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Effects of unstable elements with different hardness on lower limb loading

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Purpose: Osteoarthritis of the knee is one of the most common diseases. For this chronic disease, modified footwear structure can effectively prevent and relieve disease of the knee. The aim of this study was to explore the effects of shoe surface elastic modulus on external knee adduction moment and ground reaction force and foot loading characteristics. Methods: Sixteen healthy female volunteers were recruited, and each subject performed five walking trials under two shoes condition. The lower limb loading data was collected using force platform and in-sole pressure measurement system. Results: The results showed that the external knee adduction moment was decreased in all stance phase when wearing SS (unstable shoes with soft unstable elements), compared with HS (unstable shoes with hard unstable elements). The ground reaction force showed no obvious change under two shoes condition. Additionally, compared with HS, plantar pressure transferred from medial foot to lateral foot when wearing SS. Along with changes of contact areas, average pressure and impulse had also presented this tendency. Conclusions: These results can provide some scientific evidence and suggestions for footwear companies, and for the foot plantar medial injury disease has also certain applicability.

Key words: *knee adduction moment, plantar pressure, unstable elements, hardness, foot*

1. Introduction

Footwear is widely used as a tool to protect the foot and improve walking capability. In recent years unstable shoes have become increasingly popular as both a therapeutic and a functional tool [10], [16]. The so-called “unstable” shoe designs, which are characterized by a round sole in the anterior-posterior direction with a soft pad underneath the rear foot, are claimed by footwear companies and proponents to increase muscle activity during standing [17] and walking [23] and relieve leg, back and foot problems. These benefits are advocated for the effect of the increased activation level of additional muscles [6] and the reduction of joint reaction forces [16].

Originating from Switzerland in 1996, the unstable Masai Barefoot Technology (MBT) shoe can be considered as the original “barefoot” functional shoe. Among those functional shoes, the MBT has been the most extensively studied in the peer-reviewed literature [16], [23], [11], [19], [22].

Previous unstable shoes research has predominantly focused on Masai Barefoot Technology (MBT). Nigg et al. [16] reported that MBT unstable shoes strengthen muscles which are anatomically closer to the axes of joint movement, therefore reducing joint loading. Research on MBT has identified an increased range [11] and velocity [17] of COP motion while standing, increased tibialis anterior activation during swing, increased gastrocnemius activation during early-stance and mid-stance [6] and improvements in reactive balance after the MBT intervention [19]. Other unstable footwear brands, such as Reebok, Skechers and FitFlop, utilize different technologies, which lack the peer-reviewed published literature compared with MBT. Some independent comparative research has been undertaken, which does not support increase in muscle activation in MBT, Skechers Shape-Ups and Reebok Easy Tone when compared to a trainer [4],

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however the protocols utilized in these studies are less comprehensive including less stringent analysis of EMG than the MBT research. Unpublished studies with small participant number commissioned by the footwear companies identify increases in lower limb muscle activation by prolonging and increasing the magnitude of muscle activation. The results from these studies are utilized in marketing material, however, the full protocols and analysis procedures of these studies are not currently in the public domain. Research commissioned by footwear companies acts as a starting point, with peer-reviewed published data being a must for this footwear category.

Andriacchi and Mündermann [1] reported that the rate of progression of osteoarthritis (OA) at the knee is associated with increased loads in the joint during ambulation. Footwear modification for patients with knee OA has received extensive attention as an effective conservative intervention that can alter knee loading [18]. A series of related previous studies have shown that variable-stiffness shoes [5], mobility shoes [20], flat walking shoes [5], flexible non-heeled shoes [25], and shoes with lateral wedging and a variable-stiffness sole [2] can reduce knee joint loading while walking as compared to modern heeled shoes or stability shoes.

There were many studies about unstable shoes on effect of human biomechanics. Most of the research focused on the plantar pressure distribution and gait characteristics, but less on effect of unstable shoes with different hardness elements. Therefore, we reformed the common shoes with two hemispheric unstable elements stuck to the outsole in the middle forefoot and rearfoot into the experimental unstable shoes. The purpose of this study was to explore the effects of changing shoe outsole surface elastic modulus on external knee adduction moment and ground reaction and the foot loading characteristics.

