Movement Analysis of Lower Limb during Backward Walking with Unstable Intervention

Running title: Movement characteristic of backward walking

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ABSTRACT
Backward walking (BW), an emerging rehabilitative and training modality, was integrated with unstable sole construction with various hardness levels to analyze the kinematic and kinetic characteristics of the lower extremities. Eighteen participants volunteered to participate in the test. They performed walking tests under three conditions: 1) BW with normal shoes (NBW); 2) BW with unstable shoes with soft unstable elements (UBW-S); 3) BW with unstable shoes with hard unstable elements (UBW-H). The results show increased hip and ankle flexion and increased knee flexion-extension extent in the stance phase during BW with unstable shoes. The motor control mechanism of unstable BW enhanced the rehabilitation of lower limb deficiency. The attached unstable elements (UBW-S and UBW-H) induced local perturbation to stimulate proprioceptive ability and the neuromuscular system, changing the plantar loading distribution in a certain region. Future study should concentrate on the possible rehabilitative effect of unstable BW on neurological disorders and motor system deficiency.
Keywords: Backward walking, Unstable shoes, Gait analysis, Rehabilitation

1. INTRODUCTION

Backward walking (BW) has gained increasing popularity as a modality of rehabilitative training for patients with motor system deficiency and neurological disorders. Previous research suggested that BW can improve the muscular strength of lower extremities and locomotion balance [1,2], owing to the different motor control mechanism from that for forward walking (FW) [3,4] with greater reliance on neuromuscular control, proprioception, and protective reflexes [5]. Clinical studies have shown that BW treatment has positive effects on gait rehabilitation, with evenly distributed plantar loading for diabetic peripheral neuropathy patients, and improved motor control and asymmetric gait patterns for hemiplegic or stroke and knee osteoarthritis patients [6-9]. BW locomotion can be clinically useful for reducing stress on injured joints, particularly the knee (patellofemoral) joint [10,11]. Comparisons of the differences in lower limb biomechanics between BW and FW [1,2,12] have shown advantages of BW for patients with anterior cruciate ligament deficiency and knee osteoarthritis. The control mechanism for BW is not simply the reverse of the FW mechanism [3,13,14,15]. BW has thus been integrated by clinical therapists into conventional gait rehabilitative training sessions, including those with FW, to avoid asymmetric gait patterns and enhance activation of lower limb muscles [5,16].

Footwear can affect muscle activity level and the kinematic and kinetic characteristics of lower extremities (including ankle, knee, and hip joints) through external manipulation (i.e., the outsole forming unstable structures). The outsole of a common unstable shoe is characterized by a convexity that provides an unstable support base while standing or walking. Examples include Masai Barefoot Technology (MBT) and wobble-board unstable shoes. In addition, different sole constructions (high-heeled, negative heel, round, shoe lifts, etc.) influence the motor patterns of lower limbs, causing distinct changes of gait kinematics and muscle activity [17,18,19]. Based on previous studies, the main function of unstable shoes is increasing lower limb muscle strength and coordination, enhancing postural control, reducing perceived pain level, and rehabilitating lower limb and lower back injuries [4,17,18,20-25]. It has been reported that unstable shoes (MBT) reduce joint pain through effective shock absorption for people with knee osteoarthritis and increase static balance [26]. Another type of unstable footwear with convexity in the heel and forefoot part of the outsole was shown to stimulate the neuromuscular system and muscle activity in the lower extremities, thus increasing postural control and proprioceptive adjustment [17]. The plantar loading distribution was altered for falling prevention and malfunction rehabilitation, as the position of unstable elements varies in the coronal plane [27]. Moreover, Li et al. [28] found that compared with unstable shoes with hard unstable elements, plantar loading, in terms of impulse and maximal force, on the medial and central forefoot regions was alleviated obviously when wearing unstable shoes with soft unstable elements, which is beneficial for certain therapy shoe designs.

Both BW and unstable shoes have practical implications for rehabilitation of lower limb disorders or malfunctions. However, few studies have focused on the effect of unstable shoes on lower limb kinetics and kinematics during BW locomotion, which could potentially combine the effects of unstable shoes and BW locomotion, thus increasing the rehabilitative effect with an integrated stimulus of unstable perturbations and elimination of visual cues.

