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Statistical modelling of knee valgus during a continuous jump test

Y. NAGANO[1], M. SAKAGAMI[1], H. IDA[2], M. AKAI[3], & T. FUKUBAYASHI[4]

[1]*Graduate School of Sports Science, Waseda University, Saitama, Japan,* [2]*Kanagawa Institute of Technology, Kanagawa,* [3]*National Rehabilitation Center for Persons with Disabilities, Saitama, Japan, and* [4]*Faculty of Sports Sciences, Waseda University, Saitama, Japan*

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Abstract
Landing with the knee in a valgus position is recognized as a risk factor for anterior cruciate ligament (ACL) injury. Using linear and non-linear regression analyses, the purpose of this study was to examine the correlation between two-dimensional (2D) knee valgus and three-dimensional (3D) knee kinematics measured during a jump landing task. Twenty-eight female collegiate athletes participated. All participants were required to perform a continuous jump test. The average maximum angles of abduction and internal tibial rotation during landing were measured using the Point Cluster Technique. Average peak knee valgus angle was measured using a 2D approach. Linear and non-linear regression analyses between 2D valgus and 3D knee abduction, and between 2D valgus and 3D internal tibial rotation, were performed. The R^2 value between 2D valgus and 3D knee abduction was significantly different from zero and had a moderate correlation for all models, whereas the R^2 value between 2D valgus and 3D internal tibial rotation was not significantly different from zero. The 2D approach could be used to screen a specific group of individuals for risk of ACL injury; however, using frontal plane 2D analysis of valgus motion to evaluate internal tibial rotation is not advised.

Keywords: *Injury, kinematics, knee, landing, motion analysis*

Introduction

Acute knee injuries, especially those to the anterior cruciate ligament (ACL), often occur during landing. Krosshaug and colleagues (2007) examined the mechanism of ACL injuries that occur while playing basketball and found that the most common manoeuvre at the time of injury is landing. Accordingly, many studies have focused on landing to examine the mechanism and risk of ACL injury.

Recently, knee valgus at the time of landing has been recognized as a risk factor of ACL injury. In a prospective study (Hewett et al., 2005), female athletes with increased dynamic valgus and a high abduction load during a jump-landing task were at increased risk of ACL injury. In a biomechanical study, valgus torque and internal tibial rotation in combination with anterior force resulted in a significantly larger strain to the ACL (Berns et al., 1992).

Correspondence: Y. Nagano, Sports Orthopedic Laboratory, Graduate School of Sports Science, Waseda University, Mikajima 2-579-15, Tokorozawa, Saitama 359-1192, Japan. E-mail: yasu-nagano@moegi.waseda.jp

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Using a model-based investigation to examine injury causing kinematics, McLean et al. (2004) found that valgus loading is a likely injury mechanism, especially in females. Several researchers have reported that females tend to land and remain in a more valgus position than males (McLean et al., 1999; Malinzak et al., 2001; Ford et al., 2003). In programmes designed to help prevent ACL injury (Hewett et al., 2001; Myklebust et al., 2003; Mandelbaum et al., 2005), athletes are instructed to avoid knee valgus during landing. Consequently, it is important to determine the amount of knee valgus during athletic tasks to screen individuals at risk of injury as well as to evaluate prevention programmes.

In biomechanical studies, three-dimensional (3D) motion analysis has been considered the standard method to measure the valgus angle of the knee during athletic tasks. Although this method provides reliable data, 3D motion analysis has spatial and temporal costs that prevent large screenings and evaluations to determine successful ACL injury prevention programmes. On the other hand, measuring frontal plane knee motion with a two-dimensional (2D) approach using a standard video camera is a concise and versatile procedure. Recently, some studies have successfully used the 2D approach (Barber-Westin et al., 2005; Noyes et al., 2005; Willson et al., 2006). However, the valgus angle of the knee evaluated using a 2D approach is influenced by hip internal rotation, ankle eversion, and knee flexion. Moreover, knee valgus evaluated using a 2D approach includes not only knee abduction, but also tibial rotation.

