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# Effects of Short-Term Limitation of Movement of the First Metatarsophalangeal Joint on the Biomechanics of the Ipsilateral Hip, Knee, and Ankle Joints During Walking

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Data Interpretation D  
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**Background:** We analyzed the effect of limitation of movement of the first metatarsophalangeal joint (FMJ) on the biomechanics of the lower limbs during walking.





**Material/Methods:** Eight healthy college students completed walking under barefoot (BF) and FMJ constraint (FMJC) conditions. We synchronously collected kinematics and dynamics data, and calculated the torque, power, and work of hip, knee, and ankle joints.

**Results:** Compared with normal conditions, when the FMJ is restricted from walking, the maximum ankle dorsiflexion angle is significantly increased ( $P < 0.001$ ), the maximum plantar flexion angle is significantly reduced ( $P < 0.001$ ), the maximum plantar flexion torque ( $P < 0.001$ ) and the maximum dorsiflexion torque ( $P < 0.05$ ) increased significantly, the maximum power increased significantly ( $P < 0.001$ ), the minimum power decreased significantly ( $P < 0.001$ ), and the negative work increased significantly ( $P < 0.001$ ). The torque of hip and knee joints increased significantly ( $P < 0.05$ ).

**Conclusions:** After the movement of the FMJ is restricted, the human body mainly compensates and transfers compensation by increasing the angle of dorsiflexion, increasing work and the activity level of surrounding muscles through the ankle joint, thereby increasing the torque load of the knee and hip joints to maintain the dynamic balance of kinematics. FMJC condition increases the energy consumption of the human ankle, knee, and hip joints during walking. The load is compensated by the gradual attenuation of the ankle, knee, and hip. Long-term limitation may cause damage to the posterior calf muscles and increase the incidence of knee arthritis.

**Keywords:** **Biomedical Research • Compensation and Redress • Exercise Movement Techniques • Gait • Metatarsophalangeal Joint**

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## Background

During walking, the lower-limb movements are realized by multiple joints. If the first metatarsophalangeal joint (FMJ) is restricted, such as hallux valgus or hallux rigidity, rheumatoid arthritis, diabetes, and other diseases involving the metatarsophalangeal joints, patients often experience decreased mobility of the metatarsophalangeal joints and increased pressure on the forefoot plantar during walking [1-3]. The movement pattern of these patients is a typical restricted movement of the forefoot [4,5]. The literature has shown that the limited range of motion of the first metatarsophalangeal joint may change gait performance, such as walking speed, stride length, and joint kinematics of the lower limbs [6,7]. When a joint has dysfunction, in order to complete the action, the adjacent shutdown will make an additional reaction, which is called compensation [8,9]. The transmission of joint motion compensation affects dynamic parameters such as joint load [10]. There are FMJ lesions in many patients with hip, knee, and ankle lesions, especially those with knee arthritis [11]. At present, there is a lack of literature on this relationship, according to the preliminary experimental basis [12]. We hypothesize that FMJ function limitation leads to joint compensation and biomechanical effects, which increase the risk of knee arthritis.

In the design of anthropomorphic robots and prostheses, the foot is often taken as a whole without considering the movement of the metatarsophalangeal joints [13]. Research on whether the limited range of motion of the metatarsophalangeal joint will affect the walking gait is mostly the analysis of kinematics or plantar pressure when walking after the limited motion of the metatarsophalangeal joint [14-16]. Few studies have integrated kinematics and dynamics to systematically analyze the effects of restricted movement of the first metatarsophalangeal joint on the biomechanical characteristics of the lower limbs during walking. The present study will help to understand how the human body performs joint compensation and compensation delivery, to further understand the function of the first metatarsophalangeal joint in human motion, and provide a certain reference for improvement of the surgical plan, postoperative rehabilitation, and the design of prostheses.

## Material and Methods

### Inclusion and Exclusion Criteria

We enrolled 8 healthy young people: 4 men and 4 women, mean age  $23.2 \pm 1.1$  years, mean height  $172.6 \pm 3.1$  cm, mean weight  $66.7 \pm 6.5$  kg, and mean shoe size Euro 40. Subjects did not engage in vigorous exercise for 48 h before the experiment. It was confirmed that there was no obvious damage to

the lower limbs and feet within the last 6 months. All subjects had normal movement of the first metatarsophalangeal joint, without any disease. The dominant foot of all subjects was the right foot. Before the experiment, the subjects signed an informed consent form. This experiment was approved by the Second Hospital of Jilin University Ethics Committee (2020046).