By altering the material stiffness of unstable elements to change stability, this study investigated lower limb kinematics and kinetics under three conditions: 1) BW with normal shoes (NBW); 2) BW with unstable shoes with soft unstable elements (UBW-S); 3) BW with unstable shoes with hard unstable elements (UBW-H). The aim is to investigate the effect of unstable elements with different hardness levels on lower limb biomechanics during BW, with hypothesis that BW with unstable shoes (UBW) can effectively alter the kinematic and kinetic characteristics of lower limbs, which could contribute to the understanding of UBW gait.

2. METHODS

2.1 Participants

Eighteen healthy males (age: 24±1 years, height: 174.1±1.3 cm, mass: 68.0±2.4 kg) volunteered to take part in the test. All participants were university students. Participants were free of pain and injury and had not undergone major surgery on the lower limbs or lower back in the past six months. They were informed of the experimental procedure and objectives and gave written consent. This study was approved by the Ethics Committee of Ningbo University.

2.2 Shoes

All participants wore the same shoes during the tests. The control shoes were normal shoes with a flat sole, and the experimental shoes were made manually with unstable elements attached to the outsole of the heel and forefoot zones of the normal shoes. The unstable elements were in the form of two hemispheres with a height of 1.5 cm and a diameter of 5.5 cm (Fig. 1). The elements were made of rubber, whose elastic modulus was measured using an elastic modulus test system (INSTRON AG Grove, USA) (Fig. 2) [28]. Unstable elements with two elasticity levels were adopted for the soft and hard shoes (shoes-S and shoes-H, respectively).

2.3. Experiment protocol

The tests were performed in the Sports Biomechanics Laboratory of Ningbo University. A Vicon 8-camera motion analysis system (Oxford Metrics Ltd., Oxford, UK) was used to capture three-dimensional kinematics at a frequency of 200 Hz. A standard reflective marker set was used to define joint centers and axes of rotation. Participants were required to wear tight shorts. 16 reflective points (diameter: 14 mm) were attached with adhesive on the left and right lower limbs, respectively, over anatomical landmarks, namely the anterior-superior iliac spine, posterior-superior iliac spine, lateral mid-thigh, lateral knee, lateral mid-shank, lateral malleolus, second metatarsal head, and calcaneus. The markers for the heel and toes were adhered on the corresponding anatomical points on the shoes. A force platform (Model 9281B, Kistler, Switzerland) was used to collect ground reaction forces (GRFs) at a frequency of 1000 Hz. It was connected with the Vicon software to enable simultaneous testing. A Novel Pedar System (Germany) was utilized to collect plantar pressure data. The insoles were placed inside the shoes before testing. The plantar was divided into eight areas based on the anatomy of the foot, namely the medial rearfoot (MR), lateral rearfoot (LR), medial midfoot (MM), lateral midfoot (LM), medial forefoot (MF), lateral forefoot (LF), hallux (H), and other toes (OT). Plantar pressure parameters were peak pressure, contact area, and pressure-time integral.

2.4 Procedures

Participants walked along a 10-m walkway before testing in order to become familiar with the required walking speed and to adjust their gait so that the right foot could land on the Kistler force platform naturally. Walking speed was measured with a timing meter and a metronome; it was controlled at 1±0.2 m/s. Three testing conditions were set: 1) BW with normal shoes (NBW); 2) BW with shoes-S (UBW-S); 3) BW with shoes-H (UBW-H). Each participant performed 5 trials for each walking condition (selected randomly) to achieve good consistency and stability of gait to reduce experimental error. One BW gait cycle was defined from one ipsilateral forefoot contact with the ground to the subsequent one. The spatiotemporal parameters were stride length, stride time, and contact time; kinematic data were angle changes of lower limb joints (hip, knee, and ankle) in three planes (sagittal, coronal, and horizontal) during one gait cycle; kinetic parameters were GRF, peak pressure, contact area, and pressure-time integral. Stride length and stride time are the distance and duration between two successive forefoot strikes of a given foot, respectively. Contact time is the duration of the stance phase. In this test, the forefoot landed firstly on the walkway under the three BW conditions.

