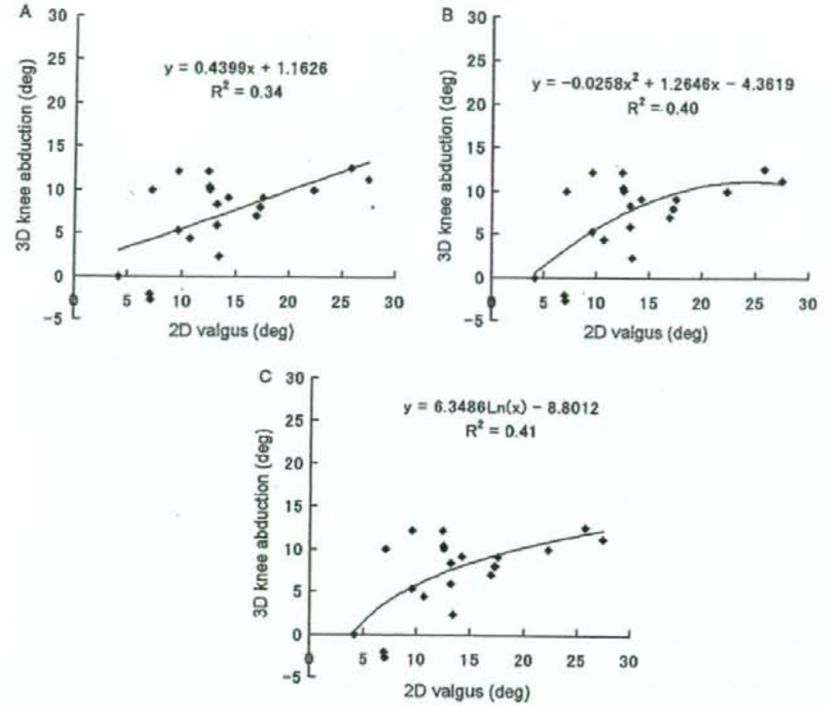

Figure 3. Associations between 2D valgus and 3D knee abduction during the continuous jump test for the linear model (A), the quadratic model (B), and the logarithmic model (C). The R^2 values of all models between 2D valgus and 3D knee abduction were significantly different from zero.

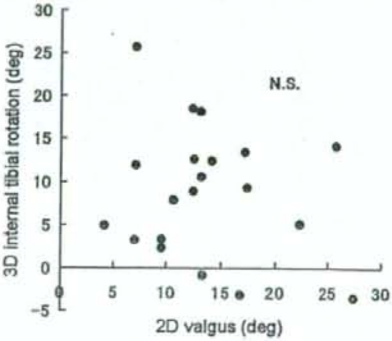

Figure 4. Associations between 2D valgus and 3D internal tibial rotation during the continuous jump test. The R^2 values of all models between 2D valgus and 3D internal tibial rotation were not significantly different from zero.

According to a prospective study (Hewett et al., 2005), individuals who went on to have an ACL injury had a 7.6° greater knee abduction angle at landing than those who did not get injured. Although a different movement task was studied, 9° of injured knee-abduction angle (Hewett et al., 2005) could be used to determine the accuracy of the regression models. By doing so, the false-negative rate was 35%, 30%, and 30% for the linear, quadratic, and logarithmic model, respectively. The false-positive rate was 0%, 10%, and 10% for the linear, quadratic, and logarithmic model, respectively. Thus, these regression models could be used as one screening tool to assess the risk of ACL injury during landing. However, the false-negative rates were slightly high. Therefore, other screening tools [i.e. lower limb strength (Barber-Westin et al., 2006) and joint laxity (Myer et al., 2008)] should be used in addition to these regression models to introduce athletes who are at risk for ACL injury to prevention training.

Since there was no significant difference between the three models based on the R^2 values, any of the models can be used to evaluate knee valgus. However, the data points scatter around the regression curves, especially between 10° and 15° of 2D valgus. In this area, other factors contributing to the 2D valgus angle (i.e. hip rotation, ankle eversion/inversion, stance width, etc.) vary between individuals in a way that is not correlated with knee valgus. Although this occurrence is a fundamental limitation of the regression model approach, the non-linear regression models take into account the scatter in this area better than the linear model. Therefore, we suggest that the logarithmic regression model, which has a damping behaviour, has most suitable to evaluate knee valgus. On the other hand, data points above 15° of 2D valgus fit well with the regression curves. In this area, the regression model can be used to screen individuals at risk for ACL injury, since the knee abduction angle is relatively large.

In this study, participants who showed knee varus during landing were excluded. In theory, the correlation should hold whether valgus or varus was measured. However, there was no significant correlation when those who showed varus landing were included. The 2D measurement of varus/valgus angle is affected by many factors including hip rotation, ankle eversion/inversion, foot position, and knee flexion. This result shows that the contributions of these factors may be different between 2D valgus and 2D varus. Since most female athletes show valgus landing and valgus landing is related to risk of ACL injury, we decided in this study to screen for valgus only.

A significant regression relationship between 2D valgus and 3D internal tibial rotation could not be determined. Increased internal tibial rotation combined with knee valgus leads to increased strain (Berns et al., 1992) and increased force (Markolf et al., 1995; Kanamori et al., 2002) in the ACL. Therefore, evaluation of internal tibial rotation during landing has benefits for screening and identifying risk of ACL injury. However, from the results of this study, using frontal plane 2D analysis of valgus motion to evaluate internal tibial rotation is not advised. It may be necessary for other parameters (e.g. foot position) to be examined or 3D analysis should be used to measure tibial rotation.

There are some limitations to this study. First, it is unclear whether the statistical regression model in this study could be applied to other athletic populations or to male athletes. Furthermore, since the participants in this study were barefoot, the effect of wearing shoes could have an influence on the results. There is also a fundamental limitation of the regression model approach. As mentioned earlier, hip internal rotation and other variables are correlated to 2D knee valgus; however, at times these factors vary among individuals in a way that does not correlate with 2D knee valgus and thus contributes to scatter within the data. Lastly, the power of this test to make comparisons among regression models was low (0.18). To examine which statistical model best describes the 3D knee kinematics, a larger sample size is needed.

Conclusion

We examined the reliability of a 2D approach to screen individuals for risk of ACL injury during a jump landing task. The results suggest that not only the linear model, but the quadratic and logarithmic models, show a moderate association between the 2D and 3D methods to measure knee abduction. The 2D approach could be used to screen a specific group of individuals who have greater 2D valgus and 3D knee abduction. However, the use of frontal plane 2D analysis of valgus motion to evaluate internal tibial rotation is not advised.

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Addition of an arch support improves the biomechanical effect of a laterally wedged insole

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ABSTRACT

In order to examine if the addition of an arch support could improve the biomechanical effect of the laterally wedged insole, three-dimensional gait analysis was performed on 20 healthy volunteers. Kinetic and kinematic parameters at the knee and subtalar joints were compared among the following four types of insoles; a 5-mm thick flat insole, a flat insole with an arch support (AS), a 6° inclined laterally wedged insole (LW), and a laterally wedged insole with an arch support (LWAS). The knee adduction moment averaged for the entire stance phase was reduced by the use of LW and LWAS by 7.7% and 13.3%, respectively, from that with FLAT. The difference in knee adduction moment between LW and LWAS was most obvious in the late stance, which was ascribed to the difference in the progression angle between those insoles. The analyses also revealed that LW tended to increase step width, and that such an increase was completely eliminated by the addition of an arch support to LW. This reduction of step width could be another mechanism for the further reduction of the moment with LWAS. The analyses of biomechanical parameters at the subtalar joints suggested that LWAS allowed the subject to walk in a more natural manner, while exerting greater biomechanical effects than LW. Thus, the addition of an arch support to the laterally wedged insole reduced knee adduction moment more efficiently, possibly through the elimination of potential negative effects of the laterally wedged insole.

1. Introduction

Osteoarthritis (OA) of the knee joint is the most prevalent joint disease among the elderly. Loading of the knee has been shown to play a key role in the development and progression of the disease [1]. Loading while walking is particularly important, because walking is the most frequently performed activity. During walking, the load is not equally distributed between the medial and lateral compartments of the joint. In a normal gait, the peak force on the medial compartment is almost 2.5 times that on the lateral compartment [2]. This uneven loading may account for the high susceptibility of the medial compartment to OA. Once OA changes are initiated, the magnitude of the medial load is associated with the severity of symptoms and progression of the disease [1,3]. Therefore, load reduction within the medial compartment could be critical in the management of patients with medial knee OA.

The load transferred through the medial and lateral compartments during walking can be estimated on the basis of the external knee adduction moment measured during three-dimensional gait analysis [2]. Using this parameter, both the symptoms and progression of medial knee OA were shown to correlate with the magnitude of load transferred through the medial compartment [1,3]. Knee adduction moment has also been used to evaluate the effects of treatments directed to reduce the medial load [4–8].