### Study Design

The first metatarsophalangeal joint constraint (FMJC) used in the experiment is composed of a low-temperature thermoplastic sheet brace (medical polymer splint), medical gauze, and Velcro provided by Convalesce<sup>®</sup> (Figure 1). During the experiment, an FMJC was made according to the foot shape of each subject, so that the movement of the first metatarsophalangeal joint was restricted.

According to the commonly used Helen Hayes motion capture model [17], during the test, 19 reflective landmarks were attached to the subject: left/right anterior superior iliac spine, midpoint of the fourth and fifth lumbar spinous processes, left/right thigh anterior, left/right tibial tuberosity, left/right fibula lateral malleolus, left/right tibia and medial malleolus, left/right toe, left/right heel, and toe point, and left/right first metatarsophalangeal joint. The data acquisition system consists of a Vicon three-dimensional optical motion capture system including 6 MX infrared cameras and 4 AMTI biomechanics force plates. The acquisition frequency of the force plate is 1000 HZ and the size is 450×500 mm. The gait walking test bench has a total length of 5m, a width of 1 m, and a height of 0.8 m. The 4 force plates are embedded in the walking test bench, and the ground angle level of the force plate can be adjusted to ensure that the force plate is flush with the surface of the test bench. An accompanying protection device is installed above the walking test bench. When the subject is walking, the operator manually controls the protection device so that the speed of the device can be changed with the change of the subject's pace. Therefore, the device can walk with the subject without disturbing the subject's natural gait, and the subject can be supported by the protective belt to ensure safety when the subject slips.

### Experimental Process

Each subject completed walking barefoot with immediate limitation of the first metatarsophalangeal joint and 30 min of limitation of the first metatarsophalangeal joint. Before the start of the experiment, the Vicon<sup>®</sup> system was calibrated, 19 reflective balls were pasted on the subjects' lower limbs, and the subjects were allowed to stand still in the test area with their arms slightly extended for about 2 s, so as to establish a static model for the subjects. Before data collection, the subjects were given appropriate practice time to adapt to



**Figure 1.** The first metatarsophalangeal joint constraint (FMJC) consists of a low-temperature thermoplastic sheet brace, medical gauze, and Velcro.

the test environment and the position of the force plate, so that the feet were stepped on the center of the force plate as much as possible, and data were collected after the exercise. The subjects were required to complete the test in a naturally relaxed state, and each set of data collected should contain 3 to 4 gait cycles. Each subject was required to complete 10 walking tests while barefoot (BF), with immediate limitation of the first metatarsophalangeal joint and 30 min of limitation of the first metatarsophalangeal joint. Subjects rested for 2 min after every 2 tests to prevent physical fatigue from affecting the subjects' walking gait. The walking speed of all subjects was the normal walking speed.

The three-dimensional coordinates of all the marker points collected were smoothed by Butterworth low-pass filtering, and the cutoff frequency was 10 Hz. The human body link coordinate system was established according to the coordinates of the marker points [18]. The hip joint center was calculated according to Bell's research [19], the center of rotation of the knee joint was the midpoint of the medial and lateral condyles of the femur, the center of rotation of the ankle was the midpoint of the medial and lateral malleolus, and the center of rotation of the metatarsophalangeal joint was the center of the first and fifth metatarsophalangeal joints. The Euler angle method was used to calculate the three-dimensional

angles of the hip, knee, and ankle, and the inverse dynamic method was used to calculate the three-dimensional net torque of the joint [20]. The human body inertia parameter uses the Zatsiorsky-Seluyanov's human body inertia parameter modified by DeLeva [21]. The joint power is the product of the joint torque and the joint angular velocity. The trapezoid method was used to calculate the joint power over time to obtain the joint power. The present study only analyzed the human sagittal plane data, and normalized each indicator according to a gait cycle (right toe off the ground to right toe off the ground again).