2.5 Statistical analysis

Statistical analysis was performed with STATA 12.0 software. One-way analysis of variance (ANOVA) with post-hoc Bonferroni correction was performed to investigate the differences between the main variables of interest. Statistical results were considered significant if $p < 0.05$. **3. RESULTS**

A comparison of spatiotemporal parameters between paired conditions is shown in Table 1. There were significant differences in contact time between NBW and UBW-S ($p < 0.05$).

3.1. Kinematics

The joint angles of the hip, knee, and ankle in three planes (sagittal, coronal, and horizontal) during one gait cycle under the three walking conditions are illustrated in Figs. 3(a)-(c). During the stance phase, significant differences were observed between NBW and UBW (both UBW-S and UBW-H) in the sagittal plane of the ankle, hip, and knee as well as the horizontal plane of the hip (highlighted with rectangles in Fig. 3). During the swing phase, the hip for NBW showed significantly larger peak flexion than that for UBW. The ankle for UBW-H showed significantly larger peak dorsiflexion than those for NBW and UBW-S. The knee for UBW-H showed significantly less peak internal rotation than those for the other two (NBW & UBW-S) conditions.

3.2 Kinetics

3.2.1 GRF

There were two main peak forces in the vertical GRF (vGRF) during the stance phase of walking. For BW, the first peak was initiated by the passive impact of forefoot contact and the second peak was caused by the initiative impact of heel-off (Fig. 4). No significant differences were found in either the first or second peak GRF among NBW, UBW-S, and UBW-H (NBW and UBW-S: $p = 0.226$; NBW and UBW-H: $p = 0.173$; UBW-S and UBW-H: $p = 0.089$).

3.2.2 Peak pressure, contact area, and pressure-time integral

For peak pressure, significant differences were found between NBW and UBW-S in MF and LF and between NBW and UBW-H in MF, LF, and LM (Fig. 5(a)). The peak pressure of the forefoot for NBW was significantly less than those for UBW-S and UBW-H.
For contact area, significant differences were found between NBW and UBW-H in MF, LF, and LM (Fig. 5(b)). The contact area of the forefoot for NBW was significantly larger than that for UBW-H. However, there were no significant differences between NBW and UBW-S for the

contact area

For the pressure-time integral, significant differences were found between NBW and UBW-S in MF, LF, MM, MR, and LR, between NBW and UBW-H in OT, MF, LF, MM, MR, and LR, and between UBW-S and UBW-H in MF and LF (Fig. 5(c)). The pressure-time integrals of both the forefoot and heel for NBW are significantly less than those for UBW-S and UBW-H. The forefoot for UBW-S showed a lower pressure-time integral than that for UBW-H.

4. DISCUSSION

This study compared the biomechanical characteristics of lower limbs during BW with unstable and normal shoes. The walking conditions clearly affected the contact time (i.e., that between NBW and UBW-S), which is in agreement with the results reported by Vilensky et al. [4] and Kramer et al. [23]. Kinematic data of the lower limb joints during BW were also consistent with the findings of previous studies [1,27,29]. The plantar loading distribution characteristics were different for UBW-S and UBW-H.

Compared with BW with normal shoes (NBW), UBW changed the kinematics greatly, particularly movement in the sagittal plane. BW with soft shoes (UBW-S) and hard shoes (UBW-H) showed greater extension in the hip and plantarflexion in the ankle, and a larger extent of flexion-extension in the knee. The increased hip extension in the stance phase can be explained by BW partially employing a reverse control mechanism (central pattern generator (CPG)) compared to that of FW [4], which increased hip flexion at the initial contact with toning shoes [30]. Regarding the smaller peak flexion angle in the swing phase, the self-protection function of neuromuscular control might be responsible, as it prevents falling, owing to the elimination of visual cues [5]. The limited abduction movement in the stance may be the result of neuromuscular control to maintain global (whole-body) and local (regional body part) stability with unstable perturbations from the outsole [27]. This reverse control mechanism (CPG) may be responsible for the greater ankle plantarflexion in the stance phase; in contrast, FW with unstable shoes presented greater dorsiflexion [4,31]. In previous research of FW with unstable shoes, Nigg et al. [31] reported that walking with unstable shoes increased ankle dorsiflexion. Due to BW is the time-reversed counterpart for FW, it can be inferred that UBW increases ankle plantarflexion, which is consistent with the kinematic results in this study.

