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Effects of the shoes with a different sole hardness on the muscle mechanics of lower limbs in female runners

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Abstract

Sole hardness (SH) is an important characteristic of cushioning and shock absorption of running impacts and affects running performance. However, the effects of SH on the muscle mechanics of the lower limbs during running remain controversial. The purpose of this study was to explore the effects of different SH on the muscle mechanics of the lower limbs during the running-stance phase in female recreational runners. 16 healthy female recreational runners (age: 22.17±0.69, height: 1.64±0.03m, body mass: 50.41±1.62kg, BMI: 18.65±0.39kg/m^2, feet size: 23.94±0.39 cm) participated in this experiment. During the test, participants were asked to wear soft shoes (SS, Shore A 35), medium shoes (MS, Shore A 45), and hard shoes (HS, Shore A 55), and to run across a 2 m-long force plate at a speed of 3.3 ± 5% m/s. The force plate recorded the ground reaction force at 1000 Hz to distinguish a complete gait cycle. An eight-camera motion capture system was used to record kinematic data during the running stance phase at 200 Hz. Each participant completed 5 valid trials each time. The peak forces (PF) of the vastus medialis and vastus lateralis when wearing HS were significantly greater than those when wearing MS. Tibialis anterior PF was significantly greater in MS than in SS. SH significantly influences muscle peak force without altering impulse. Specifically, HS increases knee extensor loading while enhancing tibialis anterior activation, suggesting that footwear selection should be personalized to balance knee injury risks with ankle stability benefits.

Keywords: sole hardness, muscle mechanics, female, running

1 Introduction

Footwear is widely used to protect the foot and improve walking capability. However, the functions of shoes can vary with differences in sole characteristics, such as hardness, materials, and structure, which affect sports performance and injury risk in the lower extremity musculoskeletal system [1]. The sole is a vital shoe component for cushioning and shock absorption during running impact [2]. This is because it can influence repetitive impact forces that may contribute to microtrauma and overuse injuries [3]. Various studies reported that sole hardness (SH) influenced biomechanical variables. The softer sole reduced the impact loading rate and tibial acceleration and might thus have a protective effect on running-related injuries [4][5]. Consistently softer SH reduced loading rates and improved comfort during heel strikes by increasing the time to the first vertical ground reaction force peak [6].

Most studies of muscle mechanics have focused on muscle activation. However, a higher degree of muscle activation meant the recruitment of more motor units, thus potentially generating greater muscle contraction force. From a biomechanical perspective, the mechanical energy generated by the muscles of the lower limb joints enables the runner to fulfill the movement task [7][8]. Previous studies report that the cushioning of running shoes makes it possible to use less muscular energy. However, the well-cushioned shoes result in energy loss as the shoe midsole is compressed. Although cushioning reduces the load on the body, it may also require more muscular energy to generate propulsive force [9]. In the study on the influence of SH on muscle balance and co-contraction, the higher SH significantly reduced the muscle activity ratio between the medial and lateral gastrocnemius muscles [10]. Indeed two tuning parameters of the tendon unit model can be optimized and adjusted by muscle activity through SH [11]. James et al. also found that lower limb muscle activity used to tune the muscles for the impact task can be altered by changing the SH [12]. Causing changes in the lower limb muscle activity during locomotion may provide a mechanism for the treatment and prevention of musculoskeletal disorders, reduce muscle fatigue, and increase performance for daily activities. Some studies have shown that vastus lateralis muscle activation was significantly higher in harder shoes than in softer shoes. Gluteus maximus muscle activation was significantly lower in harder shoes than in softer shoes [9]. Previous studies on the effects of SH on muscles have mostly focused on muscle activation and energy. Muscle impulse (force-time-integral) is the total force generated over the duration of the running [13], which reflects the effect of force on the accumulation of time. Therefore, we wanted to explore the effect of SH on muscle force and impulse separately.

Despite the recognized health benefits of running, the prevalence of running-related injuries is high [14]. It's reported that the most common site of running injuries is the knee (42–46%), followed by the foot and ankle (17–30%) and lower leg (13–22%) [15]. Risk factors for running are multifaceted and include gender, training load, and biomechanical, anatomical, and anthropometric variables [16]. There are differences in bone structure and muscle force between males and females [17]. When compared to males, females are almost twice as likely to sustain a running injury, such as patellofemoral pain syndrome, iliotibial band syndrome, or gluteus medius injury [17]. Masanori et al. found a greater knee abduction in female runners, which can be linked to a high risk of knee injury [15]. Current literature investigating the effects of shoe midsole hardness on running kinematics has often been focused on male participants [18]. Therefore, in this study, women were experimental participants.

The movements of running that were affected by the shoe midsole hardness were seen more dominantly in the sagittal plane [18]. As a result, the purpose of this study was to explore the effects of different SH on the muscle peak force (PF) and impulse of rectus femoris, vastus medialis, vastus lateralis, medial gastrocnemius, lateral gastrocnemius, and tibialis anterior.