Laterally wedged insoles are used to treat patients with medial knee OA in its earlier stages [9,10], and successful results have been reported [11,12]. However, its efficacy may be limited [9,13–15], possibly because the insole fails to reduce knee adduction moment in certain individuals [7,16]. Furthermore, its effectiveness may be reduced by the discomfort caused by its use [17]. In an attempt to relieve that discomfort, we modified the insole by adding an arch

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support. Unexpectedly, the use of the modified insole not only reduced discomfort but also enhanced the clinical results [18]. This result led us to consider whether the addition of an arch support may improve the biomechanical effect of the laterally wedged insole. The present study was conducted to examine this hypothesis.

2. Methods

2.1. Subjects

This study was performed on healthy volunteers under the approval of the institutional review boards. Sample size was determined by a published nomogram [19], based upon our previous data [7,20,21]. The result indicated that a sample size of 20 would be enough to detect a 5% difference in peak knee adduction moment or peak subtalar abduction moment, with a statistical power of 80% and a 5% level of significance. Thus, 20 healthy volunteers (11 males and 9 females) who had no known history of symptoms with their back and lower extremities were enrolled in this study. Informed consent was obtained in writing from all volunteers. Prior to gait analysis, lower limbs were clinically examined and weight-bearing antero-posterior radiographs were obtained to confirm that there was no abnormality. Details of the subjects are shown in the supplementary data (Supplementary Table 1).

2.2. Data acquisition system

Three-dimensional gait analysis was conducted with a 12-camera optoelectronic motion analysis system (Vicon 512; Oxford Metrics, Oxford, UK) combined with eight force platforms (Kistler 9281C; Kistler Instrument, Winterthur, Switzerland) as described previously [7,20,21]. Details of the data acquisition system are given in the supplementary data (Supplementary Method).

2.3. Experimental protocol

Four types of insoles were tested in this study (Fig. 1). A laterally wedged insole (LW) had the size of the entire sole and was inclined medially at an angle of 6° along the full length of the insole. A laterally wedged insole with an arch support (LWAS) was used to evaluate the effect of that added arch support. A 5-mm thick, flat insole without inclination (FLAT) and a flat insole with an arch support (AS) were used as the controls. All insoles and arch supports were made of ethylene vinyl acetate (EVA 8200, Toyo Sponge, Tokyo, Japan), which had an elasticity coefficient of 100–300 kg/mm^2. The insole size and the shape and height of the arch support were adjusted to fit each subject. The insoles were directly attached to the subjects' soles bilaterally with double-sided adhesive tape, and the subjects were requested to walk on the walkway with the insoles, without wearing shoes. The insoles were tested at a self-selected, natural walking speed. To keep the gait velocity constant during measurement, a metronome was first set to the subject's cadence, and the subject was requested to walk with the insoles at that cadence. The four insoles were tested sequentially during a single measurement session in a randomized order. For each type of insole, the first or second trials were used as accommodation trials, while the data of five subsequent trials were employed for the analysis.

2.4. Data analysis

Data were analyzed by a previously described method [7,20,21]. The rotation of the subtalar joint was defined as the rotation of the calcaneus relative to the lateral and medial malleoli. All joint moments were expressed as external moments and were normalized to the subject's body weight and height and expressed as a percentage of body weight × height (%Bw × Ht).

Three kinematic parameters were also acquired from the measurements (Fig. 2). Progression angle was the angle between the direction of gait progression and the foot axis at midstance. Step width was the distance between the centers of pressure (COPs) of the right and left foot across the direction of gait progression, and step length was the distance traversed by a single step.

The kinetic and kinematic parameters of knee and subtalar joints were first compared with the four types of insoles for the entire stance phase. Then the stance phase was divided into three sections of equal length, and the parameters for the four insoles were compared in the respective sections. The results are shown by the mean +, −, or ± standard error of means (S.E.M.).

2.5. Statistical analysis

Statistical analysis was performed with the Dr. SPSS-II software (version 11.01.1J, SPSS Japan, Tokyo, Japan). Data were initially analyzed by repeated measures one-way analysis of variance (ANOVA) and, when necessary, Dunnett's multiple comparison was employed as a post hoc test. The level of significance was set at $p < 0.05$.

3. Results

3.1. External adduction moment at the knee joint

The time–distance parameters and ground reaction force that could affect the kinetics and kinematics of the knee and subtalar

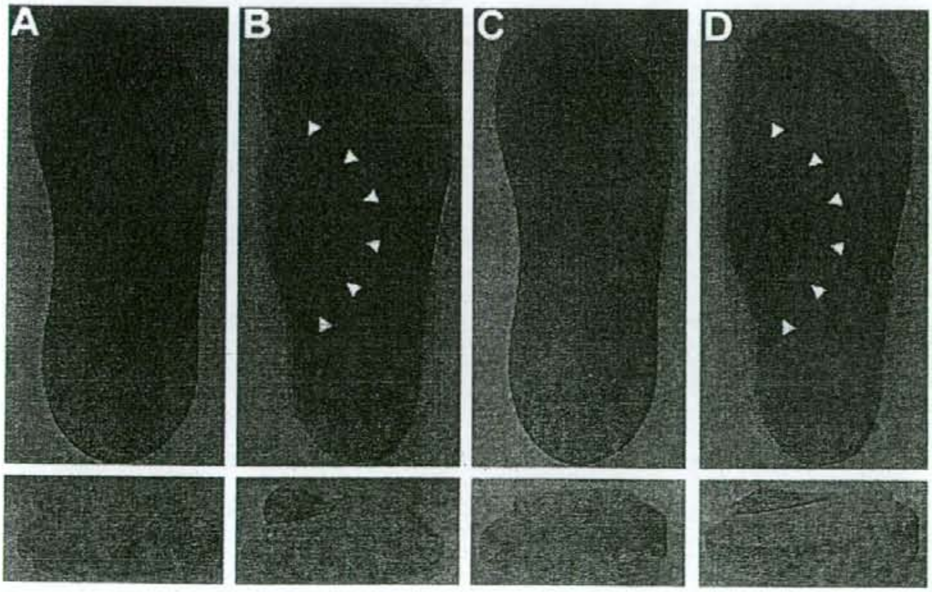

Fig. 1. Insoles used for this study. Upper (upper panels) and posterior views (lower panels) of FLAT (A), AS (B), LW (C), and LWAS (D) are shown. In (B) and (D), white arrowheads indicate arch supports added to the insoles.

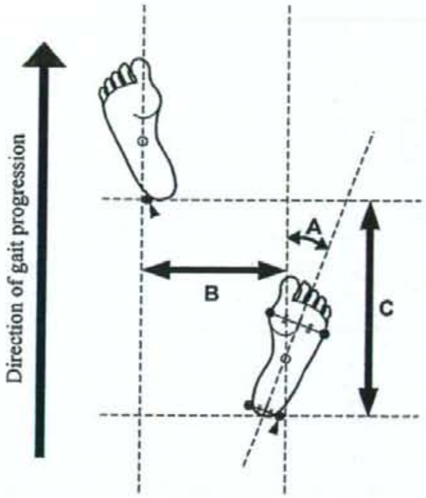

Fig. 2. Three kinematic parameters evaluated. Progression angle (A) is the angle formed by the direction of gait progression and the foot axis, which is an assumptive line passing through the middle of the first and fifth metatarsal heads and the middle of the calcaneal tuberosity at the midstance (red dotted line). Step width (B) is the distance between the COPs of right and left feet at midstance vertically across the direction of gait progression. Step length (C) is the distance of right and left feet in a single step along the direction of gait progression obtained from the positions of markers on the lateral aspects of calcaneal tuberosities at heel strike (solid arrowheads). Black dotted lines are assumptive lines drawn parallel to the direction of gait progression passing through the COPs, while blue ones are those drawn vertically to the direction of gait progression, passing markers at the lateral aspects of calcaneal tuberosities. Red circles indicate markers placed at the metatarsal heads, and blue ones denote those at the lateral and medial aspects of calcaneal tuberosities. Open circles represent COPs. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of the article.)

joints were not significantly different among the insoles (Supplementary Table 2).

In accordance with previous studies [5,8,22–25], external adduction moment of the knee joint presented a two-peak pattern during the stance phase (Fig. 3A). Considering this, the stance phase in the current study was divided into three parts of equal length (early, middle, and late sections), and the effect of the insole was evaluated in respective sections as well as for the entire stance phase.