The effect of outsole material stiffness can be eliminated, as the knee loading (adduction moment) showed no significant difference while performing a FW gait with this footwear in a previous study [28]. Accompanied with the constant vGRF (see Fig. 4), the enlarged extent of knee movement in the sagittal plane of the stance phase implies reduced knee joint loading during BW with unstable shoes [13,26], which agrees with a previous report of a reduction of knee loading during a BW gait [32]. Although Lee et al. [1] found that BW can reduce the knee flexion angle with simply reversed kinematic data characteristics comparing with FW, their data are different from the kinematics data obtained in this study. Moreover, the motor control mechanisms of FW and BW are similar only for the hip and ankle joints [3,4,24]. This should be verified in a future study.

Apart from the joint loading alleviation of BW, Tommy et al. [33] suggested that BW can reduce lower back pain and enhance the function of lumbar musculature in postural stability and dynamic function, with exhibition of greater lower back motion in the sagittal plane and lesser motion in the coronal plane after a BW exercise program. Combining the hip motion difference in the coronal plane found in this study, it can be inferred that UBW with restricted hip abduction motion can alleviate lower back pain and enhance lumbar muscle function in core stability control, thus improving sports performance [33,34].

As to kinetic parameters, the vGRF during BW with normal flat shoes and unstable shoes presented patterns similar to those previously reported [1,11,32]. The second peak of vGRF was smaller than the first peak, which is believed to result from the active lift-off movement, with the knee and hip joints lifting the limb and moving it backward [1,4]. This could contribute to reduced knee joint loading, particularly by the patellofemoral joint compressive force, as it is different from the push-off movement (increased loading) of FW or running [10,35].

In terms of the altered plantar loading distribution with UBW, MF and LF presented increased peak pressure but reduced contact area, opposite with the results of LM, which can be easily explained by the unstable element attached to the forefoot region [27]. This adjustment to external unstable perturbations stimulates the neuromuscular system to control local stability, thus improving global balance maintenance [17,22,29] and offloading (reduce even remove loading to) specific region with extreme impact or ulceration risks of certain diseases [12,27,28]. Stewart et al. [36] observed that plantar pressure decreased in the heel region and increased in the forefoot and tiptoe regions during FW with unstable shoes. The above-mentioned effect of locomotion with unstable footwear indicates that unstable shoes can adjust plantar loading combined with altered biomechanics of BW. The pressure-time integral for the forefoot (MF and LF) increased obviously from NBW to UBW-S, and again to UBW-H, reflecting the increase in the unstable element's hardness, and thus greater loading on medial and central metatarsal parts [28]. In addition, the forefoot area was more sensitive to the hardness of unstable elements than was the rearfoot area, even though the unstable elements were attached to both the forefoot and rearfoot regions. The pressure-time integral in the rearfoot (MR and LR) regions shows no consistency with that in the forefoot during FW, which can be explained by the different landing patterns (toe-heel for BW and heel-toe for FW). The forefoot strike of UBW was consistent with the kinematic result of a larger ankle plantarflexion angle in the stance during UBW compared to that for NBW. The greatly reduced impulse loading on the MM part is of great importance for the alleviation of mid-foot pain, and thus could be a rehabilitative training protocol [12,27].

5. CONCLUSION

This study analyzed the biomechanical characteristics of BW with unstable shoes with soft and hard stiffness. As hypothesized, a significant difference between NBW and UBW existed in the kinematics and plantar pressure data, which is beneficial for understanding the UBW mechanism. The reversed motor control mechanism of UBW compared to that of FW enhances the rehabilitation of lower limb deficiency. The attached unstable elements with different hardness levels induced local perturbations, which stimulated the proprioceptive ability and neuromuscular system, and changed the plantar loading distribution in certain regions. Further studies should be carried out with neurological disorder or gait motor deficient patients to determine the rehabilitative effects of UBW.