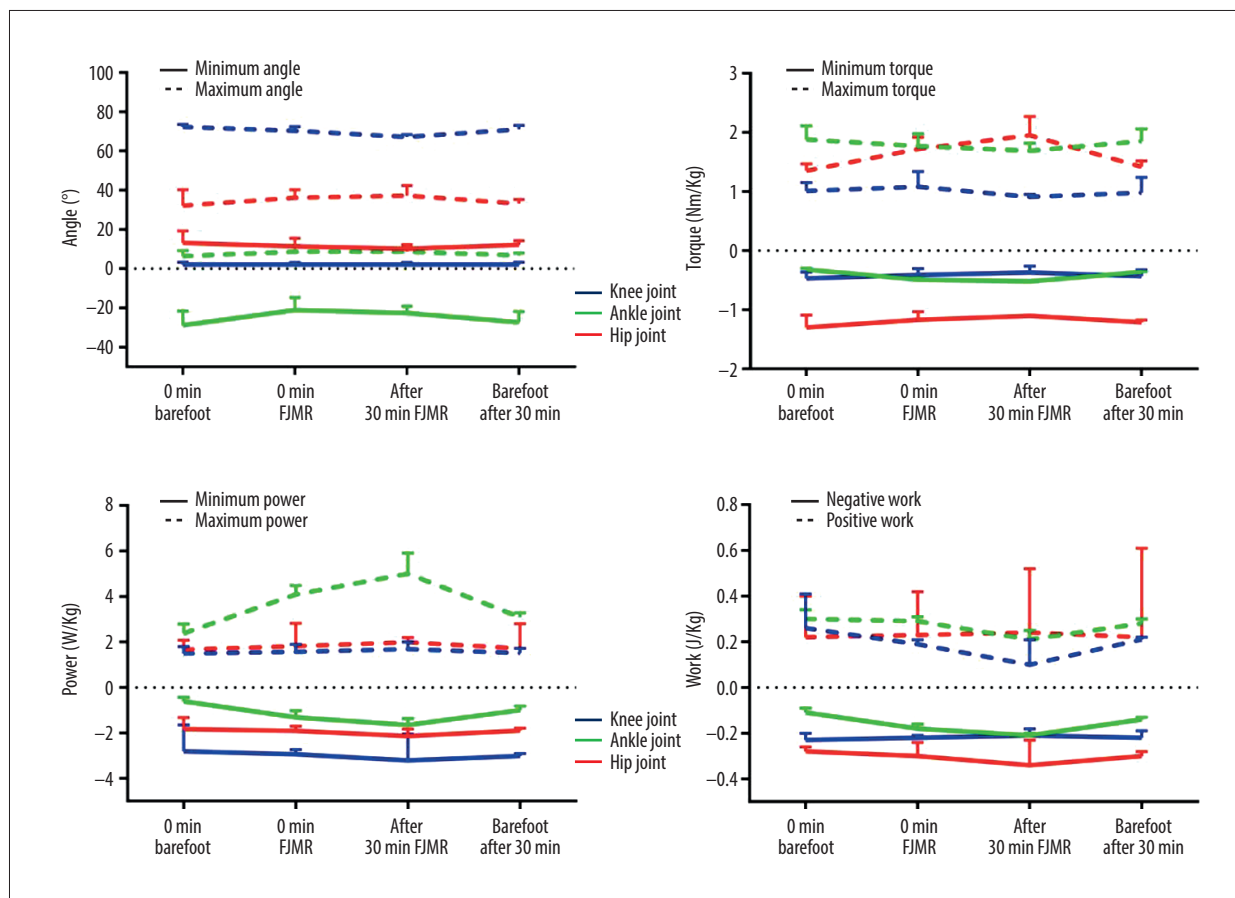
#### Data Analysis

This test needs to analyze 6 spatiotemporal parameters (step speed, stride length, stride frequency, stride time, single support period, and double support period) to avoid splitting multiple dependent variables into single factors for testing, resulting in increased type I errors. To improve the inspection efficiency, the Hotelling  $T^2$  test was performed on the collected spatiotemporal parameters [22]. For other parameters, the difference between FMJC and BF was analyzed by paired-sample  $t$  test.  $P < 0.05$ ,  $p < 0.01$  and  $P < 0.001$  were considered statistically significant, and the data processing was performed using SPSS24.0.