Using linear regression analysis, McLean et al. (2005) reported the potential of the 2D approach for screening knee valgus. While McLean et al. studied tasks that have high demands on the frontal plane (e.g. the side jump and sidestep), many researchers have examined landing tasks that have a high demand on the sagittal plane (e.g. the drop vertical jump, drop landings, etc.) (Hewett et al., 2005; Noyes et al., 2005; Yu et al., 2005). To our knowledge, no study has performed regression analyses using both linear and non-linear methods between 2D knee valgus and 3D knee kinematics data obtained during tasks that involve large movements in the sattigal plane. Most jump landing tasks are a one-shot trial with the possibility that the landing motion would include trial bias and feed-forward control. In contrast, individuals perform the continuous jump test without stopping after each landing task. Based on this characteristic, trial bias and feed-forward control should have less influence on the landing motion than one-shot tasks.

The aims of this study were to examine the regression between 2D knee valgus and 3D knee kinematics (both knee abduction and internal tibial rotation) and to determine which statistical model – linear, quadratic or logarithmic – best describes the 3D knee kinematics measured during the continuous jump test. We hypothesized that there would be a significant correlation between 2D knee valgus and 3D knee abduction in all regression models and the non-linear model would better describe the 3D kinematics than the linear model. Furthermore, since knee valgus rotation is a movement that occurs mostly in the frontal plane and tibial rotation is a movement that occurs mostly in the horizontal plane, we assumed it would be difficult for a 2D approach using kinematics from the frontal plane to evaluate movement in the horizontal plane. Therefore, we further hypothesized that there would not be a significant correlation between 2D knee valgus and 3D internal tibial rotation for all models.

Methods

Participants

Twenty-eight female collegiate basketball and lacrosse athletes gave their written informed consent to participate in the study. Approval for the study was obtained from the institutional review board of the National Rehabilitation Center for Persons with Disabilities. Exclusion criteria included a history of lower limb injury and or any musculoskeletal injury in the previous 6 months that prohibited an individual from playing sports. The mean physical characteristics of the participants were as follows: age 21 ± 1 years, height 1.66 ± 0.8 m, and body mass 58.8 ± 7.7 kg.

Continuous jump test

All testing took place at the National Rehabilitation Center for Persons with Disabilities in Saitama, Japan. The participants were measured in a static standing position. While barefoot, the participants performed five vertical jumps continuously (i.e. without resting between jumps) using both legs with maximum effort (Figure 1). They were instructed to place their hands on their lower torso, stand with their feet shoulder-width apart, and face the frontal plane during testing. A research assistant demonstrated the continuous jump test; however, the assistant did not provide any verbal instructions regarding landing or jumping technique. The participants were allowed several preparation trials. Measurement of the landing of the dominant limb from the second to the fourth jump was used for analysis.

Analysis of the 3D data

A six-camera motion analysis system (Motion Analysis Corp., California, USA) was used to record the 3D movements of the lower limb. The motion and force data were recorded at 200 Hz. For each participant, 24 reflective markers of 9 mm diameter were secured to the lower limb using double-sided adhesive tape as described previously (Nagano et al., 2007).

From the coordinate system data obtained, the angular displacements of knee abduction/adduction and internal/external tibial rotation were calculated for each landing. The knee kinematics were calculated using the Point Cluster Technique (Andriacchi et al., 1998) and the Joint Coordinate System proposed by Grood and Suntay (1983). For this algorithm, the reference zero position for these measurements was obtained during the static standing trial. The angular displacements in each trial were indicated as a variation from the position in the static standing trial. The average maximum angle during landing from the static standing position was measured as knee abduction and internal tibial rotation.

