The effect of three-dimensional postural change on shear elastic modulus of the iliotibial band

1. Introduction

Iliotibial band (ITB) syndrome, which causes pain in the lateral aspect of the knee, occurs in 5-14% of runners [van der Worp et al., 2012], and frequently occurs in patients with knee osteoarthritis [Vasilevska et al., 2009]. The excessive compression between the ITB and the lateral femoral epicondyle has been proposed as a cause of ITB syndrome [Fairclough et al., 2007]. Excessive stiffness of the ITB can increase the compression force exerted by the ITB on the lateral femoral epicondyle; therefore, investigation of the factors that increase ITB stiffness can improve our understanding of ITB syndrome.

The ITB is a lateral thickening of the fascia in the thigh, connecting the hip and knee muscles (i.e., the gluteus maximus, gluteus medius, tensor fasciae latae [TFL], and vastus lateralis), lateral intermuscular septum, lateral femoral epicondyle, patella, and tibia [Terry et al., 1986; Vieira et al., 2007; Becker et al., 2010]. Consequently, ITB stiffness could be influenced by factors such as joint angles, external joint moments, and corresponding muscle tension during movement.

Several studies have found kinematic changes at the hip and knee joints in patients with ITB syndrome, which may contribute to this pathology, though a specific cause-and-effect relationship is not well-defined [Miller et al., 2007; Noehrren et al., 2007; Hamill et al., 2008; Feber et al., 2010; Foch and Milner, 2014; Louw and Deary, 2014]. A prospective study found greater hip adduction and knee internal rotation throughout the stance phase of running in patients with ITB syndrome [Noehren et al., 2007]. These authors presumed that excessive hip adduction and knee internal rotation increases ITB strain, causing the ITB to compress against the lateral femoral condyle. Research using a musculoskeletal model has reported that increasing step width during running reduces ITB strain and the strain rates mediated by decreased hip adduction [Meardon et al., 2012]. These findings strongly suggest that kinematic changes, especially in hip motion in the frontal plane, significantly influence ITB stiffness.

Furthermore, in planes other than the frontal plane, Noehren et al. [2007] found that individuals who develop ITB syndrome exhibited greater external rotation of the femur. In addition, it was reported that female runners with ITB syndrome exhibited greater hip external rotation while running than male runners with ITB syndrome and healthy female runners [Phinyomark et al., 2015]. In a cadaver-based study, strain in the ITB was greater during hip flexion and adduction stretching, which increase the tensioning role of the gluteus maximus more than a straight leg raise; however, no difference was found in ITB strain between the hip flexion and adduction stretching and the modified Ober test, which includes hip extension and adduction [Falvey et al., 2010]. These findings demonstrate the importance of focusing on hip joint motion changes in the sagittal and transverse planes in addition to the frontal plane, as these factors may contribute to increased ITB stiffness. However, the anatomical complexity of the ITB makes it difficult to estimate the ITB stiffness by observing joint motion. The factors contributing to an increase in ITB stiffness would be revealed by comparing in vivo measurements of ITB stiffness following posture changes.

Shear-wave elastography (SWE) is a reliable, non-invasive ultrasonographic imaging technique for evaluating elasticity of soft tissue by measuring the propagation velocity of the shear waves in the tissues, enabling the calculation of the shear elastic modulus [Kot et al., 2012; Maïsetti et al., 2012; Aubry et al., 2013; Eby et al., 2013]. SWE has been used for measuring stiffness of the soft tissue such as muscle and tendon [Aubry et al., 2013, 2015; Brandenburg et al., 2014]. Recently, we have demonstrated that, compared to normal one-leg standing, an increased hip adduction angle and adduction moment at the hip and knee joints leads to an approximately 32% increase in ITB stiffness, measured in vivo using SWE [Tateuchi et al., 2015]. However, the in vivo measurement of ITB stiffness has only been performed in our two-dimensional investigation [Tateuchi et al., 2015]; the effect of postural changes on ITB stiffness is poorly understood. Therefore, the purpose of this study was to clarify the factors contributing to an increase in ITB stiffness using quantitative SWE by examining the relationship between three-dimensional postural changes and ITB stiffness. Based on the previous related research [Noehren et al., 2007; Falvey et al., 2010; Phinyomark et al., 2015], we hypothesized that ITB stiffness would be increased in the (1) hip adducted position in the frontal plane, (2) hip externally rotated position in the transverse plane, and (3) hip flexed position in the sagittal plane compared to the normal one-leg standing position.