Among the four types of insoles, the peak knee adduction moment was the highest with FLAT (Fig. 3B). While the moment with AS was similar to that with FLAT, the peak moment was significantly reduced with LW and LWAS compared to FLAT ($p = 0.010$ and $p = 0.034$, respectively). The adduction moment averaged for the entire stance phase showed a similar change with the insoles (Fig. 3C). The mean moment was highest with FLAT, followed by AS and LW, and lowest with LWAS. The change of the mean moment with the insoles differed from that of the peak moment in that the reduction was more obvious with LWAS than with LW. The reduction of the mean moment with LW and LWAS was 7.7% and 13.3%, respectively, compared to that with FLAT. The mean moment with LWAS was significantly lower than that with LW ($p = 0.002$). Next, the knee adduction moment was averaged in each of the three stance phase sections, and compared among the insoles (Fig. 3D). In the early section, the moments with LW and LWAS were slightly reduced compared to that with FLAT, but the reduction was not significant for either insole. In the middle section, the moment was significantly reduced with LW and LWAS compared to that with FLAT ($p = 0.003$ and $p < 0.001$, respectively). In the late section, the moment was obviously reduced with LWAS, which was found to be significantly lower than that with LW ($p < 0.001$) as well as that with FLAT ($p < 0.001$). Although the moment with LW was lower than that with FLAT, the reduction in this section did not reach the level of significance ($p = 0.053$).

3.2. Valgus angle at the knee joint

In order to examine whether the observed difference in the adduction moment was related to the change in the kinematics of the knee joint, the valgus knee joint angle was compared among the four insoles, and no significant difference was found among any of them (Supplementary Figure). Therefore, it is unlikely that the change of knee adduction moment with the insoles was caused by any difference in knee joint kinematics.

3.3. External abduction moment and valgus angle at the subtalar joint

Compared with FLAT, the peak abduction moment at the subtalar joint was significantly higher with LW and LWAS than with FLAT ($p < 0.001$ for both), while it was almost unchanged with AS (Fig. 4A). The level of increase was similar for LW and LWAS. A similar trend was observed when the moment was evaluated in each of the three sections during the stance phase (Fig. 4B). That is, the moment was not altered with AS in either section, but was equally increased with LW and LWAS compared to FLAT in all three sections ($p \leq 0.003$ for LW and $p < 0.001$ for LWAS).

The valgus angle of the subtalar joint was averaged for each section of the stance phase and compared among the insoles (Fig. 4C). Throughout the sections, the valgus angle was lowest with FLAT, followed by AS and LWAS, and highest with LW.

From these results, the addition of the arch support to LW may indeed tend to reduce the change of subtalar valgus angle, while exerting a similar level of abduction moment at the joint to that with LW.

3.4. Progression angle and step width

The progression angle and step width were compared for the four insoles. The progression angle was lowest with LW, and highest with LWAS (Fig. 5A). The difference in the angle between those two insoles was significant ($p = 0.037$). This result indicates that the use of LW tended to induce a toe-in gait, but this trend was completely reversed by the addition of an arch support to LW. Meanwhile, the step width increased most with LW, and declined most with LWAS (Fig. 5B). The difference in width between those two insoles was statistically significant ($p = 0.033$). Comparison between LW and LWAS revealed that the step width tended to increase with LW, but that increase was completely eliminated by the addition of an arch support. These changes in the progression angle and step width imply that the gait pattern could be altered by the use of LW, but it may be normalized by the addition of an arch support.

4. Discussion

In our study, the peak knee adduction moment was reduced by approximately 8.8% by the use of LW. This level of reduction was similar to those in previous reports [5,6,26], which would support the validity of our measurements. Although the peak moment was not changed by the addition of an arch support to LW, the knee adduction moment averaged for the entire stance phase was significantly reduced by it (Fig. 3B and C). This reduction of the moment was most obvious in the late stance (Fig. 3D). Our current

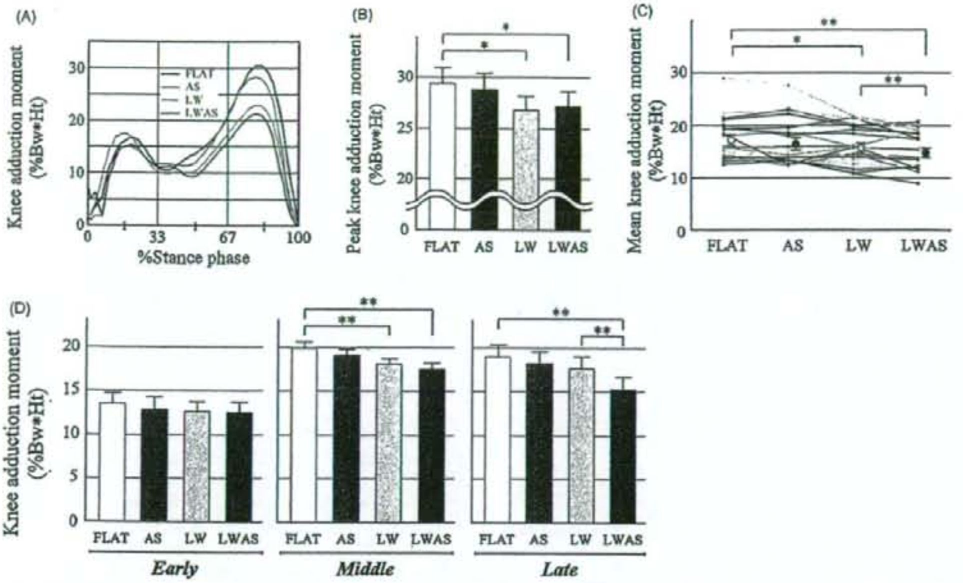

Fig. 3. External adduction moment of the knee joint with the four types of insoles. (A) Knee adduction moment with the four types of insoles during the stance phase. Representative result of a single subject is shown. Data are averages of five measurements. (B) Peak values of knee adduction moment with the four types of insoles. (C) Adduction moment averaged for the entire stance phase. Results of respective subjects are shown by lines of different colors together with the mean values of all subjects. (D) Adduction moment was averaged in respective sections of the stance phase and compared among the insoles. In (B)–(D), values are the mean + or ± S.E.M. $^{*}p < 0.05$ and $^{**}p < 0.01$, respectively.

analysis also revealed that the use of LW reduced the progression angle, and that such a change in the angle was fully reversed by the addition of an arch support (Fig. 5A). The increase in progression angle has been shown to decrease the knee adduction moment in the late stance [4,27,28]. Therefore, it is very likely that the arch support added to LW reduced the knee adduction moment through the increase in the progression angle. Meanwhile, the finding that the progression angle decreased with the use of LW implies that

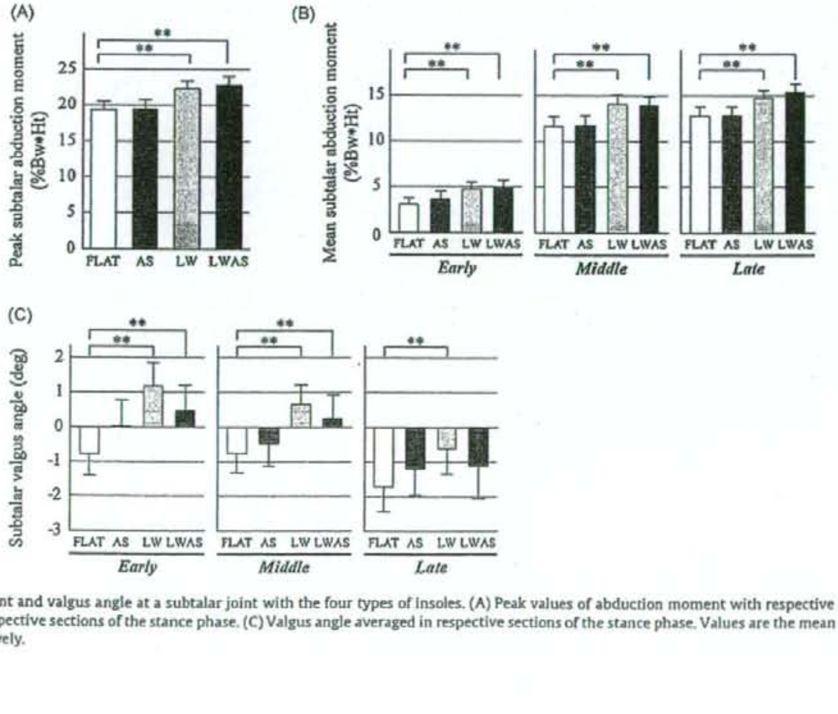

Fig. 4. Abduction moment and valgus angle at a subtalar joint with the four types of insoles. (A) Peak values of abduction moment with respective insoles. (B) Abduction moment averaged in respective sections of the stance phase. (C) Valgus angle averaged in respective sections of the stance phase. Values are the mean + or − S.E.M. $^{*}p < 0.05$ and $^{**}p < 0.01$, respectively.