2. Materials and methods

2.1. Subject selection

Sixteen healthy female subjects (age = 22 ± 3 years, height = 1.61 ± 0.05 m, weight = 57.6 ± 5.2 kg; mean ± SD) took part in the experiment. Participants were the undergraduate students in Faculty of Physical Education from Ningbo University. They took sports courses or exercise every other day, three to four days a week. The study excluded subjects who presented one or more of the following aspects: (1) history of surgeries to lower extremity; (2) background or signs of neurological dysfunction which could affect lower limb motor performance, or balance; (3) recent osteo-articular or musculotendinous injury to the lower limb; (4) the previous experience wearing any unstable shoes. The dominant feet of all the subjects are the right one, so all statistics concerning the tests are collected from the right feet. Their feet size are 37 (EU).

2.2. Experiment materials

2.2.1. Elastic modulus test

The materials of unstable elements were rubber and elastic modulus, (or Young's modulus, *E*), was measured and calculated by the elastic modulus test system (INSTRON AG Grove USA) in a cuboid shape (Fig. 1). Unstable elements were hemisphere and induced medial-lateral direction instability during single-support phase.

$$E = \sigma/\varepsilon.$$

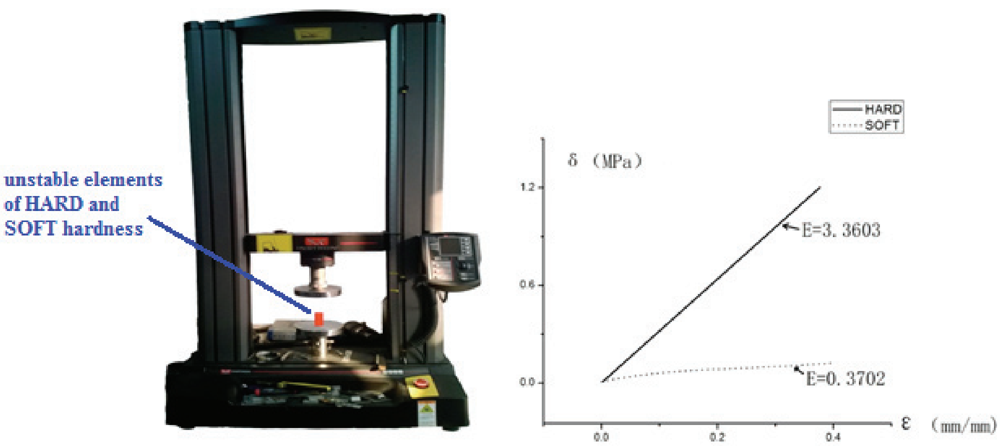

Fig. 1. The elastic modulus test system (left) and elastic modulus (Young's modulus, *E*) of both materials (HARD and SOFT) used for the experimental shoes (right); *E* is also the gradient of the two lines (hard or soft materials), respectively

E – indicates the elastic modulus or Young's modulus and is the ratio of stress with reference to strain (unit: MPa); σ – indicates the stress that is the quotient of an evenly distributed perpendicular force and the area of the surface (unit: MPa); ε – indicates strain that is the ratio of change in length versus the initial length (mm/mm).

2.2.2. Shoes condition

A comprehensive three-dimensional gait analysis was performed on all subjects wearing unstable shoes with elements of hard and soft hardness. In terms of the experimental shoes, the unstable elements of soft and hard stiffness (Fig. 1) are stuck to the outsole of normal shoes in the forefoot and rear-foot zone, with the form of the hemisphere's height of 1.5 cm and diameter of 5.5 cm (Fig. 2) [6], [7]. The unstable elements used, designated based on their position as neutral unstable (NU, mainly the middle metatarsals support) were attached to forefoot following the procedure used in previous studies [6]. The experimental shoes are of EU size of 37.