2. Methods

2.1. Participants

Fourteen healthy volunteers (7 men and 7 women; age, 22.0 ± 1.0 (mean ± SD) years; weight, 61.3 ± 11.5 kg; height, 168.6 ± 7.9 cm) were recruited from the local student population. The exclusion criteria included the presence of disease of any joint in the lower extremity or spine, neurological disease, or a history of ITB syndrome. All subjects provided informed consent and the study received ethical approval from the local ethical committee.

2.2. Experimental protocol

Prior to the one-leg standing condition, kinematic data were collected for 5 s in the bilateral standing position, which were used as a reference for calculating the joint angles. We then manipulated the pelvic rotational position while the participant maintained a stable one-leg standing position, to vary the three-dimensional hip joint angles. The pelvic rotational position is a point of observation during the retraining of running gait in patients with ITB syndrome [Hunter et al., 2014], as it is directly related to hip joint motion. One-leg standing was performed on the dominant leg, defined as the leg that the participant would use to kick a ball. Trials under the 7 one-leg standing conditions were conducted as follows (Fig. 1): normal condition (NO), normal one-leg standing with no pelvic and trunk inclination; PT, 10° posterior tilt of the pelvis (i.e. hip extension); AT, 10° anterior tilt of the pelvis (i.e. hip flexion); CD, 10° drop of the pelvic contralateral side (i.e. hip adduction); CR, 10° rise of the pelvic contralateral side (i.e. hip abduction); PR, 5° posterior rotation of the pelvic contralateral side (i.e. hip external rotation); and AR, 10° anterior rotation of the pelvic contralateral side (i.e. hip internal rotation). The pelvic rotation angle was set at 5° in the PR condition because a stable posture could not be maintained at 10°. In the PT, AT, CD, and CR conditions, inclination of the trunk in the same direction as pelvic rotation was allowed to maintain balance. The pelvic angle change was verified by an examiner using a goniometer before each data collection. 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, which minimized mechanical support. The order of the 7 conditions was randomized. ITB stiffness was measured using SWE after holding each stable posture for 5 s, based on a previous study [Kot et al., 2012]. The kinematic and kinetic variables were recorded synchronously for 3 s while maintaining each posture, following the SWE measurement.

2.3. Shear-wave elastography

ITB stiffness was measured at the level of the superior border of the patella using SWE (Aixplorer, SuperSonic Imagine, Aix-en-Provence, France) while the participant maintained each posture. Prior to the measurements, the investigator identified the ITB at the level of the superior border of the patella by palpation and examination of the B-mode image, which is two dimensional brightness mode ultrasonography, and marked the measurement site and the anterior and posterior borders of the ITB with a pen (Fig. 2a). The transducer was placed lightly on the participant using a generous amount of ultrasound gel. A single investigator performed all SWE measurements. In the present study, the transducer was placed transversely because the shear elastic modulus of the calcaneal tendon was increased in response to tendon stretching regardless of whether the measurements are made longitudinally or transversely, and the reproducibility of the longitudinal measurement tended to be decreased in a tendon stretched position because it reaches the upper limit of the measurement device [Aubry et al., 2013, 2015]. Therefore, we utilized transverse measurements of the ITB, which is regarded as a tendinous tissue [Fairclough et al., 2006].