**Table 1.** Comparison of gait parameters between normal walking, walking immediately after FMJC, and walking 30 minutes after FMJC (n=8).

|                        | Step length (cm) | Step width (cm) | Step frequency (steps/s) | Percentage of swing phase (%) | Percentage of support phase (%) |
|------------------------|------------------|-----------------|--------------------------|-------------------------------|---------------------------------|
| Normal                 | 82.31±4.5        | 12.55±4.31      | 1.89±0.10                | 0.361±0.021                   | 0.610±0.015                     |
| Immediately after FMJC | 81.92±4.41       | 14.71±3.12      | 1.88±0.21                | 0.389±0.029                   | 0.582±0.021                     |
| 30 minutes after FMJC  | 79.11±2.3*#      | 15.41±3.11      | 1.85±0.31                | 0.401±0.031                   | 0.571±0.025                     |

Compared with the normal group \* p<0.01; Compared with immediate limitation # p<0.01.



**Figure 2.** Comparison of parameters of ankle, knee, and hip joints during normal walking, walking immediately after FMJC, and walking 30 minutes after FMJC (n=8).

## Results

### Gait Parameters

Compared with the normal situation, there was no significant difference in the step width, swing phase, support phase, step length, and step frequency of the ankle, knee, and hip when walking on the first metatarsophalangeal joint ( $P>0.05$ ) (Table 1).

### Kinematics and Biomechanical Parameters

#### Ankle Joint

The results of kinematics and dynamics parameters of each joint of the lower limbs when walking in BF and FMJC conditions (Figure 2, Table 2) show that, under restricted conditions, the maximum ankle dorsiflexion angle was significantly greater than the BF case ( $P<0.001$ ), and the maximum plantar flexion

**Table 2.** Comparison of parameters of ankle, knee, and hip joints during normal walking, walking immediately after FMJC, and walking 30 minutes after FMJC (n=8).

| Parameters             | Ankle joint     |                        |                       | Knee joint     |                        |                       | Hip joint      |                        |                       |
|------------------------|-----------------|------------------------|-----------------------|----------------|------------------------|-----------------------|----------------|------------------------|-----------------------|
|                        | Normal          | Immediately after FMJC | 30 minutes after FMJC | Normal         | Immediately after FMJC | 30 minutes after FMJC | Normal         | Immediately after FMJC | 30 minutes after FMJC |
| Maximum angle (°)      | 6.47<br>±2.91   | 5.51<br>±3.29*         | 6.11<br>±2.91*#       | 72.37<br>±1.36 | 70.25<br>±2.14*        | 67.21<br>±1.31*       | 32.15<br>±8.12 | 36.13<br>±4.15**       | 38.35<br>±5.12**      |
| Minimum angle (°)      | -28.81<br>±7.31 | -21.01<br>±6.49***     | -22.51<br>±3.61*      | 2.16<br>±1.13  | 2.17<br>±1.01          | 2.21<br>±1.03         | 13.16<br>±6.21 | 11.41<br>±4.12*        | 10.16<br>±2.21*       |
| Maximum torque (Nm/Kg) | 1.88<br>±0.23   | 1.77<br>±0.21*         | 1.69<br>±0.13**       | 1.01<br>±0.14  | 1.08<br>±0.26          | 1.41<br>±0.04*        | 1.35<br>±0.12  | 1.72<br>±0.2*          | 1.95<br>±0.32**       |
| Minimum torque (Nm/Kg) | -0.32<br>±0.02  | -0.49<br>±0.03*        | -0.58<br>±0.01***     | -0.47<br>±0.11 | -0.41<br>±0.11         | -0.31<br>±0.11*       | -1.3<br>±0.21  | -1.17<br>±0.14         | -1.10<br>±0.01        |
| Maximum power (W/Kg)   | 2.38<br>±0.41   | 4.1<br>±0.39**         | 5.0<br>±0.91***       | 1.49<br>±0.31  | 1.57<br>±0.33          | 1.73<br>±0.31*        | 1.67<br>±0.41  | 1.82<br>±1.01          | 1.99<br>±0.21*        |
| Minimum power (W/Kg)   | -0.61<br>±0.18  | -1.31<br>±0.29***      | -1.65<br>±0.28***     | -2.81<br>±1.16 | -2.94<br>±0.21         | -3.41<br>±1.16*       | -1.83<br>±0.51 | -1.91<br>±0.21         | -2.13<br>±0.31*       |
| Positive work (J/Kg)   | 0.3<br>±0.04    | 0.29<br>±0.02          | 0.21<br>±0.04*#       | 0.26<br>±0.15  | 0.19<br>±0.02          | 0.10<br>±0.11*        | 0.22<br>±0.18  | 0.23<br>±0.19          | 0.24<br>±0.28         |
| Negative work (J/Kg)   | -0.11<br>±0.02  | -0.18<br>±0.02*        | -0.21<br>±0.01**      | -0.23<br>±0.03 | -0.22<br>±0.01         | -0.21<br>±0.03        | -0.28<br>±0.02 | -0.30<br>±0.06         | -0.34<br>±0.11        |