2 Material and Methods

2.1 Participants

The sample size for the within-factor repeated measures ANOVA (1 group × 3 measurements) was calculated using G*Power 3.1 (Franz Faul, Germany), taking into consideration Cohen's medium effect size = 0.5, α error probability = 0.05, and power (1 − β) = 0.8. Based on these parameters, it was estimated that a minimum of 8 participants would be required for this study. 16 healthy female recreational runners (age: 22.17±0.69, height:1.64±0.03 m, body mass: 50.41±1.62 kg, BMI: 17.66±1.27 kg/m2, feet size: 23.94±0.39 cm) participated in this experiment. The recruitment criteria for recreational runners in this experiment were running for at least 6 months and running a minimum distance of 10 km per week [19]. The dominant leg was defined as the preferred leg when kicking a ball, and the participants' right-side limb was the dominant limb. All participants were free from health problems and/ or neuromuscular disorders and/or known gait impairments, and had no lower limb injuries in the previous six months. All participants were rearfoot strikers and were recruited from Ningbo University for this study. Before the experiment, all participants gave written consent. The Ethics Committee of the Research Institute at Ningbo University approved the study.

2.2 Procedures

The general process of the experiment is shown in Figure 1(b). Before testing, participants had 10 minutes to warm up and familiarize themselves with the experimental settings, and were allowed 3 trials to get acquainted with the test maneuvers. During the test, participants were asked to wear soft shoes (SS, Shore A 35), medium shoes (MS, Shore A 45), and hard shoes (HS, Shore A 55) and ran through a 10 m track at a speed of 3.3±5% m/s. At the same time, participants were asked to complete a full gait cycle on a 2m force platform (Kistler, Winterthur, Switzerland). The force platform was located in the middle of the track. In this test, a full gait cycle was defined as from heel strike to forefoot off the ground. The shoe information is shown in Table 1. The SH was determined by the American Society of Testing and Materials standard test method D2240-04 for Rubber Property-Durometer Hardness [20]. The force plate recorded the ground reaction force at 1000 Hz to distinguish a complete gait cycle. An eight-camera motion capture system (Vicon Metrics Ltd., Oxford, United Kingdom) was used to record running kinematic data during the stance phase at 200 Hz. The force plate and the camera synchronously collected data. A full foot (heel-toe) contact with the force plate was defined as a valid trial. Each participant completed 5 valid trials each time. A threshold of 20 N on the vertical ground reaction force was applied to identify the initial foot contact and toe-off. To manage running speed, Brower timing lights (Brower Timing System, Draper, UT, USA) were used. Before the experiment, participants were asked to apply 38 reflective markers (diameter: 14 mm) to their bodies.

Table 1. Description of shoe information

	SS	MS	HS
Hardness value (Shore A)	35	45	55
Appearance			

2.3 Data analyses

OpenSim 4.3 (Stanford University, Stanford, CA, USA) was used to process and calculate biomechanical parameters during this study. A musculoskeletal model in OpenSim (gait 2392), which had 10 rigid bodies, 23 degrees of freedom, and 92 musculotendon actuators, was used. Import the statics model into the OpenSim 4.3 software. Then use the scale tool to obtain the anthropometric model of each participant. Identify the muscle starting and ending points, and ensure the moment arms are consistent with the length of the participants' limbs [21] . Using a static optimization tool with the kinematic data calculated and motion file, calculate the muscle force [22]. Muscle forces were normalized to body weight.

2.4 Statistical analyses

Tests for normality and homogeneity of variances (Shapiro–Wilk and Levine's, respectively) were conducted on all data before the analysis. One-way repeated measures ANOVA was performed to determine the effects of SH on muscle force of females during running, and Tukey's post hoc tests were used to analyze the specific significant differences. The level of statistical significance was set at $p < 0.05$.

3 Results

The changes in muscle PF and impulse in different SH shoes during the running stance phase are shown in Table 2 and Figure 1. According to the statistical results, vastus medialis and vastus lateralis PF values were significantly greater in HS than in MS in the running stance phase ($P < 0.05$). Tibialis anterior PF when wearing MS was significantly larger than that of SS in the running stance phase ($P < 0.05$). There was no significant difference in the impulse of each muscle in the running stance phase ($P > 0.05$).

Table 2. The peak force of muscles during running in different SH shoes

Variables	Muscles	SS	MS	HS	*P*-value
PF (N/kg)	Rectus Femoris	15.61±8.63	13.95±8.10	14.96±6.59	0.693
	Vastus Medialis	16.53±1.97	15.53±1.61[c]	16.94±2.54[b]	**0.023**
	Vastus Lateralis	31.56±2.92	31.02±3.55[c]	33.22±3.96[b]	**0.038**
	Medial Gastrocnemius	28.96±5.98	30.58±4.03	28.11±6.14	0.192
	Lateral Gastrocnemius	13.06±2.91	13.40±1.74	13.50±1.90	0.708
	Tibialis Anterior	11.73±1.21[b]	13.39±1.45[a]	12.45±1.44	**0.027**
Impulse (N·s/kg)	Rectus Femoris	0.95±0.55	0.82±0.32	0.78±0.42	0.247
	Vastus Medialis	1.77±0.34	1.68±0.36	1.90±0.43	0.080
	Vastus Lateralis	3.43±0.65	3.34±0.73	3.68±0.80	0.159
	Medial Gastrocnemius	1.29±0.27	1.28±0.28	1.23±0.30	0.670
	Lateral Gastrocnemius	0.55±0.08	0.58±0.10	0.56±0.08	0.428
	Tibialis Anterior	0.70±0.05	0.74±0.08	0.68±0.09	0.151

Data in bold indicate significant differences, a= there was a significant difference with the data corresponding to SS, b= there was a significant difference with the data corresponding to MS, c= there was a significant difference with the data corresponding to HS, $P < 0.05$.