Based on the thickness (approximately 1.9 mm) and width (approximately 5.3 mm) of the ITB determined using ultrasonography [Goh et al., 2003; Wang et al., 2008], regions of interest (ROIs) with a diameter of 1.5 mm were set in the ITB. Although elastic modulus value is not influenced by the ROI's size when the mean value in the ROI is used [Kot et al., 2012], the ROI's size in our study was considerably smaller, approximately half the size of that in the previous study in the calcaneal tendon [Aubry et al., 2013]. Therefore, the 3 ROIs were set horizontally to cover the entire region of the ITB although only one ROI was commonly used in the previous study using SWE, while the mean value of the 3 ROIs was used to determine a more representative measure (Fig. 2b). Based on the local shear wave propagation velocity, c, the Young's modulus, E, was calculated from $E = 3 \rho c^2$ where density ρ is assumed to be constant (1000 kg/m^3) [Aubry et al., 2013]. The observed values of Young's modulus were divided by 3 to obtain the shear elastic modulus. All 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 one examiner who was blinded to the experimental conditions. Excellent intra-rater reliability of the SWE measurement of the ITB during one-leg standing was confirmed in an earlier investigation (ICC [1,2]: 0.94) [Tateuchi et al., 2015].

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. The reflective markers were placed by one examiner. A total of 20 markers were placed bilaterally on the 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. The pelvic segment contained 4 markers, at the bilateral anterior superior iliac spine and posterior superior iliac spine. The thigh segment had 3 markers, at the superior aspect of the greater trochanter and the medial and lateral femoral condyles. The shank segment had 4 markers, at the medial and lateral femoral condyles and the medial and lateral malleoli. In accordance with a previous study [Tateuchi et al., 2015], we calculated the three-dimensional joint angles and external joint moments of the hip and knee joints using BodyBuilder software (Vicon Motion Systems Ltd. Oxford, England). The joint angles during each condition were calculated in reference to the joint angles determined from that obtained during the bilateral standing position. The external joint moments were normalized for body weight and height.

2.5. Statistical analysis

For each parameter, a paired *t* test with a Shaffer correction for multiple comparisons was used to determine the differences between the NO and the other 6 conditions. A p value <0.05 was considered statistically significant. SPSS version 19.0 (IBM Japan Ltd.) was used for statistical analysis.

3. Results

Confirming the prescribed postural changes, compared to the NO, the hip extension angle increased in the PT and decreased in the AT conditions; the hip adduction angle increased in the CD and decreased in the CR conditions; and the hip external rotation angle increased in the PR and decreased in the AR conditions. All differences were statistically significant. In addition, the hip external rotation angle also increased in the CR condition (Fig. 3). These results confirm that changes in the hip joint angles occurred.

ITB stiffness was significantly increased in the PT, CD, and PR conditions compared to that in the NO. Conversely, ITB stiffness was significantly decreased in the AT and CR conditions compared to that in the NO (Fig. 4). The percentage changes of the ITB stiffness in the PT, AT, CD, CR, PR, and AR conditions to the NO condition were 29.6%, –29.2%, 33.5%, –25.1%, 28.0%, and –2.4%, respectively.

For the other kinematic and kinetic parameters accompanying the postural changes, there were multiple significant changes, compared to the NO, in knee angle and hip and knee joint moments (Fig. 3). The knee flexion angle was significantly changed in the PT and AT conditions, and the knee rotation angle was significantly changed only in the CR condition. In the kinetic analysis, the hip extension moment was significantly increased in the PT and decreased in the AT conditions, compared to that in the NO. The hip extension moment was slightly changed even in the PR and AR conditions. For the hip adduction moment, there was clearly increase in the CD and decrease in the CR conditions, and in addition, there were small changes in the PT and AT conditions. The hip external rotation moment was significantly increased in the CD and decreased in the CR conditions, whereas there was no significant difference in the hip rotation moment in the PR and AR conditions. The knee extension moment was significantly decreased in the PT, CD, and PR conditions. The knee adduction moment was significantly changed in the PT, CD, CR, and PR conditions; however, an increase in the knee adduction moment was found only in the CD condition. The knee internal rotation moment was significantly decreased in the PR condition. A summary of the significant changes in the joint angles and joint moments at the hip and knee joints in each condition are shown in Table 1.