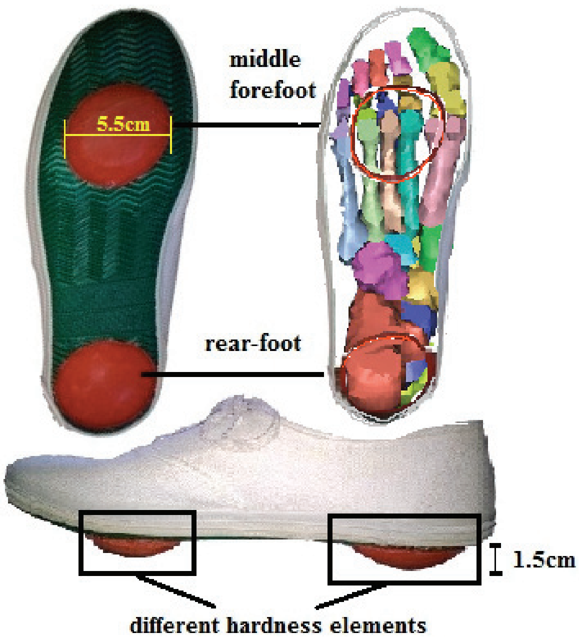

Fig. 2. Unstable shoe and position of unstable element

2.3. Experimental protocol

Participants were required to attend testing session in the Sports Biomechanics Laboratory of Ningbo University. Before the experiment, all subjects were required to stand or walk with unstable shoes at least 1 h per day for two weeks in order to adapt to unstable shoes, so that they can perform their normal gait characteristics in the biomechanical experiments. Each subject performed five walking trials in the soft unstable shoe and hard unstable shoe, with randomized order of the two footwear for all subjects. Infrared timing gates were used to guide the subjects to walk at 1.5 ± 0.5 m/s. And in order to reduce experimental error, our experiment parts were done in the same day.

2.3.1. Joint adduction moment

A three-dimensional motion analysis system (Vicon MX, Vicon Motion System Ltd., Oxford, UK) was used with 8 cameras (8 MX3 and 2 MX40) to capture and analyze motion of the knee with a sampling frequency of 200 Hz. The following measurements were taken for the calculation of the joint centers: height, weight, leg length, and width of knee and ankle. To assess the three-dimensional motion of the lower limb, retro-reflective markers were attached in accordance with Plug-In Gait (PIG). The markers were attached with a division to two groups. A group of makers attached to pelvis (four): left and right anterior superior spine, left and right posterior superior iliac spine. Another group of markers attached to lower limb (twelve): left and right lateral mid-thigh, left and right lateral knee, left and right lateral mid-shank, left and right lateral malleolus, left and right second metatarsal head, left and right heel. In order to distinguish between left and right sides, the marker to right lateral mid-thigh was slightly lower than the left one. The markers were attached to the skin using double-sided tape.

Kinetic data were collected at 1000 Hz with a Kistler force platform (Kistler Instrumente AG, Winterthur, Switzerland). The collected data were exported into Vicon Nexus 1.8.1 (Oxford Metrics Limited, UK), with the kinematic and kinetic data being filtered using a low-pass Butterworth filter with cut-off frequencies of 12 and 50 Hz, respectively [16]. The internal joint moments were calculated with Inverse Dynamics method using the anthropometric, ground reaction force and motion data.

2.3.2. The plantar pressure

The plantar pressure measurements were taken with the Pedar-X system (Novel Electronics, Germany). The system consists of pressure sensing insoles connected to a box which attaches around the subjects waist and transmits information to the Pedar-X software via Bluetooth wireless communication. The 2.5 mm thickness insoles contain 99 capacitive pressure sensors that produce a grid representing pressure distribution on a laptop screen. The sensors are sampled at a rate of 50 Hz. The data of plantar pressure from right foot heel landing to this toe off-ground were adopted for analysis. Using the PEDAR software, the area of the insoles was divided up into eight sections by creating "masks" that grouped sensors into anatomical areas: Big toe, 2–5 toes, Medial metatarsal, Mid-metatarsal, Lateral metatarsal, Midfoot, Medial rearfoot and Lateral rearfoot (Fig. 3). For each mask, average pressure, maximal force and force time integral over all the frames, were computed. The total contact area was also recorded.

