Gait & Posture 41 (2015) 522–528

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

The effect of angle and moment of the hip and knee joint on iliotibial band hardness

Hiroshige Tateuchi [a,*], Sakiko Shiratori [b], Noriaki Ichihashi [a]

[a] Human Health Sciences, Graduate School of Medicine, Kyoto University, Kyoto, Japan
[b] Division of Physical Therapy, Rehabilitation Units, Shiga University of Medical Science Hospital, Shiga, Japan

ARTICLE INFO

Article history:
Received 8 August 2014
Received in revised form 10 October 2014
Accepted 3 December 2014

Keywords:
Posture
Iliotibial band
Hip joint
External joint moment

ABSTRACT

Although several studies have described kinematic deviations such as excessive hip adduction in patients with iliotibial band (ITB) syndrome, the factors contributing to increased ITB hardness remains undetermined, owing to lack of direct in vivo measurement. The purpose of this study was to clarify the factors contributing to an increase in ITB hardness by comparing the ITB hardness between the conditions in which the angle, moment, and muscle activity of the hip and knee joint are changed. Sixteen healthy individuals performed the one-leg standing under five conditions in which the pelvic and trunk inclination were changed in the frontal plane. The shear elastic modulus in the ITB was measured as an indicator of the ITB hardness using shear wave elastography. The three-dimensional joint angle and external joint moment in the hip and knee joints, and muscle activities of the gluteus maximus, gluteus medius, tensor fasciae latae, and vastus lateralis, which anatomically connect to the ITB, were also measured. ITB hardness was significantly increased in the posture with pelvic and trunk inclination toward the contralateral side of the standing leg compared with that in all other conditions (increase of approximately 32% compared with that during normal one-leg standing). This posture increased both the hip adduction angle and external adduction moment at the hip and knee joint, although muscle activities were not increased. Our findings suggest that coexistence of an increased adduction moment at the hip and knee joints with an excessive hip adduction angle lead to an increase in ITB hardness.

1. Introduction

Iliotibial band (ITB) syndrome is an overuse injury associated with pain on the lateral aspect of the knee. The incidence of ITB syndrome in runners has been estimated to be between 5 and 14% [1]. The excessive compression between the ITB and lateral femoral epicondyle has been advocated as a cause of the ITB syndrome [2]. A recent review postulated that abnormal increase in the compression forces between the ITB and lateral femoral epicondyle causes irritation and inflammation in the tissue deep to the ITB [3]. Excessive hardness of the ITB can increase the compression force exerted by the ITB on the lateral femoral epicondyle. Thus, investigation of the factors increasing ITB hardness would shed light on the ITB syndrome.

Previous studies have investigated the biomechanics of the lower extremity during running by comparing healthy individuals and patients with ITB syndrome [4–11]; however, the research findings related to the kinematic and kinetic characteristics are controversial, probably because they have been reported in retrospective studies that include causal and compensational effects. In a prospective study, Noehren et al. [5] reported greater hip adduction and knee internal rotation throughout the stance phase of running in patients with ITB syndrome. They proposed that this motion increases ITB strain causing it to compress against the lateral femoral condyle. However, a weak relationship between the peak strain and strain rate, and the kinematic change of the lower extremity was reported [6]. These studies suggest that kinematic change is partly responsible for increasing strain in the ITB; however, the factors contributing to an increased ITB hardness are not well understood.

More importantly, direct in vivo ITB hardness measurements have not been reported to date. Therefore, such measurements are required to determine the factors affecting ITB hardness. Shear-wave elastography (SWE) is a reliable non-invasive ultrasonographic imaging technique for evaluating soft tissue properties by measuring the propagation velocity of the shear waves in tissues, allowing the shear elastic modulus to be calculated [12–19]. SWE has been used to measure the elastic properties of

* Corresponding author. Tel.: +81 75 751 3964; fax: +81 75 751 3964.
E-mail address: tateuchi.hiroshige.8x@kyoto-u.ac.jp (H. Tateuchi).

http://dx.doi.org/10.1016/j.gaitpost.2014.12.006
0966-6362/

the patellar and Achilles tendons, as well as the thigh and calf muscles [12,15,18]. Thus, a quantitative SWE could be useful in the measurement of ITB hardness.

The ITB is a lateral thickening of the fascia in the thigh, connecting the hip and knee muscles, i.e., the gluteus maximus (Gmax), gluteus medius (Gmed), tensor fasciae latae (TFL), and vastus lateralis (VL) [20–22]. Therefore, an excessive hip adduction, increase of external joint moment, and contraction of the above-described muscles could contribute to the increase in ITB hardness. The factors contributing to an increase in ITB hardness could be revealed by comparing the ITB hardness between the conditions in which the related factors are manipulated. Consequently, we aimed to clarify the factors contributing to an increase in ITB hardness using quantitative SWE. We hypothesized that ITB hardness would be increased by the increase in external load in addition to the increase in hip adduction angle. By examining the relationship between postural characteristics and ITB hardness as a basic research, we will gain insight into the potential cause of the ITB syndrome.