4. Discussion

Our study found that ITB stiffness was increased in the PT, CD, and PR conditions, in which the hip joint was extended, adducted, and externally rotated, respectively. Therefore, our hypothesis was supported in that ITB stiffness was increased when the hip was adducted and externally rotated. However, contrary to our hypothesis, ITB stiffness was increased when the hip was extended.

Because the ITB receives fibers from the TFL and gluteal muscles and runs caudally outside the hip and knee joint, it is reasonable that ITB stiffness was increased by elongation in the hip adducted position (CD condition), which could produce a passive elastic moment against the increased hip and knee adduction moment [Yoon and Mansour, 1982]. Furthermore, we found an increase in ITB stiffness in the hip externally rotated position (PR condition).

Previous studies have reported that increases in external rotation of the femur or the hip joint during running represent pathokinematics of ITB syndrome [Noehren et al., 2007; Phinyomark et al., 2015]. Our result provides a biomechanical basis for their argument that patients with ITB syndrome should focus on decreasing hip external rotation. Of the tissues that connect to the ITB, it appears that the tension in the TFL was increased the most following external rotation of the hip. The TFL inserts directly into the ITB, and the ITB behaves as an elongated tendon insertion of the TFL [Falvey et al., 2010]; thus, tensioning effect of the TFL would lead to an increase in ITB stiffness. In addition, the hip extension moment was inevitably increased in the PR condition by the backward displacement of the contralateral side of the body. The TFL can produce a braking force against the hip extension moment; therefore, it may have partly contributed to the increase in ITB stiffness.

We hypothesized that ITB stiffness would be increased in the hip flexed position. However, we found that while ITB stiffness was increased in the hip extended position (PT condition), it was decreased in the hip flexed position (AT condition). A previous study reported that ITB strain was greater in the hip flexion and adduction during non-weight bearing stretching, pointing to the tensioning effect of the gluteus maximus on ITB stiffness [Falvey et al., 2010]. In contrast, our result indicates that ITB stiffness was increased in the PT condition, which could load the TFL by increasing the hip extension angle and moment, rather than the AT condition, which could load the gluteus maximus by increasing the hip flexion angle and moment. Therefore, as with the PR condition, the TFL would have a greater impact on ITB stiffness. The difference between weight bearing and non-weight bearing conditions, along with the difference between cadavers and living individuals, could explain the inconsistent findings.

In our experiment, during one-leg standing, the hip abductors need to exert some force to maintain frontal-plane balance. Added to this, the hip angle and moment in the sagittal plane was changed in the PT and AT conditions. In a related study that used the same method as the current study [Tateuchi et al., 2015], the muscle activity of the gluteus maximus was relatively low (2.3% of maximum voluntary contraction) compared to that of the gluteus medius (10.3%) and TFL (7.3%) during one-leg standing. The TFL has a relatively high level of tension during one-leg standing, thus applying an additional load to the TFL by hyperextending the hip would result in increase of ITB stiffness.

Posture and gait re-training has been included as a treatment intervention for the patients with ITB syndrome [Fredericson and Weir, 2006; Meardon et al., 2012; Allen, 2014; Hunter et al., 2014]. Various points of modification have been recommended, including widening the step width, altering the pelvic external rotation, running at a fast pace that increases knee flexion, and increasing the cadence to reduce the work at the knee [Fredericson and Weir, 2006; Meardon et al., 2012; Allen, 2014; Hunter et al., 2014]. Relief of lateral knee pain was also reported following treatment that emphasizes gait re-training [Allen, 2014; Hunter et al., 2014]. To complement these findings, our results suggest that re-training of posture and gait should include reducing the pelvic posterior tilt, contralateral drop, and contralateral backward rotation, and from the standpoint of hip kinematics, reducing excessive hip extension, adduction, and external rotation, as needed. Future research is needed to establish effective ways to retrain gait to treat ITB syndrome, considering individual differences.