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Fig. 1. Illustration of normal shoes and experimental shoes-S and shoes-H.

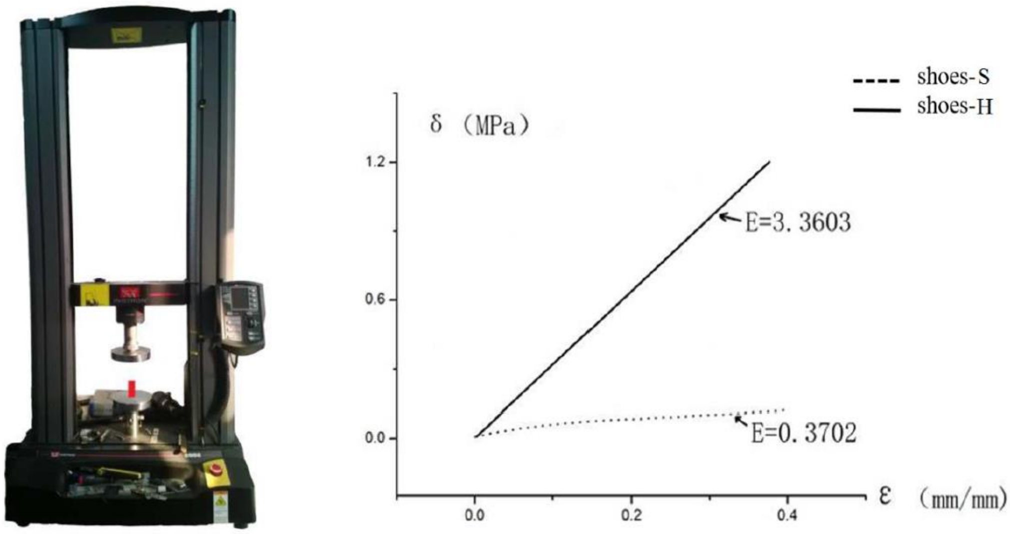


Fig. 2. Elastic modulus test system (left) and elastic modulus (Young’s modulus, E) of hard and soft materials used for experimental shoes (right).