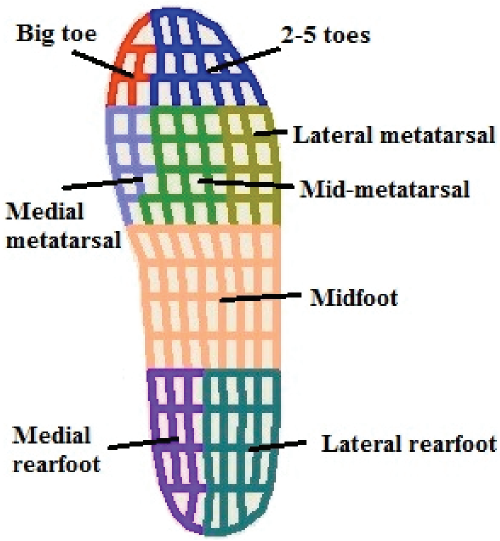

Fig. 3. Eight regions of different foot anatomy

2.4. Statistical analysis

Collected and processed data were analyzed using SPSS version 17.0 software (SPSS Inc., Chicago, IL, USA). Paired-samples T Test was employed to study the effects of different experimental shoes. The significance level of $p = 0.01$ was set in order for the results of the statistical analysis to be considered statistically significant.

3. Results

The gait cycle was divided from the right foot heel landing to the next landing. Analyzed data was the right feet of all subjects.

3.1. Joint adduction moment

During the stance phase of the gait cycle, joint adduction moment curve of the right knee was shown in Fig. 4 under the two kinds of unstable shoes conditions (unstable shoe with soft element, SS; and unstable shoe with hard element, HS).

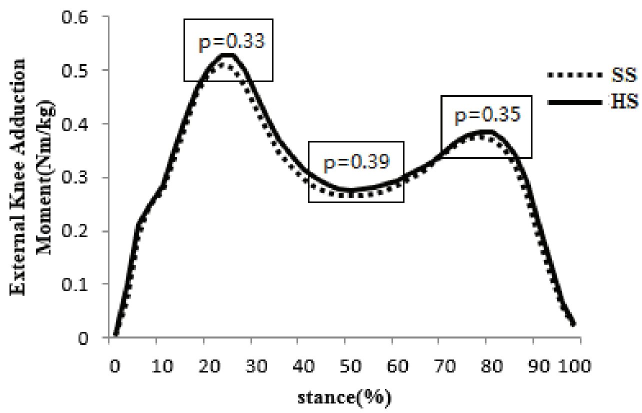

Fig. 4. Mean knee joint moment curve of the stance phase while walking in unstable shoes with different hardness elements (Note: SS indicates the unstable shoes with soft element and HS indicates the unstable shoes with hard element)

From Fig. 4, we can find that the first and the second peak of knee adduction torque under SS condition was lower than in the case of wearing HS. Additionally, through changes of the curve, we can see that knee adduction angle was also reduced when wearing SS.

The comparison of the soft unstable shoe and hard unstable shoe conditions indicated that the first peak external knee adduction moment of soft unstable shoe was decreased (Fig. 4 and Table 1). During the mid-stance, external knee adduction moment with soft

Table 1. The analysis of biomechanical variables under the two shoes conditions and *p* value

Variable	SS	HS	*p* value
First peak knee adduction moment (Nm/kg)	0.51 ± 0.11	0.53 ± 0.12	0.33
Second peak knee adduction moment (Nm/kg)	0.38 ± 0.04	0.39 ± 0.02	0.39
Trough knee adduction moment (Nm/kg)	0.26 ± 0.04	0.27 ± 0.04	0.35

Note: SS represents the unstable shoes with soft elements; HS represents the unstable shoes with hard elements; mean ± SD (Standard Deviation).

unstable shoes showed slight decrease and the second external knee adduction moment also showed slight decrease (Fig. 5 and Table 1).