This study has some limitations. First, the angle changes at the hip joint accomplished by pelvic rotation were only approximately 13° in each plane. More dynamic movements such as running require a greater range of hip joint motion, especially in the sagittal plane [Novacheck, 1998]; thus, the tensioning effect of the gluteus maximus may affect ITB stiffness in a position with greater hip flexion. However, given that hip flexion led to a decrease in ITB stiffness under our experimental conditions, possibly due to the decreased tensioning of the TFL, tensioning of the gluteus maximus may have a limited effect on ITB stiffness even in a position of greater hip flexion. Despite this, our findings are relevant to the relationship between three-dimensional postural changes and ITB stiffness, because the one-leg standing posture might correspond to the posture during the midsupport phase of running, which is altered in patients with ITB syndrome [Noehren et al., 2007; Hamill et al., 2008]. Second, kinematic change in the knee joint, which was suggested to have an effect on ITB stiffness [Miller et al., 2007; Noehren et al., 2007; Feber et al., 2010; Louw and Deary, 2014], was not evaluated, although supplemental changes of the knee joint were observed with postural changes. The effect of knee kinematic changes (e.g., knee internal rotation and flexion angle) on ITB stiffness requires further examination. Finally, this study was conducted in a sample of healthy subjects. Therefore, the results can be applied to this cohort only. Further studies will be necessary to examine the effect of hip and knee positional changes on ITB stiffness in patients with ITB syndrome.

In conclusion, ITB stiffness was increased not only in the hip adducted position but also in the hip extended and externally rotated positions. This was accompanied by increased moments of hip adduction and extension. In contrast, a significant decrease in ITB stiffness was found in the hip flexed and abducted positions. These findings suggest that increases in ITB stiffness could be attributed to the tensioning effect of the TFL. Further investigations of evaluation and treatment in patients with ITB syndrome would need to include ITB stiffness along with three-dimensional hip positional changes.

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Table 1

Summary of the changes of the joint angles and moments in each condition relative to the NO

	PT	AT	CD	CR	PR	AR
Joint angle changes						
Hip	↑ Ext	↑ Flex	↑ Add	↑ Abd, ER	↑ ER	↑ IR
Knee	↑ Flex	↑ Ext		↑ IR		
External joint moment changes						
Hip	↑ Ext, Add	↑ Flex, ↓ Add	↑ Add, ER	↓ Add, ER	↑ Ext	↑ Flex
Knee	↓ Ext, Add	↑ Ext	↓ Ext, ↑ Add	↓ Add	↓ Ext, Add, IR	

Arrow represents significant increase/decrease compared with NO condition.

Ext, extension; Flex, flexion; Add, adduction; Abd, abduction; ER, external rotation; IR, internal rotation.