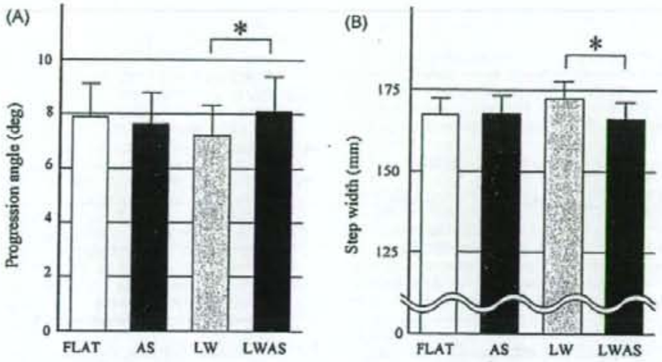

Fig. 5. Progression angle (A) and step width (B) with the four types of insoles. Values are the mean + S.E.M. $^{*}p < 0.05$.

the effect of LW in reducing the knee adduction moment could be impaired to some extent by a toe-in gait induced by it. A reduction of the progression angle by LW was reported in another study [22]. This would pose a potential drawback with that type of insole. As shown here, this drawback may be completely eliminated by the addition of an arch support. Reduction of the progression angle increases the risk of progression of medial knee OA, probably through an increase of the knee adduction moment in the late stance [29]. Therefore, such changes of progression angle should be considered carefully when insoles are used to treat knee OA patients.

The present study also revealed another potential problem with the conventional laterally wedged insole. Our current observation and those of others have consistently indicated that the use of LW increased step width (Fig. 5B) [8,12]. The wider the step width becomes, the more lateral the position of the ground reaction force would be from the center of gravity of the body, and this would increase knee adduction moment. Therefore, in addition to the change of progression angle, the increase in step width may be yet another factor limiting the effect of LW in reducing the knee adduction moment. Since the step width with LWAS was smaller than that with the control insole, the addition of an arch support to LW appeared to completely eliminate this second possible drawback of the conventional wedged insole.

Another advantage of the additional arch support was suggested by an analysis of the kinetic and kinematic parameters at the subtalar joint. We previously reported that a laterally wedged insole alters the kinetics and kinematics of the subtalar joint [7,20,21]. In accordance with those results, the use of a laterally wedged insole increased the abduction moment and valgus angle at the subtalar joint. The addition of an arch support to LW tended to reduce the valgus angle of the joint (Fig. 4C), while keeping the abduction moment equal to the level of LW (Fig. 4A and B). During the measurements, some subjects complained of instability or foot discomfort when wearing LW, but that feeling was considerably relieved with LWAS. The discomfort associated with the use of LW may be surmised to have stemmed from over-abduction of the subtalar joints. This may have been alleviated by the addition of an arch support, which reduced the degree of abduction.

Thus, the addition of an arch support to the laterally wedged insole changed all of the progression angle, step width, and valgus angle at the subtalar joint closer to the levels of the control insole. Therefore, it may be reasonable to assume that the addition of an arch support to LW allowed the subjects to walk in a more "natural" manner, while increasing the effect of the wedged insole in reducing the knee adduction moment.

A significant limitation of this study is that the biomechanical effect of insoles was investigated in healthy volunteers but not in the actual OA patients. The differences in the types of wedged insoles is another issue that was not addressed in this study, as insoles with shorter wedging or other inclinations may be used in clinics [13,14,30]. Since insoles are used more often within the shoes, our measurement without shoes may not reflect the actual situation of their use. This could also be a limitation of this study. These points should be addressed in future studies.

Medial knee OA deteriorates in a vicious circle of increasing varus angulation and loading of the medial compartment. The use of a laterally wedged insole is expected to prevent further disease progression by breaking this circle [5]. However, at present, insole therapy has not yet become a common treatment for knee OA, primarily because of its limited efficacy [7,9,13–16]. We hope that our current findings will be useful in modifying the conventional lateral wedged insole to obtain better clinical results.

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Conflict of interest statement

None of the authors has any conflict of interest regarding this study.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.gaitpost.2008.08.007.

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ELSEVIER

Biomechanical characteristics of the knee joint in female athletes during tasks associated with anterior cruciate ligament injury

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ABSTRACT

This study was designed to compare biomechanical characteristics of the knee joint for several athletic tasks to elucidate their effects and to examine what tasks pose a risk for ACL injury.

Three athletic tasks were performed by 24 female athletes: single-limb landing, plant and cutting, and both-limb jump landing. Angular displacements of flexion/extension, abduction/adduction, and external/internal tibial rotation were calculated. Angular excursion and the rate of excursion of abduction and internal tibial rotation were also calculated.

During plant and cutting, from foot contact, subjects rotated the tibia more rapidly and to a greater degree toward internal tibial rotation. Moreover, excursion of knee abduction is greater than that during single-limb landing. During both-limb jump landing, the knee flexion at foot contact was greater than for either single-limb landing or plant and cutting; peak knee abduction was greater than for either single-limb landing or plant and cutting.

In plant and cutting, the risk of ACL injury is increased by greater excursion and more rapid knee abduction than that which occurs in single-limb landing, in addition to greater internal tibial rotation. Although single-limb tasks apparently pose a greater risk for ACL injury than bilateral landings, both-limb landing with greater knee abduction might also risk ACL injury.

1. Introduction

Anterior cruciate ligament (ACL) injury is a serious injury in sports activities. After ACL injury, most athletes must undergo ligament reconstruction and continue rehabilitation for 6 months to a year before returning to sports activities [1]. The rate of ACL injury is reportedly much higher for female athletes than for males [2,3]. Additionally, almost 70% of situations causing ACL injury are noncontact situations: landing from a jump, stopping after fast running, and cutting to a different direction [2,4].

Understanding the mechanisms of ACL injury is important for its prevention. Olsen et al. [5] described ACL injury mechanisms from viewing videotapes of ACL injuries. They concluded that the main injury mechanism for ACL injuries is a forceful valgus collapse with the knee close to full extension, combined with external or internal rotation of the tibia. However, ACL injuries occur rapidly during games and practice sessions. In most cases, it is difficult to determine the mechanisms of ACL injury from videotapes or pictures recording the injury situation because of the image quality. Therefore, many researchers have examined injury mechanisms from motion capture images taken in laboratory conditions.

Numerous studies using motion capture systems have examined the mechanism and risk factors of ACL injury during athletic tasks according to gender differences. As described previously, female athletes are more prone to sustaining ACL injury than male athletes. Therefore, female characteristic kinematics and kinetics are thought to be risk factors related to ACL injury mechanisms. Earlier studies have shown that female athletes demonstrate larger knee valgus than male athletes during landing or many other athletic tasks [6–12]. Hewett et al. [13] measured kinematics and joint loads using kinetics during a jump-landing task prospectively: results showed that female athletes with increased dynamic valgus and high abduction loads are at increased risk of anterior cruciate ligament injury. Therefore, knee valgus has been recognized as a risk factor and one mechanism of ACL injury. Tibial rotation during athletic tasks has been examined recently: we examined gender differences of tibial rotation during single-limb drop landing and estimated that the risk factor and mechanism of ACL injury would be greater for tibial internal rotation combined with knee valgus [14].

Another approach to examination of the mechanism of ACL injury using motion capture systems is analysis of biomechanical characteristics during tasks that pose a high injury risk for ACL injury. In fact, ACL

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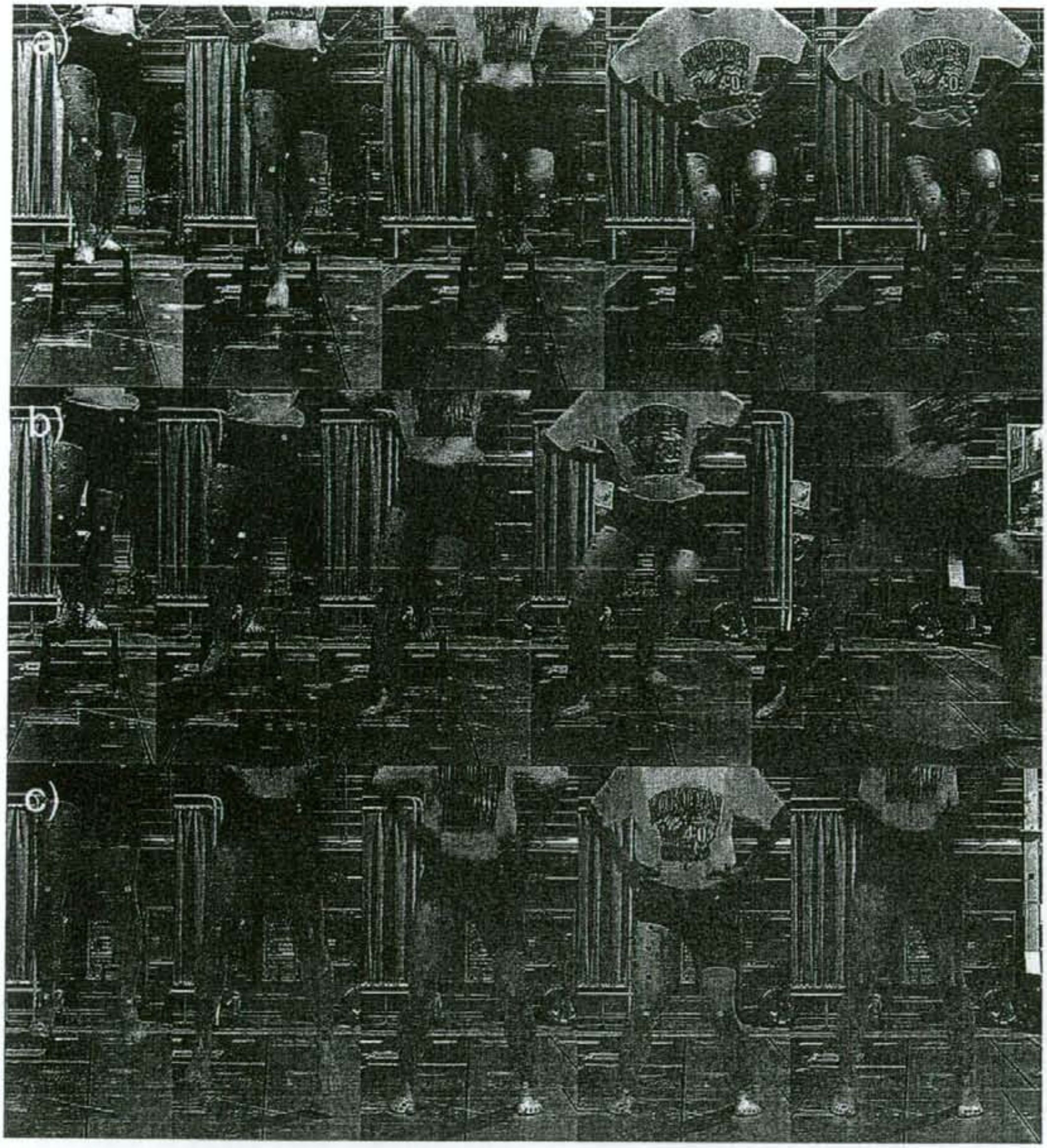