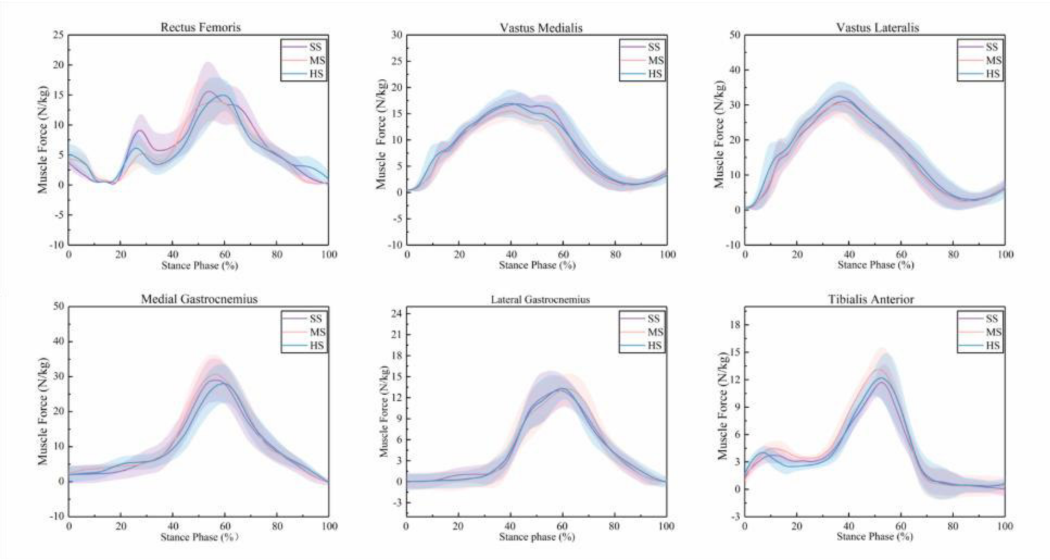

Figure 1. Illustration of the muscle force of wearing shoes with different SH during the running stance phase.

4 Discussion

When wearing HS, the PF of the PF values of the vastus medialis and vastus lateralis were significantly greater than that of MS. When running in MS, the PF of the tibialis anterior muscle was considerably greater than that of SS. Studies have shown that muscle activity was controlled by optimally adjusting two tuning parameters of the muscle-tendon unit model with respect to shoe hardness [11]. In the study of Lee et al. [9], Vastus lateralis activation was also greater when wearing harder shoes. At the same time, a higher degree of muscle activation meant the recruitment of more motor units, thus potentially generating greater muscle contraction force. Some studies have found that softer shoes improve running economy and lower energy consumption during running. Therefore, it could lead to the vastus medialis and vastus lateralis producing less propulsion force. In the study by Nicolas et al., vastus lateralis activation and force were found to increase when wearing aged shoes with higher SH [23]. Indeed, harder shoes caused significantly greater vibration through the body during running. In response to these vibrations, muscular activities were modified to minimize soft tissue vibration [24]. The higher activation and force of the knee extensor vastus lateralis muscle with harder shoes probably induce muscle stiffening, thus amplifying impact magnitude [25]. The more damage that may be done to the knee, such as an anterior cruciate ligament injury. Perhaps because the vastus medialis and vastus lateralis have similar functions, the vastus medialis and vastus lateralis in this study produced similar changes. The muscle activation and force of the tibialis anterior increased under a harder SH condition to maintain ankle stability during running [1], reducing the risk of ankle sprains. Recreational runners can run in harder shoes to train tibialis anterior force and improve ankle stability. The lower limb muscle activity responded to the different SH in a subject-specific manner. This can be explained by the fact that the vibration characteristics of soft tissues vary between individuals and are affected by factors like tissue mass, subcutaneous fat content, and blood pressure [12]. Developing a universal shoe that regulates muscle activity in the same way would be difficult. But there was no doubt that SH caused muscle mechanics' change. Therefore, this also provides an idea for customizing shoes according to individual-specific circumstances. A significant difference was not found in muscle impulse. It indicated that SH did not influence the accumulation of force during the stance phase of running.

5 Conclusion

In conclusion, shoe hardness significantly alters muscle peak force without affecting impulse during the running stance phase. Specifically, harder shoes increase loading on the knee extensors (vastus medialis and lateralis), potentially increasing the risk of knee injury, while also enhancing tibialis anterior activation, which benefits ankle stability training. These findings indicate that runners modulate force magnitude rather than duration to adapt to shoe properties. Therefore, footwear selection should be customized to balance individual injury risk and training goals.

6 References

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