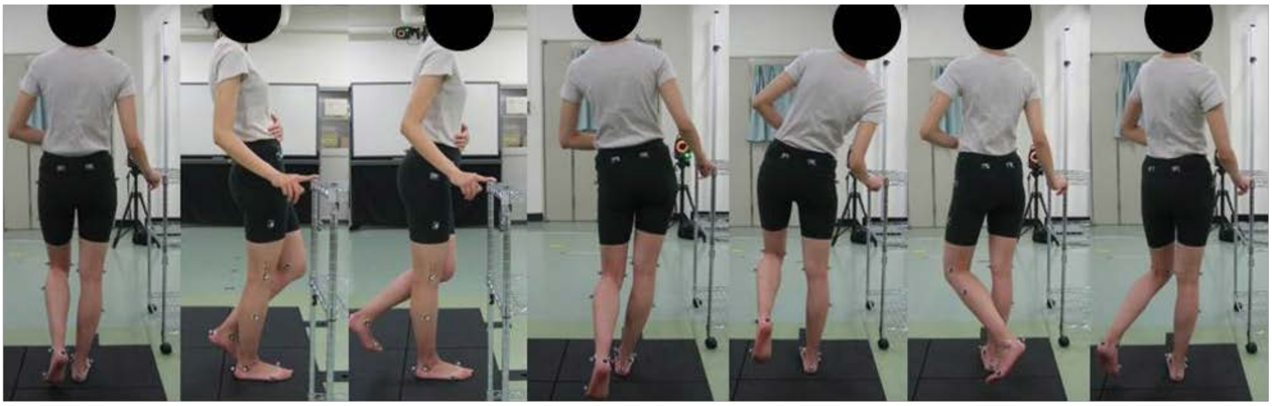
Figure captions

Fig. 1. Experimental conditions in the one-leg standing with applied marker set. NO, normal one-leg standing condition with no pelvic and trunk inclination; PT, 10° posterior tilt of the pelvis (i.e. hip extension); AT, 10° anterior tilt of the pelvis (i.e. hip flexion); CD, 10° drop of the pelvic contralateral side (i.e. hip adduction); CR, 10° rise of the pelvic contralateral side (i.e. hip abduction); PR, 5° posterior rotation of the pelvic contralateral side (i.e. hip external rotation); and AR, 10° anterior rotation of the pelvic contralateral side (i.e. hip internal rotation).

Fig. 2. (a) ITB stiffness was measured at the superior border of the patella (arrow). The measurement site and anterior and posterior borders of the ITB were marked. (b) A typical example of the color-coded box superimposed on the gray scale image. Three regions of interest (Q-box: 1.5 mm diameter) were arranged horizontally in the ITB.

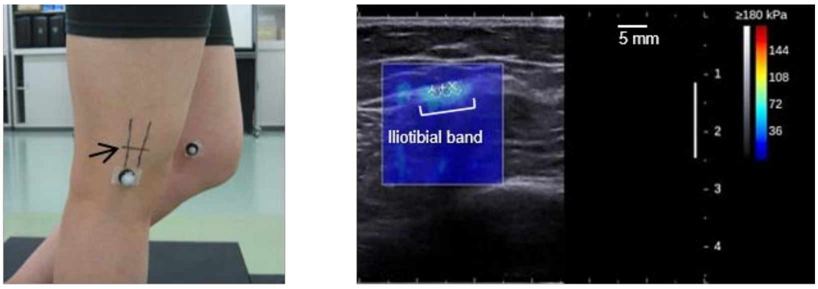
Fig. 3. Joint angles of the hip (a) and knee (b) joints and external joint moments of the hip (c) and knee (d) joints in each condition. Data are mean (SD). Dagger on the graph represents a statistically significant difference from NO ($p < 0.05$).

Fig. 4.
The shear elastic modulus of the iliotibial band in each condition. Data are mean (SD). Dagger on a graph represents a statistically significant difference from NO ($p < 0.05$).



NO PT AT CD CR PR AR

Fig 1.



(a) (b)

Fig 2.