Compared with the normal group \*  $p < 0.05$ , \*\*  $p < 0.01$ , \*\*\*  $p < 0.001$ ; Compared with immediate FMJC #  $p < 0.05$ .

angle was significantly smaller than the BF case ( $P < 0.001$ ); The maximum plantar flexion torque of the ankle joint ( $P < 0.001$ ) and the maximum dorsiflexion torque ( $P = 0.029$ ) were significantly greater than the BF case. The maximum value of ankle joint power was significantly greater than that of BF ( $P < 0.001$ ), and the minimum was significantly smaller than that of BF ( $P < 0.001$ ). The negative work of the ankle joint was significantly greater than that of BF ( $P < 0.001$ ), and there was no significant difference between the 2 cases of positive work ( $P > 0.05$ ).

#### Knee Joint

The maximum flexion angles of the knee joint at the 2 FMJC time points were significantly greater than normal ( $P < 0.05$ ). Compared with normal conditions, the maximum torque, minimum torque, maximum power, minimum power, and positive power after 30-min FMJC were significantly different ( $P < 0.05$ ). Compared with the normal group, the knee joint torque of 30-min FMJC increased significantly ( $P < 0.05$ ), and the joint motion tended to compensate in the transfer of mechanical load.

#### Hip Joint

Compared with the normal group, the maximum angle, minimum angle, and maximum torque of the hip joint at the 2 FMJC time points were significantly different ( $P < 0.01$ ,  $P < 0.5$ ). Compared with the normal group, the maximum torque of the hip joint in the 30-min FMJC group was significantly increased ( $P < 0.01$ ), and there was a tendency of joint movement in the transfer of mechanical load.

#### To Sum Up

In the case of FMJC, the angle of the ankle, knee, and hip joints changed significantly. The torque of the ankle, knee, and hip joints showed consistent changes, indicating that there is joint motion compensation between the hip, ankle, and knee. In the attenuation of torque compensation transfer, the torque changes of joints have a decreasing trend.

#### Discussion

When a person needs to lift the heel when walking, the metatarsophalangeal joint will inevitably produce movement, and



the metatarsophalangeal joint cannot be ignored in assessing the human body's movement. Under normal circumstances, the metatarsophalangeal joint has a large range of motion; however, in many cases, the range of motion of the metatarsophalangeal joint will be limited. Patients with these diseases often receive metatarsophalangeal joint fusion, especially the first metatarsophalangeal joint fusion. The operation can reduce or eliminate the pain, but often sacrifices the joint range of motion [23,24]. Prolonged limited movement of the metatarsophalangeal joints increases the risk of foot ulcers in patients with foot neurological diseases [1], which can even cause low back pain [25].

### Short-Term Limitation of Movement of the First Metatarsophalangeal Joint and Changes in Parameters of Hip and Knee Joints

When the movement of the human metatarsophalangeal joints is limited, the fulcrum moves forward when the foot is off the ground during walking, and the contact time between the forefoot and the ground is shortened, so that the support phase time of the walking cycle is reduced. There is a negative correlation between the support phase time and the pace. When there is a decrease in the percentage of the support phase, the subjects could reduce the step length or the step frequency, or adjust both at the same time to keep the pace constant. There was no significant difference in cadence between the 2 groups, which may be caused by different adjustment strategies adopted by different subjects. The angle changes of the hip and knee joints during the gait cycle are important factors affecting the time distance parameters [26]. The results of this study show that the limitation of movement of the first metatarsophalangeal joint has a significant impact on the angle of the hip and knee joints during walking, and there is a tendency for joint motion compensation to shift, which may also be why the step length and step frequency did not change significantly. We have reason to believe that the hip and knee joints play a role in compensating the ankle joint load, meaning that the lower-extremity exercise load is transferred from the proximal end to the distal end.