2. Methods

2.1. Participants

Sixteen healthy volunteers (8 men and 8 women; age, 21.9 ± 1.0 (mean $\pm$ SD) years; weight, 61.5 ± 10.8 kg; height, 169.3 ± 7.8 cm) participated in this study. Exclusion criteria included the presence of disease of any joint in the lower extremity/spinal joints and neurological disease. All subjects provided informed consent and the protocol was approved by the Ethics Committee of the Kyoto University Graduate School and Faculty of Medicine.

2.2. Experimental protocol

Prior to data collection for the one-leg standing, data were collected for 5 s in the bilateral standing position that was used as a reference for calculating the joint angle. Trials under five one-leg standing conditions on the dominant leg (defined as the leg the participant would use to kick a ball) were then conducted as follows (Fig. 1a): normal condition, normal one-leg standing with no pelvic and trunk inclination (NO); Pdrop, 10° drop of the pelvic contralateral side without trunk inclination; PTdrop, 10° drop of the pelvic contralateral side with trunk inclination toward the contralateral side; Prise, 10° rise of the pelvic contralateral side without trunk inclination; and PTrise, 10° rise of the pelvic contralateral side with trunk inclination toward the ipsilateral side. The angle change was checked by an examiner using goniometer. The contralateral hand was held at the abdomen. To maintain a stable posture, the participant was allowed to touch a fixed device with an index fingertip so that it was less likely to provide mechanical support [23]. The 5 conditions were measured in random order. ITB hardness was measured by SWE after holding each stable posture for 5 s [12]. The kinematic/kinetic variables and electromyographic (EMG) values were recorded synchronously for 3 s while maintaining each posture sequentially following the SWE measurement (Fig. 1b).

2.3. Shear-wave elastography

ITB hardness was measured at the level of the superior border of the patella using SWE (Aixplorer, SuperSonic Imagine, Aix-en-Provence, France) while maintaining each posture. ITB was identified by palpation and B-mode image, and the anterior and posterior borders of the ITB as well as the superior border of the patella were marked. The transducer was placed lightly with a generous amount of ultrasound gel for 5 s by a single investigator. In the present study, the transducer was placed transversely considering the following situation. The shear elastic modulus was consistently greater in the longitudinal than the transverse measurement [24]. In our preliminary experiment, we determined the shear elastic modulus in the ITB near the upper limit by longitudinal measurement. The reliability of the values obtained near the upper limit of the measurement device was reduced [15]. Furthermore, in the skeletal muscle, only slight increases in the shear elastic modulus with increased strain were found in transverse measurement compared to that in longitudinal measurement [17]. However, in the calcaneal tendon, elastic modulus was increased in response to tendon stretching regardless of whether the measurements were made longitudinally or transversely, and the reliability of values in the stretched positions was better in the transverse than the longitudinal measurement [15]. The inconsistency of these results may be attributed to the differences in histology and hardness of the tissue. The ITB is composed of dense fibrous connective tissue, and it is regarded as a tendinous tissue [25]. Therefore, we adopted transverse measurement to determine the ITB hardness.

Based on the thickness (approximately 1.9 mm) and width (approximately 5.3 mm) of the ITB determined using sonography [26,27], three regions of interest (ROI) with diameter of 1.5 mm for the measurement of Young's modulus were set horizontally to cover the entire region of the ITB and their mean was calculated (Fig. 1c). The observed values of Young's modulus were divided by 3 to obtain the shear elastic modulus. All the measurements were performed twice and the mean shear elastic modulus values for 2 trials were used in the analysis. The determination of the ROI and calculation of the shear elastic modulus were performed by 1 examiner who was blind to the experimental conditions.

2.4. Motion capture

The kinematic and kinetic measurements were recorded using a 7-camera Vicon motion system (Vicon Nexus; Vicon Motion Systems Ltd. Oxford, England) at a sampling rate of 100 Hz and a fourth order Butterworth low-pass filter with a 6 Hz cutoff, and force plates (Kistler Japan Co., Ltd. Tokyo, Japan) at a sampling rate of 1000 Hz and a low-pass filter (20 Hz), respectively. Reflective markers were placed by one examiner. A total of 27 markers were placed bilaterally on the acromioclavicular joint, anterior superior iliac spine, posterior superior iliac spine, superior aspect of the greater trochanter, lateral femoral condyle, medial femoral condyle, lateral malleoli, medial malleoli, heel, fifth metatarsal head, and first metatarsal head. Additionally, markers were placed on the jugular notch, xiphoid process, seventh cervical spinous process, tenth thoracic spinous process, and right scapula. The thoracic segment had six markers at the jugular notch, xiphoid process, seventh cervical spinous process, tenth thoracic spinous process, and bilateral acromioclavicular joints. The pelvic segment contained 4 markers at the bilateral anterior superior iliac spine and posterior superior iliac spine. The thigh segment had three markers at the superior aspect of the greater trochanter and at the medial and lateral femoral condyles. The shank segment had four markers at the medial and lateral femoral condyles and at the medial and lateral malleoli. According to a previous study [28], we calculated the three-dimensional joint angle and external joint moment of the hip and knee joints using the BodyBuilder software (Vicon Motion Systems Ltd. Oxford, England). In the analysis, segments were regarded as rigid and the joint moment was calculated using a link segment model, in which segments were connected together at nodal points. To compute the joint moment, coordinate data were added to the GRF data, in which the position of the center of mass, the weight portion, and the moment of inertia of each segment were used as parameters. The joint