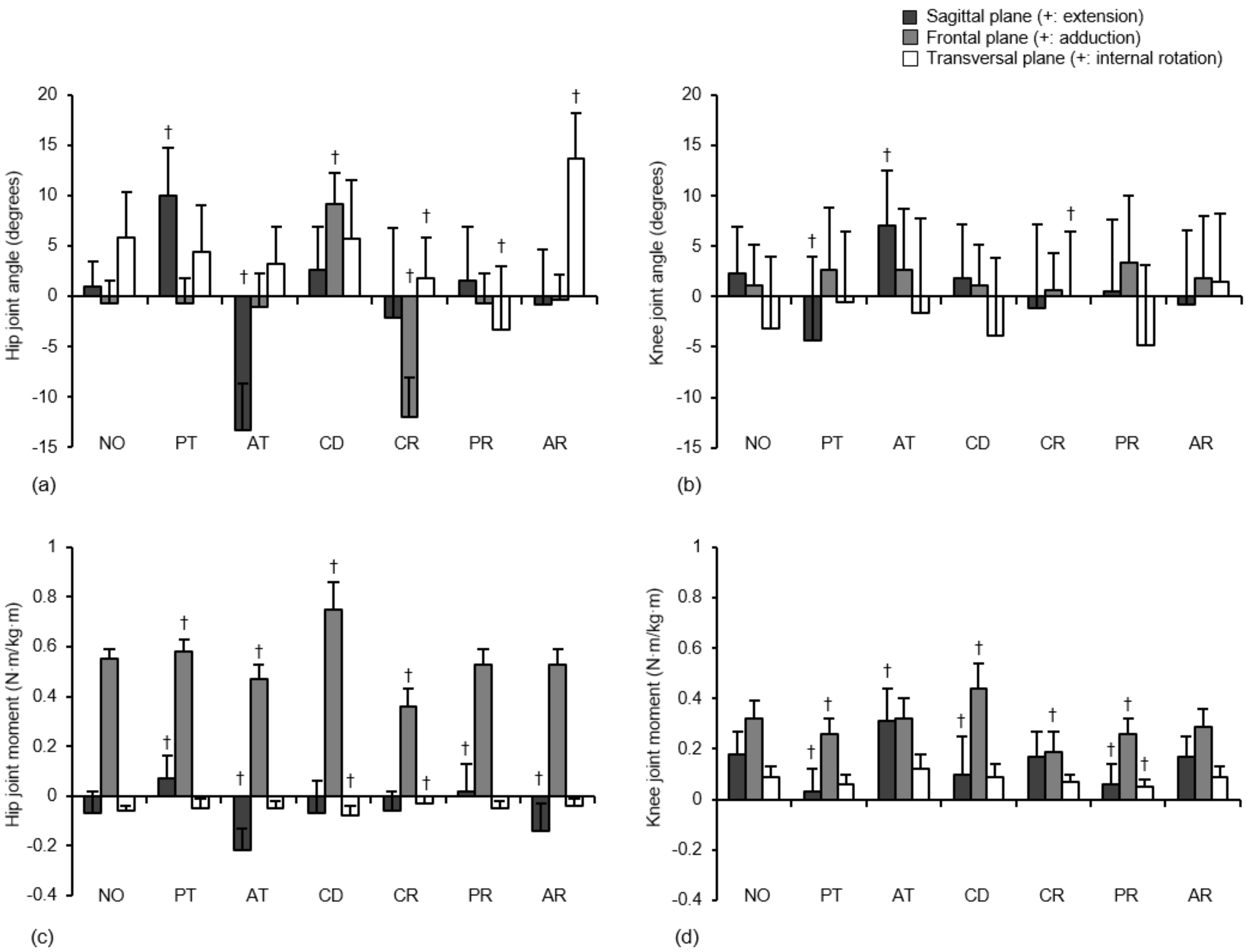

Sagittal plane (+: extension)
Frontal plane (+: adduction)
Transversal plane (+: internal rotation)
Hip joint angle (degrees)
Knee joint angle (degrees)
Hip joint moment (N·m/kg·m)
Knee joint moment (N·m/kg·m)
NO
PT
AT
CD
CR
PR
AR
(a)
(b)
(c)
(d)

Fig 3.

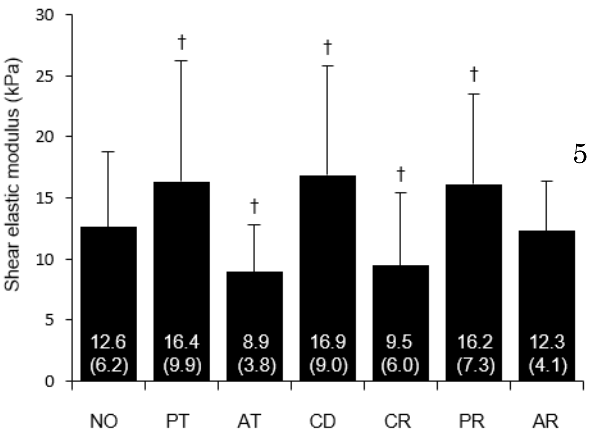

Shear elastic modulus (kPa)
12.6 (6.2)
16.4 (9.9)
8.9 (3.8)
16.9 (9.0)
9.5 (6.0)
16.2 (7.3)
12.3 (4.1)
NO
PT
AT
CD
CR
PR
AR

Fig 4.