### Short-Term Limitation of Movement of the First Metatarsophalangeal Joint and Changes in Ankle Parameters

When a normal person walks, the first metatarsophalangeal joint has a larger range of motion. After the heel is off the ground, the metatarsophalangeal joint is bent, causing the sole to roll along the ground. The rolling axis is the metatarsophalangeal joint, which has a large dorsal curvature angle displacement before the toe is off the ground [27]. When the metatarsophalangeal joint is constrained, the heel is lifted and rotated forward with the toe as the center, and the radius of

gyration increases significantly, making the angle of ankle dorsiflexion increase significantly when off the ground ( $P < 0.05$ ). The angle of plantar flexion of the ankle joint at the initial stage of swing was significantly smaller than normal ( $P < 0.05$ ), because the angle of ankle dorsiflexion increases at the time of departure. When entering the swing phase, the ankle passively turns to plantar flexion under the action of gravity, and the degree of turning to plantar flexion of the ankle joint in a larger dorsiflexion state at the beginning of the swing decreases. When the heel touches the ground, the line of action of the ground reaction force passes behind the center of the ankle joint, and it generates an external torque to perform ankle joint plantar flexion. This external plantar flexion torque is opposed by an internal ankle joint dorsiflexion torque. This dorsiflexion torque (related to the eccentric activity of the ankle dorsiflexor) controls the front of the foot to touch the ground, and its value is small, mainly because the ground reaction force is small and the line of force is closer to the center of the ankle joint [28,29]. When the first metatarsophalangeal joint is restricted, the ankle joint is relatively dorsiflexed at the time of touchdown, and the ground reaction force on the ankle joint's torque arm is larger than normal. Therefore, the maximum ankle dorsiflexion torque is larger when restricted. Once the foot is flat, the center of pressure quickly moves forward below the metatarsal of the foot, and the line of ground reaction force passes in front of the center of the ankle joint, which generates an external dorsiflexion torque and quickly increases, balancing its internal plantar flexion torque (related to the eccentric activity of the plantar flexors of the ankle joint), which controls the forward movement of the lower leg over the foot. At this stage, although the ankle plantar flexion torque increases rapidly, the dorsiflexion rate is very slow, so the joint power is still small. The angle of dorsiflexion of the ankle joint reaches its maximum at 90% of the gait cycle, and plantar flexion begins. At this time, the speed of plantar flexion increases. The combination of internal plantar flexor torque and significant plantar flexion angular velocity forms an ankle joint power burst, which lifts the heel off the ground and moves ankle joint center upward and causes forward acceleration. This power is attributed to the centripetal movement of the plantar flexors of the ankle joint [30,31]. When the first metatarsophalangeal joint is constrained, after the heel is lifted, it rotates forward with the toe as the center, the pressure center moves forward, and the ground reaction force increases the torque arm of the ankle joint center, resulting in an increase in the external dorsiflexion torque.

This part of the dorsiflexion torque is resisted by the plantar flexion torque generated by the internal muscles. Therefore, the ankle joint plantar flexion torque, when pushing off the ground, is significantly greater than normal in the restricted situation ( $P < 0.05$ ).

