than in NS, and with lower force and moment. According to the previous study [41], this modification can effectively reduce ankle varus sprains, which fundamentally solves one of our major problems. BS reduces ankle pronation by restoring the original ontological state, activating its own balance ability and discarding the inherent functional support shoe. Muscle force is one of the most important biomechanics parameters which was used to reflect the function or ability of lower limbs [51], [63]. Our results of muscle force show that running in BS has a significantly higher difference in PL and PB muscle force than in NS. Compared to a previous studies [31], [45], [47], we conclude that increased PL and PB muscle force significantly reduce the risk of ankle angle inverse injury. It is further proved that BS plays an important role in the stability of ankle joint including the reduction of injury probability.

Podiatric and orthopedic doctors often face hallux valgus, which is one of the most prevalent foot abnormalities [53]. MPJ1 joint subluxation, as well as valgus angulation, and proximal phalanx pronation, are all symptoms of hallux valgus, which is also often called "bunion deformity" [25], [37], [38]. A previous study indicated that lower MPJ1 flexion can effectively reduce the probability of bunion deformity [10], [14], which is consistent with our results. The reason for this result, combined with the results of joint moment and power obtained by us, is likely to be that BS increases the stiffness of MPJ1 under such unstable conditions, thus reducing the degree of buckling of MPJ1, and thus reducing the risk of bunion deformity. Compared to the analysis of medial and lateral GRF, we found that medial GRF was significantly higher than that of NS when BS was used for running, which further confirmed our idea. Furthermore, we found that BS running has a significantly lower vertical GRF than NS during the running stance phase at the first peak value. This strengthens the evidence that using BS can reduce damage.

We have to admit that there are some limitations in the present study. First, we only had male volunteers in our study. According to prior research, women's pelvic anatomy makes them more susceptible to accidents to the lower extremities. Second, we did not explore the differences between two different shoes during walking and running by using other tools like finite element analysis. Third, we did not detect the change of the foot arch changes between two different shoes during the walking and running phases, by using multiple foot models such as the oxford foot model. Last, different age and different countries of subjects may also cause the final result to be different. Those limitations could allow us to look at it from a different sight, a more precise sight.

5. Conclusions

This study investigated the differences in ankle joint between using the NS and the BS during the running stance phases. We found that BS may be better suited to the human condition than other unstable shoes, or even NS. In addition, our results suggest that BS plays an important role in reducing ankle injuries during running by increasing muscle participation in unstable conditions while better restoring the most primitive instability of foot condition that humans have.

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