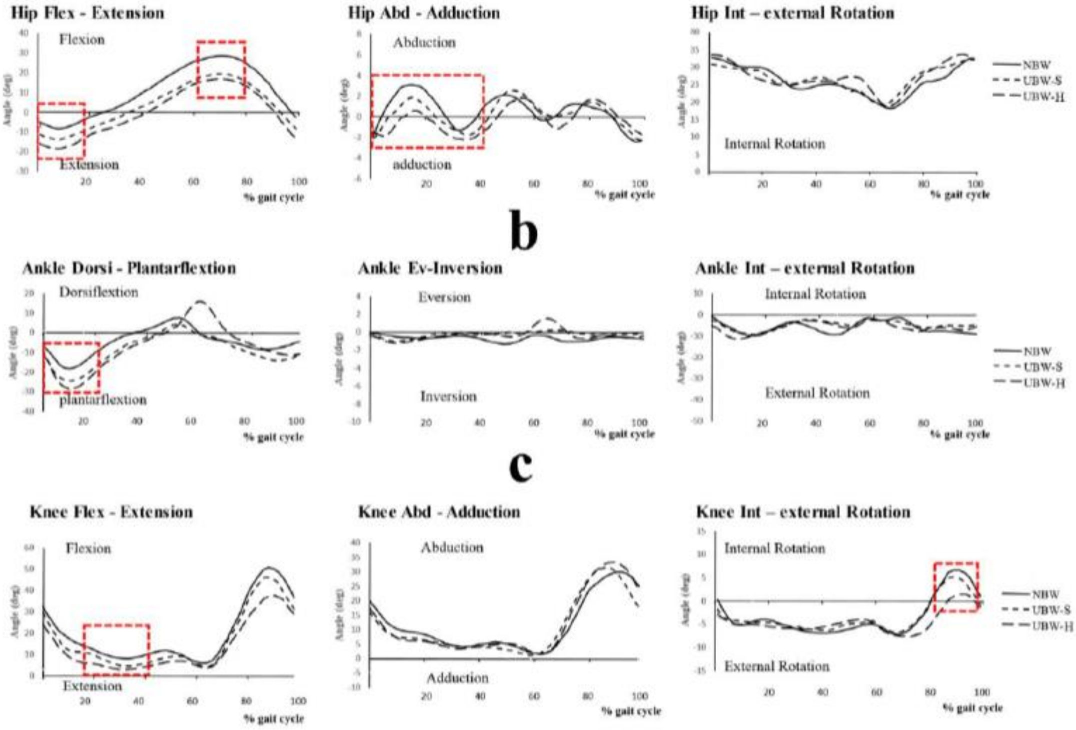


Fig. 3. Joint angles curves for (a) hip, (b) ankle, and (c) knee in sagittal, frontal, and horizontal planes during one gait cycle. Red squares indicate significant difference between NBW and UBW (both UBW-S and UBW-H).

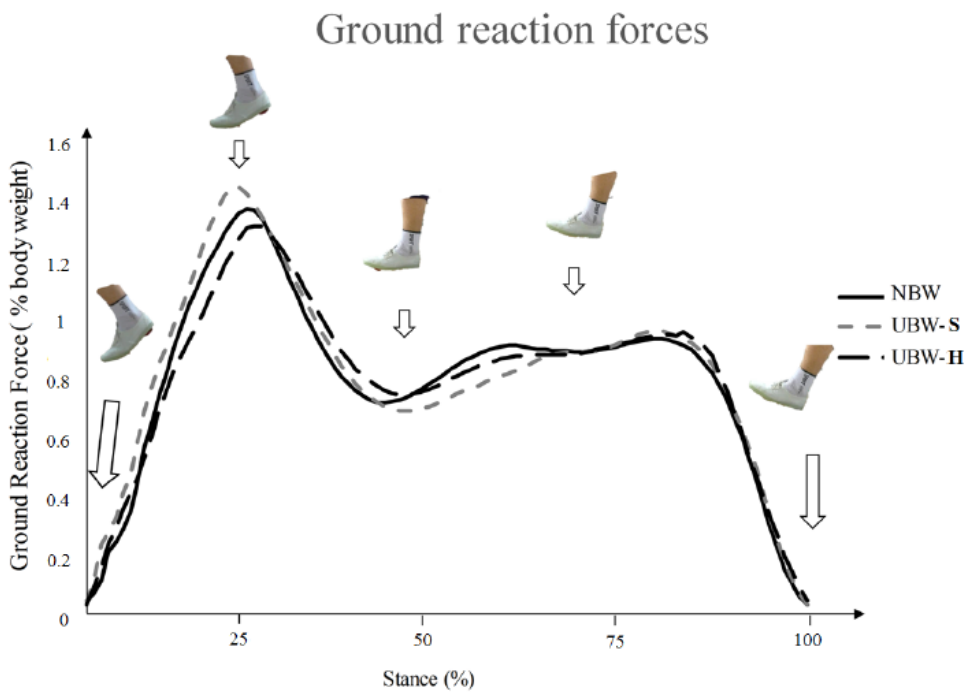
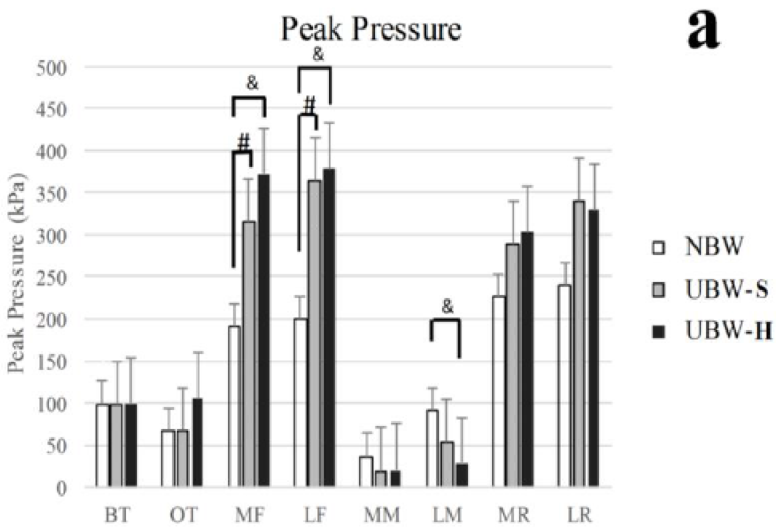


Fig. 4. Mean resultant vGRF for stance phase while walking for various experimental conditions.