3.2. The plantar pressure distribution

The data of right foot collected from heel landing to toe off-ground was analyzed. Compared with HS, average pressure of big toe, 2–5 toes, lateral metatarsal, lateral rearfoot regions were significantly increased under SS condition. While average pressure of medial metatarsal, mid-metatarsal, medial rearfoot regions were significantly decreased. And there was no significant change in midfoot regions (Table 2).

Table 2. Average pressure distribution (kPa)

Areas	SS	HS	*p* value
Big toe	100.73 ± 12.55	94.39 ± 22.93	0.000**
2–5 toes	45.56 ± 13.33	39.25 ± 14.62	0.001**
Medial metatarsal	67.76 ± 13.66	76.68 ± 11.98	0.002**
Mid-metatarsal	169.07 ± 41.67	193.30 ± 52.71	0.000**
Lateral metatarsal	60.61 ± 17.23	47.16 ± 16.22	0.000**
Midfoot	0.14 ± 0.30	0.00 ± 0.00	0.168
Medial rearfoot	69.68 ± 26.83	94.08 ± 28.96	0.000**
Lateral rearfoot	126.20 ± 18.86	106.19 ± 28.29	0.000**

Note: ** indicates that there was a significant difference between the two groups ($p < 0.01$).

Table 3. Force distribution (N/kg)

Areas	SS	HS	*p* value
Big toe	1.13 ± 0.10	1.11 ± 0.28	0.626
2–5 toes	27.52 ± 44.22	0.84 ± 0.71	0.040*
Medial metatarsal	1.29 ± 0.25	1.47 ± 0.20	0.002**
Mid-metatarsal	4.87 ± 1.23	5.53 ± 1.36	0.000**
Lateral metatarsal	1.16 ± 0.33	0.90 ± 0.29	0.000**
Midfoot	0.01 ± 0.02	0.00 ± 0.00	0.168
Medial rearfoot	1.83 ± 0.65	1.76 ± 1.62	0.840
Lateral rearfoot	4.55 ± 0.57	2.70 ± 2.33	0.003**

Note: ** indicates that there was a significant difference between the two groups ($p < 0.01$).

With the change of materials' stiffness, plantar pressure and contact area also changed. From Table 3, we could find that there was a series of changes. Compared with HS, maximal force of 2–5 toes, lateral metatarsal and lateral rearfoot regions were significantly increased under SS condition. The maximal force of medial metatarsal, mid-metatarsal regions were significantly decreased, but there was no difference in midfoot under SS condition. Contact area of big toe was significantly increased under SS condition. There were also a significant increase of contact area in lateral metatarsal and lateral rearfoot regions (Table 4).

Table 4. Contact area (cm^2)

Areas	SS	HS	*p* value
Big toe	6.23 ± 0.01	4.23 ± 0.02	0.000**
2–5 toes	12.58 ± 1.52	12.05 ± 1.94	0.056
Medial metatarsal	8.57 ± 0.62	8.44 ± 0.67	0.343
Mid-metatarsal	12.95 ± 1.54	13.59 ± 1.94	0.015
Lateral metatarsal	8.70 ± 1.89	7.29 ± 1.71	0.001**
Midfoot	0.25 ± 0.54	0.00 ± 0.00	0.168
Medial rearfoot	10.91 ± 1.45	10.37 ± 0.46	0.138
Lateral rearfoot	15.54 ± 1.77	13.05 ± 1.82	0.000**

Note: ** indicates that there was a significant difference between the two groups ($p < 0.01$).

Table 5. Impulse distribution (N*s)

Areas	SS	HS	*p* value
Big toe	10.53 ± 1.07	9.41 ± 2.56	0.018*
2–5 toes	7.96 ± 2.24	6.80 ± 2.70	0.001**
Medial metatarsal	12.02 ± 2.12	16.95 ± 2.84	0.000**
Mid-metatarsal	52.24 ± 14.85	69.09 ± 15.52	0.000**
Lateral metatarsal	10.65 ± 2.97	9.60 ± 2.04	0.027*
Midfoot	0.05 ± 0.11	0.00 ± 0.00	0.234
Medial rearfoot	16.71 ± 11.44	22.19 ± 10.20	0.003**
Lateral rearfoot	41.63 ± 15.70	35.63 ± 15.82	0.001**

Note: ** indicates that there was a significantly difference between the two groups ($p < 0.01$).