Fig. 1. Sequential photographs of experimental tasks: Single-limb landing (a), plant and cutting (b), and both-limb jump landing.

injuries often occur in plant and cutting movements while leaning on one leg and forcing a knee valgus [4,5]. Sell et al. [15] examined the effects of direction during a two-legged stop-jump task and concluded that lateral jumps are the most risky manoeuvres for ACL injury. Pappas et al. [16] compared bilateral and unilateral landings and found that, in unilateral landings, subjects performed high-risk kinematics with increased knee valgus, decreased knee flexion, and decreased relative hip adduction. However, they only analyzed knee valgus at initial contact during landings and did not examine the plant and cutting manoeuvre, which is thought to pose greater risk for ACL injuries. The characteristics of plant and cutting and several athletic tasks have never been well established.

This study was intended to compare biomechanical characteristics of the knee joint between plant and cutting tasks and normal single-limb landing, and to compare characteristics between both-limb jump landing and single-limb tasks. Comparison of kinematics among tasks can elucidate the characteristics of these tasks, and enable examination of what tasks pose a risk for ACL injury. Understanding risky tasks and movements can help prevent ACL injury because team trainers and coaches might thereby be better able to instruct their athletes to avoid such movements. Our hypotheses were two. During a plant and cutting manoeuvre, subjects demonstrate riskier kinematics for ACL injury than during normal single-limb landing because of greater knee valgus and greater internal tibial rotation. In addition, during single-

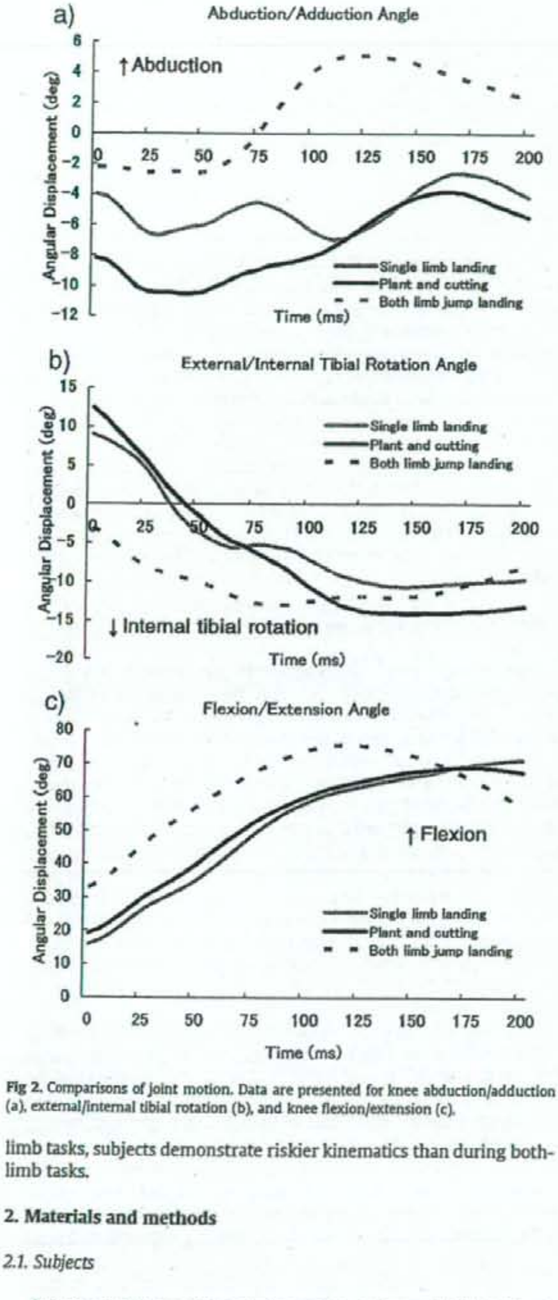

Fig 2. Comparisons of joint motion. Data are presented for knee abduction/adduction (a), external/internal tibial rotation (b), and knee flexion/extension (c).

limb tasks, subjects demonstrate riskier kinematics than during both-limb tasks.

2. Materials and methods

2.1. Subjects

A power analysis conducted during a pilot study revealed that at least 24 subjects were necessary to achieve 80% statistical power with an α level of 0.05. In all, 24 female athletes were recruited for the experiment. Half were basketball players; others were lacrosse players. Subjects were excluded from the study if they had a history of serious musculoskeletal injury, any musculoskeletal injury within the past 6 months, or any disorder that interfered with sensory input, musculoskeletal function, or motor function. Before participation, all subjects provided written informed consent in accordance with approval by the Institutional Review Board of National Rehabilitation Center for Persons with Disabilities. The average age of subjects was 21.1 (1.3) yr (Mean (SD)); their average height was 166.1 (8.3) cm and their average weight was 59.3 (8.2) kg. All subjects were right-leg dominant. The dominant leg was determined as the leg used to kick a ball.

Table 1
Mean (SD) for tasks observed power of joint angle at the time of foot contact

	Knee abduction		External tibial rotation		Knee flexion	
Single limb landing	−4.0 (2.6)		9.0 (3.4)		15.8 (5.0)	
Plant and cutting	−8.2 (3.1)**	*	2.4 (4.3)**	**	19.2 (7.0)**	**
Both limb jump landing	−2.2 (3.4)**		−3.0 (5.2)**		32.8 (7.1)**	
Observed power	1.0		1.0		1.0	

*: $p<0.05$, **: $p<0.01$.

2.2. Experimental task

All subjects were measured in a static standing position and during performance of three athletic tasks: single-limb landing, plant and cutting, and both-limb jump landing. For the single-limb landing, subjects stood on a 30-cm-high platform with the left limb, and landed on a platform 30 cm away with the right limb (Fig. 1a). They were required to unyoke their left foot from a platform, and, when they start a landing motion, not to land the right limb along with their left limb on a platform. A trial was considered successful if they retained the landing position. For the plant and cutting, subjects stood on a platform, as in the single-limb landing. They were required to land with their right foot 45 abducted from the original direction and to push off their foot perpendicularly (to the left) with the right foot to make a cut (Fig. 1b). They also were required to make three steps after the cut. A trial was considered successful if they landed with their foot at the prescribed angle and made a cut to the prescribed direction. For both-limb jump landing, subjects performed vertical jumps five times using both legs with maximum effort [17] (Fig. 1c). They were instructed to stand with their feet shoulder-width apart and face the frontal plane during testing. The subjects were given verbal instruction to shorten their foot contact time as much as they were able and to jump as high as they were able. The landings from the second to fourth time of their dominant limb were measured for analysis. Throughout the experiment, the subjects were barefoot and kept their hands on their lower torso. The subjects were allowed to perform several preparation trials. Measurements were continued for three successful trials: each was conducted consecutively.

2.3. Data collection

All experiments were performed at the National Rehabilitation Center for Persons with Disabilities in Saitama, Japan. A seven-camera high-speed motion analysis system (Hawk; Motion Analysis Corp., Santa Rosa, CA) was used to record the lower-limb movements three-dimensionally. The motion and force data were recorded at 200 Hz. The laboratory was equipped with six force plates (9287A; Kistler Japan Co., Ltd., Tokyo, Japan). Vertical ground-reaction force was used to signal the initial contact to determine the data capture period.