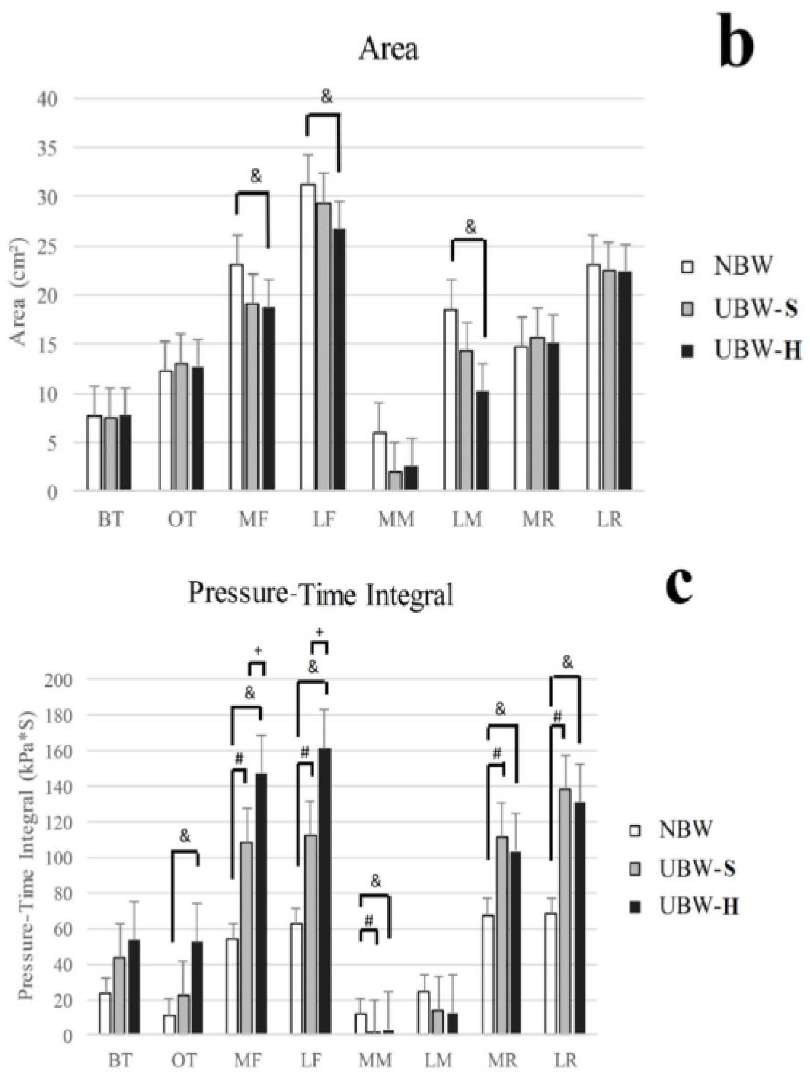


Fig. 5. Comparison of (a) peak pressure, (b) contact area, and (c) pressure-time integral.

Note: #, &, and + indicate significant differences ($p < 0.05$) for comparison between NBW and UBW-S, between NBW and UBW-H, and between UBW-S and UBW-H, respectively.

Table. 1 Spatiotemporal parameters (n = 18)

		NBW	UBW-S	UBW-H
Stride length (m)	mean	0.94	0.98	1.01
	SD	0.042	0.035	0.01
Stride time (s)	mean	1.18	1.05	1.21
	SD	0.000	0.113	0.014
Contact time (s)	mean	0.78#	0.73	0.77
	SD	0.014	0.007	0.048