Impulse of different areas also appeared to be significantly changed. Compared with HS, impulse of big toe, 2–5 toes, lateral metatarsal, lateral rearfoot areas significantly increased, while impulse of medial metatarsal, mid-metatarsal and medial rearfoot areas obviously decreased under SS condition (Table 5).

4. Discussion

4.1. Joint adduction moment

One aim of this study was to investigate the differences of knee joint moments and ground reaction force with unstable shoes of different hardness outsole. Various design factors of footwear such as heel width, arch support, sole shape, flexibility or cushioning ability might influence joint loading [8].

Changes of knee adduction torque and the ground reaction force under the conditions of different hardness soles were analyzed in this study. From Fig. 4, we could find that the first and the second peak of knee adduction torque under SS condition was lower than in the case of wearing HS. Additionally, through changes of the curve, we could know that knee adduction angle was also reduced when wearing SS. Landry et al. [12] have shown that the decrease of the knee adduction torque mainly appears in the early stance when wearing MBT walking shoes. The experimental shoes used in this study were similar to MBT shoes, especially the unstable shoes with soft hardness element (SS). Results of this study showed that knee adduction torque was reduced during heel-strike gait period under SS condition compared with HS. This may be due to the characteristic of soft hemispheric structure under the rearfoot part of unstable shoes. The soft element to the unstable shoes under stress would produce more compression deformation, reducing the range of knee joint activity in coronal plane during heel-strike gait period, and thus shorten the force arm of knee adduction. In addition, soft material was similar to the material of MBT, which would be beneficial to the cushioning function of ground reaction force [12]. The MBT shoes have mainly produced instability in the anterior and posterior direction [24]. The unstable shoes in this study are similar to the Easy Tone of Reebok [4], which mainly produced instability in the medial and lateral direction. Therefore, knee angle in coronal plane increased in mid-stance period, and the knee adduction torque was also relatively increased when wearing HS [3]. Moreover, knee adduction torque also appeared to be slightly reduced under SS condition during toe-off period when compared with HS. The reasons may be inconsistent with the soft unstable elements under heel position, but the amount of reduction was less obvious compared with heel-strike period. Although the rearfoot region and forefoot region add unstable elements of the same material property, the impact on the knee of unstable element in the forefoot is smaller than that of the rearfoot part. Also, the unstable elements to the forefoot and rearfoot part could stimulate static and dynamic stability, thus enhancing postural control and preventing falling risk [6]. A similar unstable shoe with height-adjustable elements in forefoot and rearfoot part, called Apos-System or Apos-Therapy shoes, was proven to effectively improve the symptoms of knee osteoarthritis (OA) [7]. The future research of this kind of unstable shoes shall focus on the combined effect of both the stiffness and the height of the unstable elements on the impact of knee.

In this study, there was no significant change of ground reaction force under the condition of two experimental shoes during the gait cycle. This result was different from some previous studies. Isabel et al. [9] reported that ground reaction force was slightly increased when wearing MBT shoes compared with barefoot and ordinary shoes. The high ground reaction force could raise the risk of damage [15], while wearing MBT shoes would accelerate the pace of walking, make momentum of distal joint increase and contribute to expansion of ground reaction force [15]. The experimental shoes mainly produced effect of instability in medial and lateral direction. And there was same effect in a study of Easy Tone; it was reported that the ground reaction force seemed to have a rising trend during early stance period, but did not give a certain answer [24]. This may suggest that unstable shoes with different structure would not significantly affect the ground reaction force.