Table 2
Mean (SD) for tasks observed power of peak joint angle

	Knee abduction		Internal tibial rotation		Knee flexion	
Single limb landing	−1.2 (5.2)		12.3 (5.5)		72.5 (6.7)	
Plant and cutting	−2.6 (6.1)	*	14.4 (6.0)*	**	70.4 (8.5)	
Both limb jump landing	7.1 (5.5)**		14.9 (5.5)		80.3 (16.4)	*
Observed power	1.0		0.96		0.88	

*: $p<0.05$, **: $p<0.01$.

Table 3
Mean (SD) for angular excursion (deg) and rate of excursion (deg/ms)

	Knee abduction		Internal tibial rotation	
	Excursion	Rate	Excursion	Rate
Single limb landing	6.6 (3.6)	0.12 (0.05)	21.4 (6.4)	0.15 (0.06)
Plant and cutting	9.8 (3.8)**	0.13 (0.04)	26.8 (6.8)**	0.22 (0.07)
Both limb jump landing	11.2 (3.6)	0.14 (0.05)	12.1 (4.9)**	0.14 (0.05)

*: $p<0.05$, **: $p<0.01$.

To each subject, 25 reflective markers of 9 mm diameter were secured to the lower limb using double-sided adhesive tape, as described in a previous study [14]. The markers were used to implement the Point Cluster Technique (PCT) [18]. We calculated knee kinematics using the joint coordinate system proposed by Grood and Suntay [19]. For PCT, the skin markers are classified into two groups: a cluster of points representing a segment and points representing bony landmarks. For a cluster of points, 10 and 6 markers were attached respectively to the thigh and shank segments. The bony landmarks were the great trochanter, the lateral and medial epicondyles of the femur, the lateral and medial edges of the tibia plateau, the lateral (fibula) and medial malleoli, and the fifth metatarsophalangeal joint.

2.4. Data analysis

The coordinate data obtained from the markers were not smoothed because of the expected noise-cancelling property of the PCT. In each trial, we calculated the angular displacements of flexion/extension, abduction/adduction, and external/internal tibial rotation using the PCT. The reference position for these measurements was obtained during the static trial. We analyzed each variable at the time of foot contact and the peak value from the foot contact to 200 ms thereafter. Additionally, angular excursion for knee abduction and internal tibial rotation was calculated. A rate of excursion for knee abduction and internal tibial rotation was also calculated.

All dependent variables were calculated for each trial, then averaged across the three trials. A repeated measures one-way ANOVA was used to test for task differences in joint angle at the foot contact and peak joint angle. The alpha level was set at $p<0.05$. A post hoc Bonferroni multiple comparison test was performed for each variable to determine differences among tasks. Intraclass correlation coefficients (ICC (1, 3)) were calculated to determine the measurement consistency.

3. Results

Acceptable ICC (1, 3) values at the time of foot contact and a peak value were established for knee abduction/adduction (0.98, 0.97), external/internal tibial rotation (0.93, 0.98), and flexion/extension (0.96, 0.89). Fig. 2 portrays mean time course comparisons across tasks for the three angular displacements of the knee (abduction/adduction, external/internal tibial rotation, and flexion/extension).

Means, standard deviations and observed power for all variables at the time of foot contact are presented in Table 1. The adduction angle in plant and cutting was significantly larger than that for either single-limb landing or both-limb jump landing ($p<0.01$, respectively); that in single-limb landing was significantly larger than that of both-limb jump landing ($p<0.05$). The external tibial rotation angle in plant and cutting was significantly larger than for either single-limb landing or both-limb jump landing ($p<0.01$); that in single-limb landing was significantly larger than that of both-limb jump landing ($p<0.01$). The flexion angle in both-limb jump landing was significantly larger than that of either single-limb landing or plant and cutting ($p<0.01$); that in plant and cutting was significantly larger than that of single-limb landing ($p<0.01$).

Means and standard deviations of peak values for all variables are presented in Table 2. The peak abduction angle in both-limb jump landing was significantly larger than that of either single-limb landing or plant and cutting ($p<0.01$ and $p<0.05$, respectively). During single-limb landing or plant and cutting, their knee was abducted from foot contact with time. However, even at their peak, it is adducted. The peak internal tibial rotation angles in plant and cutting and both-limb jump landing were significantly larger than that of single-limb landing ($p<0.05$ and $p<0.01$, respectively). The peak flexion angle in plant and cutting was significantly smaller than both-limb jump landing ($p<0.05$).

The angular excursion and velocity for knee abduction and internal tibial rotation are presented in Table 3. The excursion for knee abduction in plant and cutting and both-limb jump landing was significantly larger than that for either single-limb landing ($p<0.01$, respectively). The rates of excursion for knee abduction among three tasks were not significantly different. The excursion for internal tibial rotation in plant and cutting was significantly larger than for either single-limb landing or both-limb jump landing ($p<0.01$, respectively), whereas that in single-limb landing was significantly larger than that of both-limb jump landing ($p<0.01$). The rate of excursion for internal tibial rotation in plant and cutting was significantly faster than that for either single-limb landing or both-limb jump landing ($p<0.01$, respectively).

4. Discussion

The primary purpose of this study was to analyze the biomechanical characteristics of the knee joint during several athletic tasks, and to examine what tasks present a risk for ACL injury. A plant and cutting manoeuvre is a movement that commonly causes ACL injury, of which most situations were single-foot push-offs [5]. However, biomechanical characteristics of plant and cutting and several athletic tasks are unknown. Therefore, to compare a plant and cutting and normal single-limb landing as well as both limb landing, we can understand these athletic tasks and examine what tasks are risky for ACL injury. The results of this study showed that greater excursion and more rapid knee abduction occur in plant and cutting than that which occurs in single-limb landing, in addition to greater internal tibial rotation. Furthermore, compared to similar single-limb tasks, both-limb jump landing knee flexion and knee abduction were greater; external tibial rotation at the foot contact was smaller.

4.1. Plant and cutting versus single-limb landing

Some recent studies have compared biomechanical characteristics across different athletic tasks [8,15,20]. Nevertheless, these studies present some limitations. Although Chappell et al. [8] compared knee kinematics of forward, vertical, and backward stop-jump tasks, they did not examine lateral movement. Sell et al. [15] compared two-legged stop-jump tasks in three different directions. Although their results indicate that lateral jumps are the most dangerous of the stop-jumps, all tasks were two-legged tasks, not single-leg tasks. Besier et al. [20] compared the joint load during running, sidestep cutting, and crossover cutting. They inferred that external moments applied to the knee joint during the stance phase of the cutting tasks place the ACL and collateral ligaments at risk of injury, but they did not analyze joint kinematics and the frequency of the motion analysis system was too slow to support examination of high-speed athletic tasks. Therefore, the results of this study, along with those of the prior study, provide some implications of mechanisms causing ACL injury.

The results of this study showed that, during plant and cutting, external tibial rotation at the foot contact and peak internal tibial rotation were greater than during single-limb landing. During plant and cutting, from foot contact, subjects rotated the tibia more rapidly and to a greater degree toward internal tibial rotation than during single-limb landing. Previous studies [8,15,16] that examined the mechanism of ACL injury have not analyzed tibial rotation during high-risk movement, probably because of technical issues. In this study, we analyzed tibial rotation using PCT. An anatomical study has demonstrated that internal tibial rotation increases the strain of ACL [21]. Therefore, biomechanically and anatomically, plant and cutting presents a high risk for ACL injury.

During plant and cutting, subjects demonstrated more increased knee adduction at foot contact than during single-limb landing. After foot contact, during single-limb landing, subjects showed twin peaks of knee abduction. During plant and cutting, subjects moved toward knee abduction with time, although subjects did not exhibit a great magnitude of knee abduction. Consequently, during plant and cutting, excursion of knee abduction was greater than during single-limb landing. Therefore, during plant and cutting, greater excursion of knee abduction occurred than during single-limb landing combined with greater internal tibial rotation to push off their body to the other side and change direction.

4.2. Both-limb jump landing versus single-limb tasks

Some studies have analyzed kinematics or kinetics during bilateral landing to examine ACL injury mechanisms [11,12,22]; other studies have screened risks for ACL injury [13] or lower limb injury [23,24]. However, few studies have examined the characteristics of bilateral landing in comparison to single-limb landing. Only Pappas et al. [16] compared bilateral and unilateral landings. Their results indicated that, in unilateral landings, subjects performed high-risk kinematics with increased knee valgus, decreased knee flexion, and decreased relative hip adduction. However, they showed no peak knee valgus or tibial rotation during landing.