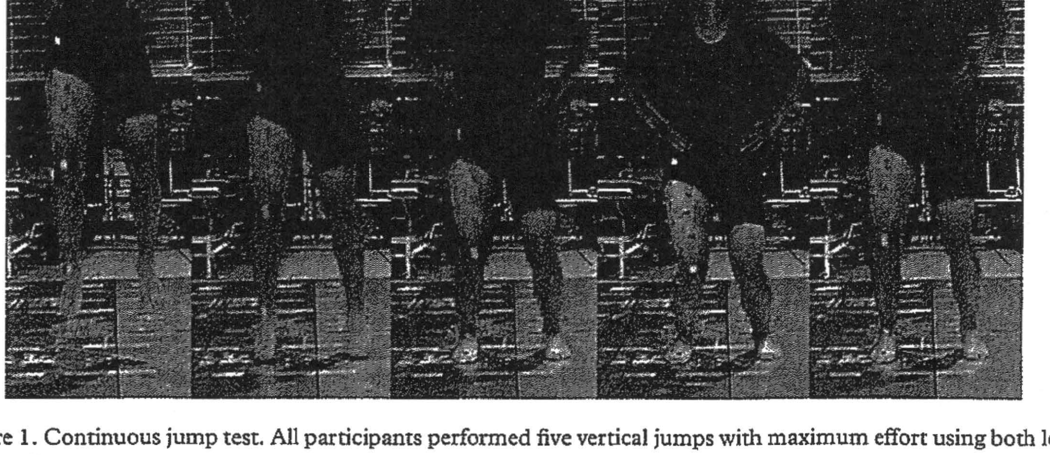

Figure 1. Continuous jump test. All participants performed five vertical jumps with maximum effort using both legs and landing.

Analysis of the 2D data

Each trial was recorded from the frontal plane using a digital video camera (30 Hz; Sony Product, Japan). The camera was placed 3.8 m from the landing point at the height of the knee joint. For each participant, 3 square plastic tape markers with an area of 3.24 cm^2 were secured to the lower limb. Markers were placed at the anterior superior iliac spine (ASIS), the midpoint of the patella, and the midpoint of the ankle joint. Captured images were imported into a digitizing software program (Dartfish, Dartfish Japan Co., Ltd., Japan). The angle between the lines formed from the marker on the ASIS to the midpoint of the patella and that formed from the midpoint of the patella to the midpoint of the ankle joint was recorded as the knee valgus angle (Figure 2). The average peak 2D knee valgus angle from the static position was measured for analysis. As the purpose of this study was to screen for knee valgus, when participants showed knee varus during landing, these data were excluded from analysis.

Statistical analyses

Linear and non-linear regression analyses between 2D valgus and 3D knee abduction, and between 2D valgus and 3D internal tibial rotation, were performed to identify the model with the best fit. The models were expressed by the following equations:

Linear model: $y = a + bx$
Quadratic model: $y = a + bx + cx^2$
Logarithmic model: $y = a + b\ln(x)$

The coefficient of determination (R^2 value) for each model was calculated and tested for statistical significance. When the R^2 value for all three models was significant, an analysis