4.2. Plantar pressure distribution

This study showed that plantar pressure patterns were significantly different when wearing unstable shoes of different outsole hardness. The results were consistent with previous studies. Some studies reported that the design of the arc sole can effectively reduce plantar pressure in forefoot regions [15]. This study adopt materials of different hardness on the basis of the arc sole (elastic modulus E = 0.3702 of sole material for SS, elastic modulus E = 3.3603 of sole material for HS). Compared with HS, average pressure of big toe, 2–5 toes, lateral metatarsal, lateral rearfoot regions were significantly increased when wearing SS, while average pressure of medial metatarsal, mid-metatarsal, medial rearfoot areas were significantly decreased and average pressure of midfoot area does not appear to be significantly changed. Changes of pressure are determined by the plantar pressure and contact area, which are positively related with plantar pressure and negatively correlated with contact area. In terms of pressure, compared with HS, pressure of big toes, 2–5 toes, lateral metatarsal and lateral rearfoot regions were significantly increased when wearing SS, while pressure of medial metatarsal, mid-metatarsal regions were significantly decreased. This suggested that SS dispersed pressure on medial and mid-metatarsal and medial rearfoot areas or transferred pressure from medial and mid-metatarsal and medial rearfoot regions to toes and lateral rearfoot areas, so plantar pressure of medial and mid-metatarsal and medial rearfoot regions were correspondingly reduced. This may be that the relatively soft materials of SS under stress will produce more compression deformation. The shape change variable is larger when under larger stress, if this area is too large, the area is prone to compensatory deformation, so as to make the average plantar pressure distribution more evenly. Additionally, proprioception system of foot will make human body adjust to a stable state by transferred pressure to seek a stable support [6]. At the same time, the size of the plantar pressure is determined by the contact area. In this study, contact area of midfoot was significantly increased, while lateral metatarsal and lateral rearfoot regions were significantly decreased with the unstable support base, so the increasing pressure of lateral foot may be due to the reduction of contact area in lateral metatarsal and lateral rearfoot regions. The increase of contact area also explained the cause of pressure reduction of the mid-metatarsal. In addition, midfoot region formed the hollow form due to the outsole support of unstable elements. Therefore, average pressure of midfoot region was not significantly changed under two kinds of unstable shoes with different hardness.

Impulse is defined as force and time integral, and the force produced an accumulative effect during a certain period of time for plantar regions. Therefore, the size of the impulse pressure was in connection with pressure and contact time in each anatomical part. Compared with HS, impulse of big toe, 2–5 toes, lateral metatarsal, lateral rearfoot regions were significantly increased when wearing SS, while medial metatarsal, mid-metatarsal and medial rearfoot were significantly decreased in this study. The reason of impulse increase in toes and lateral foot regions may be due to the rise of force in this region (Table 3) due to the contact time or stance period being fixed. During the gait of wearing SS, these regions may bear more pressure, but the soft element to the outsole of unstable shoes could function as a spring or cushion to relieve force transmission [14] in walking process and protect foot, then avoiding injuries.

There are still several limitations in this study. Firstly, this testing session was completed in the laboratory. There are some differences between daily life environment and laboratory. Secondly, the use of young healthy participants also is a limitation because it limits the application of the findings in this study to patients with lower extremity disorders and clinical settings for rehabilitation therapy. Thirdly, in this study the unstable elements of different hardness are utilized, but the forefoot and rearfoot unstable elements only lead to medial and lateral instability. Future research can add unstable element to different locations combined with unstable elements of different stiffness for further analysis. Lastly, the experimental shoes chosen in this study were unstable shoes with different hardness, without an ordinary shoes for comparison. Those factors should be taken into consideration in future research.

5. Conclusion

Overall, the unstable shoes with the unstable elements leading to medial and lateral instability could increase the activity of the knee joint in the process of walking, leading to increase of knee adduction torque, but the adduction torque with the SS reduced compared with the HS. The unstable element to the rearfoot region has more significant effect on the knee compared with forefoot region in the gait cycle. The ground reaction force under the condition of two experiment shoes has no significant change. Additionally, compared with HS, plantar pressure transferred from medial foot to lateral foot when wearing SS. These results can provide some scientific evidence and suggestions for footwear companies, and also have certain application into the prevention and rehabilitation of some lower limb injuries.

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