The results of this study demonstrated that, during both-limb jump landing, knee flexion at foot contact was greater than for single-limb landing and plant and cutting, and that peak knee flexion was greater than plant and cutting. These results were consistent with those of a previous study [16]. Pappas et al. [16] speculated that subjects might attempt to prevent falls by limiting excessive knee flexion during unilateral landing compared to bilateral landing, while simultaneously increasing the forces in ACL. Additionally, in slight knee flexion, i.e. less than 30 , contraction of the quadriceps strains the ACL [21,25,26]. For that reason, slight knee flexion is inferred as a risk factor of ACL injury. During a process of prevention training leading athletes to increased knee flexion can decrease the incidence of ACL injury. On the other hand, during both-limb landing, external tibial rotation at the foot contact was less than that during single-limb landing and plant and cutting, while peak internal tibial rotation was not significantly different with plant and cutting. Unilateral landing has a greater excursion of tibial internal rotation than bilateral landing. As described above, an anatomical study has demonstrated that internal tibial rotation increases the ACL strain [21]. Consequently, characteristics of unilateral landing that have less knee flexion and greater internal tibial rotation present a higher risk for ACL injury than bilateral landings.

During both-limb jump landing, peak knee abduction was greater than for either single-limb landing or plant and cutting, while knee adduction at foot contact was smaller. These results did not support our hypothesis. We speculate that knee abduction was limited compensatory for greater internal tibial rotation and smaller knee flexion to prevent ACL injury during single-limb tasks. The possibility of ACL injury arose when subjects allowed greater knee abduction during single-limb tasks. Another reason might be that, because ACL injury occurs not only in single-limb situations but also in both-limb jump landing, the latter also poses a risk for ACL injury. Krosshaug et al. [27] analyzed videos of ACL injury situations and reported that ACL injury occurred during two-legged landing in 9 of 22 cases of female player situations, although it occurred in only four cases of one-legged landing. Therefore, it is thought that both-limb landing with greater knee abduction might also pose a risk for ACL injury.

Greater knee abduction was apparent during a both-limb jump landing task. For screening of ACL injuries, we detected knee abduction well in this task. It is difficult to detect a risk demonstrating greater knee abduction during single-limb tasks because of these characteristics, which demonstrate limited knee abduction. Moreover, knee abduction during both-limb landing can be evaluated using a two-dimensional approach, which uses a video recorder and analyzes a frontal projected knee valgus angle [17]. Some studies have been conducted using comparable methods [23,28]. Consequently, considering convenience and efficiency, both-limb jump landing is thought to be valuable for screening the risk of ACL injury.

4.3. Limitations

This study has important limitations. Influences of the hip and ankle have recently been suggested [9,29]. However, the present study analyzed the kinematics of the knee only. Additionally, although joint kinetics holds great importance for analyses of athletic tasks and for examination of the mechanisms of injuries, we only analyzed knee kinematics because we have not developed a joint-moment calculation system corresponding to PCT. Future studies should examine the relation between kinematic data and kinetics data to assess the ACL injury mechanism.

5. Conclusion

We compare the biomechanical characteristics of the knee joint for several athletic tasks to elucidate the characteristics of single-limb landing, plant and cutting and both-limb landing, and to examine what tasks present a risk for ACL injury. The results indicate that, in plant and cutting, knee abduction combined with internal tibial rotation poses a risk of causing ACL injury. Both-limb landing with greater knee abduction might also pose risks for ACL injury.

6. Conflict of Interest

No author of this manuscript has any conflict of interst.

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卒後
研修講座

変形性膝関節症のリスクファクター*

大森　豪**

はじめに

変形性膝関節症（膝OA）は年齢に伴う膝関節の退行性疾患であり，日常診療の場において変形性関節症の中では腰椎に次いで頻度が高い．本症の病態は，関節軟骨の変性と摩耗を主体として骨や軟骨下骨，滑膜，半月板，靱帯といった関節構成体に炎症反応や増殖性変化，変形性変化が生じ，結果的に関節破壊の進行にいたる一連の過程として理解される．膝OAは"common disease"であり，その発症と進行には多数の因子が関与している．これらの因子は，膝関節に限局する局所因子と全身性因子あるいは遺伝要因と環境要因などに区分され，疫学や生体力学，生化学，画像解析などさまざまなアプローチで研究が行われている．

膝OAの治療や予防を考えるうえでリスクファクターを理解することはきわめて重要であり，本稿では膝OAの発症・進行因子について，これまでに明らかになっているものおよび残された課題を含めて概説する．なお，ここで述べる膝OAとはX線像にて診断されたradiographic OAであり，膝痛などの症状を有するsymptomatic OAではない．

1 年齢と性別

男女とも40歳代以降年齢とともに膝OAの頻度は増加し，70歳代では男性で30〜40%，女性で50〜60%に達する[1]（図1）．40歳以降の各年代では女性が1.5〜2.5倍発症率が高くなっており，これらの点から加齢および女性は膝OAの危険因子といえる．興味深いのは40歳以下の年代では，Lawrenceら[2]は男性で5.5%，女性で3.9%，NHANES-Ⅰ[3]でも35〜44歳の群で男性1.75%，女性1.44%と逆に男性の発症率がわずかながら高く，比較的若年者の膝OA発症に靱帯，半月，軟骨損傷といった膝外傷が潜在的に影響している可能性を示唆する所見と考えられる．

2 人　種

これまでの報告では，欧米の白人，日本人，中国人における年代別の膝OA発生率はいずれも男女とも加齢とともに増加する（図1）．また，NHANES-Ⅰ[3]では米国内の黒人は白人に比べて男性で1.4倍，女性で2.8倍膝OAに対する危険度が大きいことも示されている．近年，異なる二つの人種を同一の解析手法で比較した研究が行われ，Zhangら[4]はFramingham studyのプロトコルを用いて調査を行い，中国人女性が白人女性に比べて有意に膝OAが多いことを示した．また，Yoshidaら[5]も同様の手法で日本人女性は白人女性に比べて1.9倍膝OAの危険度が高いと述べており，今後，同様の研究がすすむに従い人種間の相違や特徴が明らかになることが期待される．

Key words：medial knee osteoarthritis, risk factor, review

* Risk factor of knee osteoarthritis

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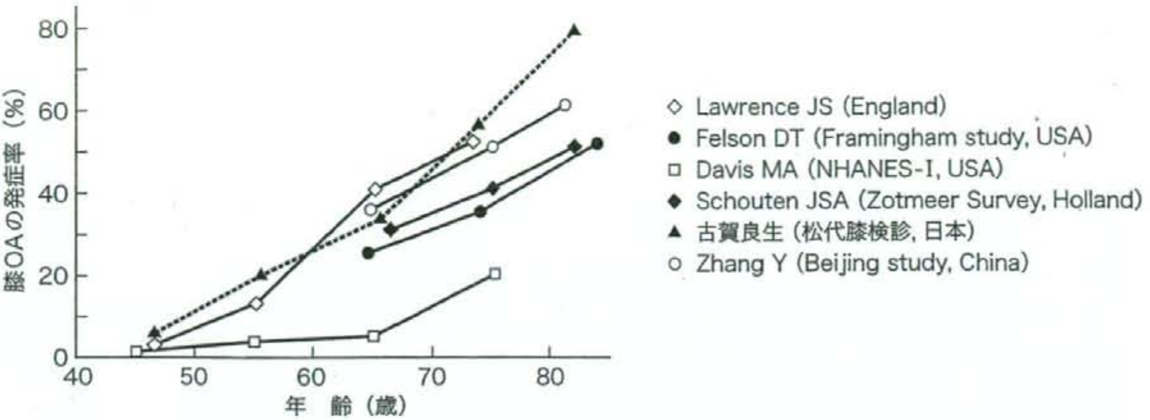
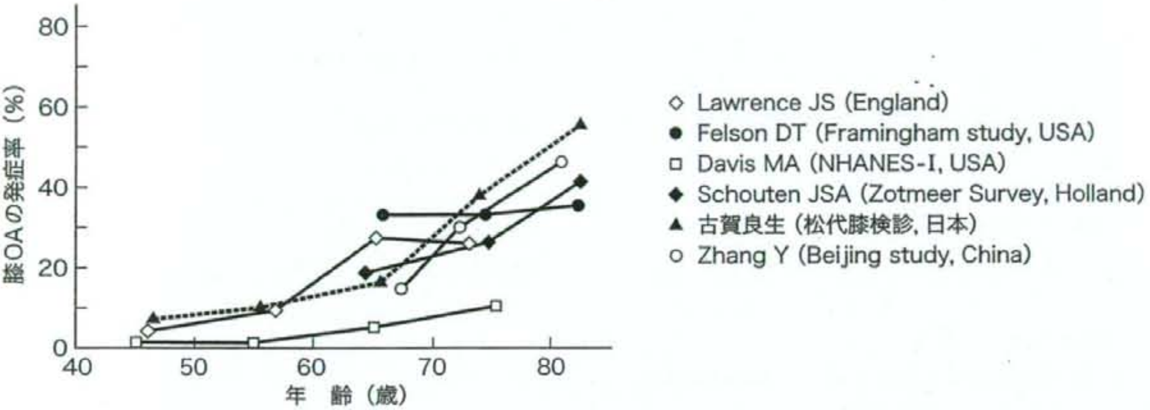


a．女



b．男

図 1．疫学調査による膝 OA の年齢別発症率

3 肥　満

膝関節には立位で体重の 1〜1.5 倍，階段昇降で 2〜3 倍，ジャンプ動作では 4〜6 倍の荷重がかかるとされ，機械的要因の観点から体重と膝 OA の関連性は明らかである．疫学調査においても膝 OA と肥満の有意な関連を示す報告は多く，肥満の指標として BMI［body mass index：体重（kg）/身長（m）^{2}］を用いた欧米の調査3,6)では BMI が 25 以上で 1.8〜3.8 倍，BMI が 30 以上で 3.8〜4.8 倍，35 以上の超肥満では 4〜7.8 倍の相対危険度が報告されている．日本では，われわれが行った調査（松代膝検診）7)において BMI が 24 以上で危険度が 2.1 倍に高くなり，また Yoshimura ら8)は和歌山県のコホートに対し case-control-study を行い，過去の肥満の既往が膝 OA 発症に関連することを述べている．男女差については，男性のほうが肥満の影響が大きいとする報告と女性のほうが大きいとする報告があり一定の見解は得られていない．また，肥満が膝 OA に影響するメカニズムとして膝関節への荷重負荷増大による機械的作用と高脂血症や糖尿病などの代謝性疾患による作用が考えられているが，次項に述べるように代謝性疾患による影響は少なく機械的作用が主体と考えられる．