## The Ankle Plantar Flexor Muscles Need to Consume More Energy When Walking in FMJC

The net joint torque is the equivalent of the muscle action effect [32]. The work done by the muscle torque will affect the mechanical energy of the system. When the joint net torque is consistent with the direction of the joint angular velocity, the joint power is positive, and the joint muscles contract centripetally to do positive work, which means that the muscles generate energy and transfer energy to the limbs. When the joint net torque is opposite to the angular velocity of the joint, the joint power is negative, and the eccentric contraction of the joint muscles does negative work. At this time, the work done by the external force on the muscles is the total work, which reflects the flow of energy from the limbs to the muscles, and the muscles absorb energy [33,34]. When walking, negative work mainly makes the limbs absorb energy when resisting gravity, while positive work makes the human body move forward. Whether it is positive work or negative work, muscle contraction consumes energy. When the first metatarsophalangeal joint is restricted, the negative work of the ankle joint was significantly greater than normal ( $P<0.05$ ), which indicates that the first metatarsophalangeal joint activity was restricted and the ankle flexor muscles needed to consume more energy when walking. In this study, the increase in energy consumption of the ankle joint caused by the FMJC was mainly due to the change of the forefoot support point and the center of pressure in the support phase, which makes the torque arm change, which affects the muscle torque and increases the mechanical work. The low-temperature thermoplastic sheet brace used in this research is light in weight, has ductility and stability, and can be made according to individual requirements. It is an ideal material for making lower-limb orthopedic braces [35], and the weight of each foot only increases by about 30 g. Our results also showed that the negative work of the ankle joint increased by 63.6% (from 0.11 J/kg to 0.18 J/kg) under restricted conditions, so the increase in the mass of the low-temperature thermoplastic sheet brace is not the main reason for the increase in energy consumption.

## When the Movement of the First Metatarsophalangeal Joint Is Limited, It Mainly Passes Through the Ankle, Knee, and Hip Joints for Sequential Motion Compensation Transmission

In this experiment, FMJC significantly affected the biomechanical indicators (eg, torque and power) of the ankle, knee, and hip joints during walking, while there was no significant change in the traditional kinematics data (eg, step length, step width, and pace). To ensure the integrity of movement, the adjacent and distal joints have compensatory effects [36,37]. If a joint has a problem, the other adjacent joints will use motion compensation to ensure the integrity of the movement; for example, in

the case of FMJC, the load will be compensated by the ankle, knee, and hip, in theory. The results of this experiment show that there is significant ankle, knee, and hip load compensation transmission, and the biomechanical changes of the ankle, knee, and hip are gradually attenuated.

Although short-term FMJC will only be shown in the data as the transfer of ankle, knee, and hip load, studies have shown that long-term knee load increase greatly increases the incidence of knee joints problems [38], seriously affecting the quality of life of patients, so the potential risk of FMJC is not only for the ankle joint, but also for the knee joint and even the hip joint. Many patients with knee arthritis have unsatisfactory treatment results because they do not pay attention to foot and ankle function. Studies have shown that foot and ankle function have an impact on knee joint disease [39]. However, the impact of FMJC has not been defined. The results of the present study prove the hypothesis that FMJ has a potential impact on the occurrence and development of knee arthritis.

## Shortcomings and Prospects

Of course, this experiment has some shortcomings and defects. Our sample size was small and there were certain differences among individual subjects. We will expand the number of subjects in the future. We only studied the short-term restriction of the first metatarsophalangeal joint, and did not conduct long-term follow-up studies. Because the subjects were healthy, long-term wearing of the device would cause irreversible damage to joints and muscles and other tissues, so the next step can be studying a patient with limited first metatarsophalangeal joint function, that is, a patient with limited forefoot movement, and rehabilitation treatment will be carried out according to their limitation. The treatment effect can be observed by comparing before and after treatment and comparing with the results of the present experiment. The transmission direction of compensation, the threshold of the transmission of compensation, and the criteria for judging the occurrence of compensation are all issues that we will address in future research. For patients with limited joint function, we believe that there are multiple compensatory transmissions at the same time, but the mutual influence and ultimate manifestation between them need further study. The possibility of lower-extremity compensation via the pelvic/sacroiliac joint to the other lower extremity/spine needs further research.

## Conclusions

After the movement of the first metatarsophalangeal joint is restricted, the human body mainly compensates by increasing the angle of dorsiflexion, increasing work, and increasing the activity level of surrounding muscles through the ankle

joint, so as to maintain the stride length, pace, and dynamic balance during exercise. Restricting the first metatarsophalangeal joint increases the energy consumption of the human ankle, knee, and hip joints during walking, and the load is compensated by the gradual attenuation of the ankle, knee, and hip. Long-term restriction can cause fatigue or damage to the muscles of the back of the calf, and also increase the incidence of knee arthritis.

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## Conflicts of Interest

None.