4 代謝性疾患

偽痛風の原因であるピロリン酸カルシウム結晶（CPPD）やほかのカルシウムリン酸結晶（BCP）と膝 OA との関連性は古くから指摘されており，膝 OA の 50〜60% に関節液中に

表 1. 職業および日常活動性の膝OAへの影響

職業および日常活動性	膝OAへの影響
炭鉱労働者	男で影響あり
港湾労働者	男で影響あり
膝屈曲を要する職業（大工，トラック運転手など）	男で2.5，女で3.5（OR*）
力を要する職業（農夫，大工など）	男で1.8，女で3.1（OR）
膝屈曲＋力仕事	男で2.2，女で0.3（OR）
しゃがみ込み動作（1日30分以上）	6.9（OR）
膝つき動作（1日30分以上）	3.9（OR）
階段昇降（1日10段以上）	2.7（OR）
しゃがみ込み動作（1日1時間以上）	女で1.2（OR）
階段昇降（1日30段以上）	女で1.19（OR）
椅子の腰掛け（1日2時間以上）	女で0.77（OR）
しゃがみ込み動作（1日2時間以上）	女で2.4（OR），男で2.0（OR）

*OR：オッズ比

CPPDが存在するといわれている[9]．しかし，全身的な高尿酸血症の影響については明らかになっていない．また高脂血症，糖尿病，高血圧についても関連性があるとする報告とないとする報告があり一定の見解は得られておらず，現時点で代謝性疾患の膝OAへの直接的な関与は少ないと考えられる[10,11]．

5 喫煙

これまでの疫学調査では，タバコおよび葉巻の喫煙習慣と膝OAの発症は逆相関が認められ，喫煙は膝OAに無関係とするものから予防的効果の可能性すら指摘する報告もある[12,13]．しかしニコチンやタール，アンモニアといったタバコに含まれる成分が膝関節に及ぼす生物学的な影響についてはまったく解明されていない．

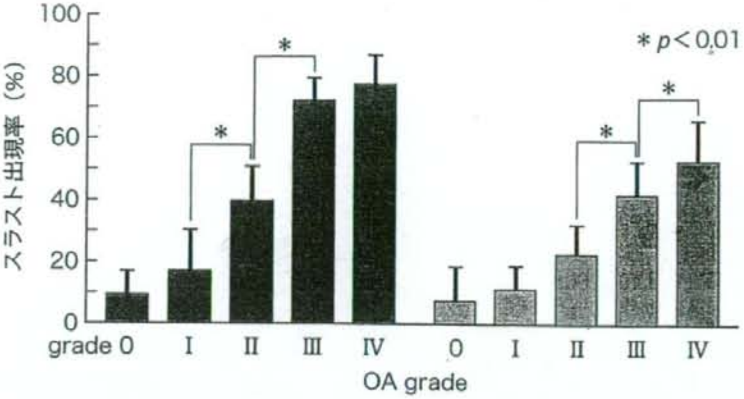
6 職業，生活様式，日常活動性と運動

職業や日常動作と膝OAとの関連性については多数の研究があり，膝の屈伸を伴う重労働の影響が大きいとする報告が多い（表1）．地域での生活習慣については，グリーンランドの狩猟民族やジャマイカの裸足生活者には膝OAが多いという報告もある[14,15]．

運動と膝OAとの関連では，ジョギングのように軽度～中等度の負荷にとどまる場合は影響が少ないとされている[16]．これに対し膝関節への負荷が増大する運動強度の高い種目では，次項に述べる半月板損傷や軟骨損傷，靱帯損傷といった膝外傷の合併との関連で検討され，膝OAに大きく影響するといった報告が多い．Sandmarkら[17]の行った調査によると，クロスカントリースキーやアイスホッケーでは，男性で2.9倍相対危険度が増すと述べている．

7 膝外傷

膝OAに影響する外傷としては靱帯損傷，半月板損傷，軟骨損傷，骨折があるが，未治療の膝外傷については診断が明確とならないため特定することは困難である．半月板損傷の影響は，生体力学研究により半月板切除による膝関節への著明な応力集中が証明されているが[18]，臨床的には治療として行った切除術後のOA変化を検討するものが多く，変性半月板断裂，半月板切除量が多いことが成績不良因子としてあげられている[19]．また，前十字靱帯損傷に関しては，保存的治療例または放置例において受傷後10～20年の経過で高率に膝OAが発症することが報告されている[20,21]．しかし，これらの臨床研究では対象者の年齢が40～50歳と比較的若いこと，X線像上のOA所見のわりに臨床症状が少ないことが指摘されている．われわれも膝半月板切除後30年以上の長期経過例を評価し，膝OA発症例に臨床症状が少なく可動域が良好な例が多いことを経験している[22]．さらに，既述した膝OAの発症率が40歳代までは男性が女性より多くその後逆転することを考えると，膝外傷後にみられる膝関節のOA変化は外傷に対する関節の生体変化である可能性が示唆される．そして，これが最終的に真の膝OAとなりさらに進行するかどうかは，



a. 女　　b. 男

図 2. 膝 OA grade とスラスト運動出現率

その個人のもつ膝OAのリスクファクターによって左右されると考えられる．しかし，この点に関する科学的なエビデンスはなく今後明らかにすべき課題である．

8 下肢筋力

大腿四頭筋力が膝OAの進行に伴い低下することは，疫学的な横断調査や患者を対象とした臨床研究において多数の報告がみられ，両者に関連性があることは明らかであるが，因果関係については不明な点が多い．最近の研究では単なる大腿四頭筋力低下ではなく，日常生活動作（ADL）における大腿四頭筋反応時間[23]や，膝屈筋とのバランス[24]，関節位置覚[25]，スラスト運動を含めた関節安定性[26]などほかの要素を含めて膝OAとの関連性を述べたものが多い．われわれは松代膝検診における縦断解析より，大腿四頭筋力低下が後述するスラスト運動を介して膝OA発症に影響することを明らかにしている[27]．また大腿四頭筋力と膝OAの進行との関係については，これまでのところ有意な関連性を示した基礎研究は見当たらず，大腿四頭筋力強化により疼痛やADLが改善したという臨床研究がみられるのみである．

このように，現在臨床の場において大腿四頭筋力強化が膝OAの予防や治療として有効であると推奨されているが，そのエビデンスは意外に乏しく，今後明らかにすべき多くの課題が残されている．

9 下肢アライメント，スラスト運動

生体力学的研究から膝内反により膝関節内側の荷重負荷が増大することが証明されているが，近年，膝内反アライメントと内側型膝OAの関連性がわれわれ[27]やSharmaら[28]により疫学調査や臨床研究から示されている．また，スラスト運動は立脚歩行初期における急激な内反運動で，われわれは歩行解析を行った膝OA患者にスラスト運動が多くみられたことから膝OAの有力なリスクファクターと考えている．松代膝検診でも膝OAの進行とともにスラスト運動の出現が増加し，さらに縦断解析によりスラスト運動が膝OA発症に関与していることが明らかとなった（図2）．欧米ではスラスト運動は膝内反モーメントとして評価され，歩行解析を用いた臨床研究から膝OAとの関連性が指摘されているが，近年，疫学研究においてスラスト運動と膝OAとの関連性を述べた報告もみられる[29,30]．

10 骨粗鬆症

従来，変形性関節症と骨粗鬆症は逆の病態と考えられ，膝OAについても高骨密度との関連性を示した報告が多い[31~33]．しかしその一方で，高骨密度は膝OA発症に関与するが，膝OAの進行には逆に低骨密度が影響するという報告[34]や，脛骨近位の内反変形と腰椎骨密度と相関するという研究[35]もみられる．また，動物