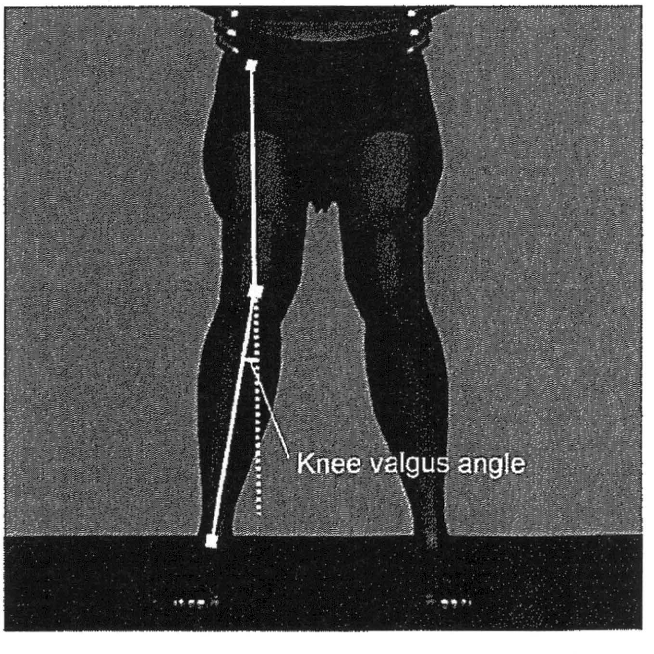

Figure 2. Measurement of knee valgus using the 2D method. The angle between the line formed from the marker on the anterior superior iliac spine (ASIS) to the midpoint of the patella and that formed from the midpoint of the patella to the midpoint of the ankle joint was recorded as the knee valgus angle.

of variance (ANOVA) together with a *post-hoc* LSD test was conducted to investigate the effect of each model (linear, quadratic or logarithmic) on the R^2 value. These statistical analyses were referred to in a previous study of similar design (Coorevits et al., 2005) and conducted using the statistical software package SPSS (v. 11.0, SPSS Inc., Chicago, IL). Statistical significance was set at $P < 0.05$.

Based on the R^2 value calculated from a pilot study (0.36) and the power of 0.80, a sample size calculation revealed that 19 participants were required to have sufficient power to test the regression analysis.

Results

When measuring 2D valgus during the continuous jump test, eight participants showed knee varus. Therefore, the data for only 20 participants were analysed in this study. No significant differences were observed in age, height or body mass between those participants included in and excluded from the analysis. For the test–retest trial, the intraclass correlation for the 2D valgus was 0.73, demonstrating substantial reliability of the videographic test and software capturing procedures.

For all models, the R^2 value between 2D valgus and 3D knee abduction was significantly different from zero: linear model ($R^2 = 0.34$, $P < 0.01$) (Figure 3A), quadratic model ($R^2 = 0.40$, $P = 0.01$) (Figure 3B), and logarithmic model ($R^2 = 0.41$, $P < 0.01$) (Figure 3C). Regarding the results of the ANOVA tests, no significant differences were observed between three models based on the R^2 values. For all models, the R^2 value between 2D valgus and 3D internal tibial rotation was not significantly different from zero (Figure 4).

Discussion and implications

The present study used regression analysis to examine the potential of a 2D approach using a standard video camera to evaluate 3D kinematics. We also examined the best fit statistical model to describe 3D kinematics. By developing a regression relationship between 2D valgus and 3D knee kinematics, a 2D approach was able to be used to screen participants at risk for ACL injury as well as to evaluate prevention training programmes that attempt to reduce ACL injury rates. Additionally, researchers, coaches, and trainers should be able to conduct adequate evaluation without having to use a complicated 3D approach, since the continuous jump test procedure has substantial reliability and requires simple equipment.

The results of this study showed that there was a moderate correlation between 2D valgus and 3D knee abduction in all regression models. McLean et al. (2005) used a linear regression analysis to show a correlation between 2D analysis of valgus and 3D analysis of valgus during frontal plane athletic tasks. The R^2 values in the present study were significant, but lower than those reported by McLean and colleagues. When the hip joint is internally rotated, the knee flexion angle is projected onto the frontal plane as a 2D valgus angle. Stance width and ankle eversion/inversion also contribute to the 2D valgus angle. The jump-landing task examined in the present study is more easily influenced by these other factors than the tasks examined in McLean's study, which have a high demand on the frontal plane. Therefore, the correlation for this study would naturally be lower.

To determine whether the 2D valgus angles measured during the continuous jump test can be used to screen for individuals who are at risk for ACL injury, the following error analysis was performed. In the correlation plots of this study, the maximum residual error was 7.03°, 6.76°, and 6.60° for the linear, quadratic, and logarithmic model, respectively.

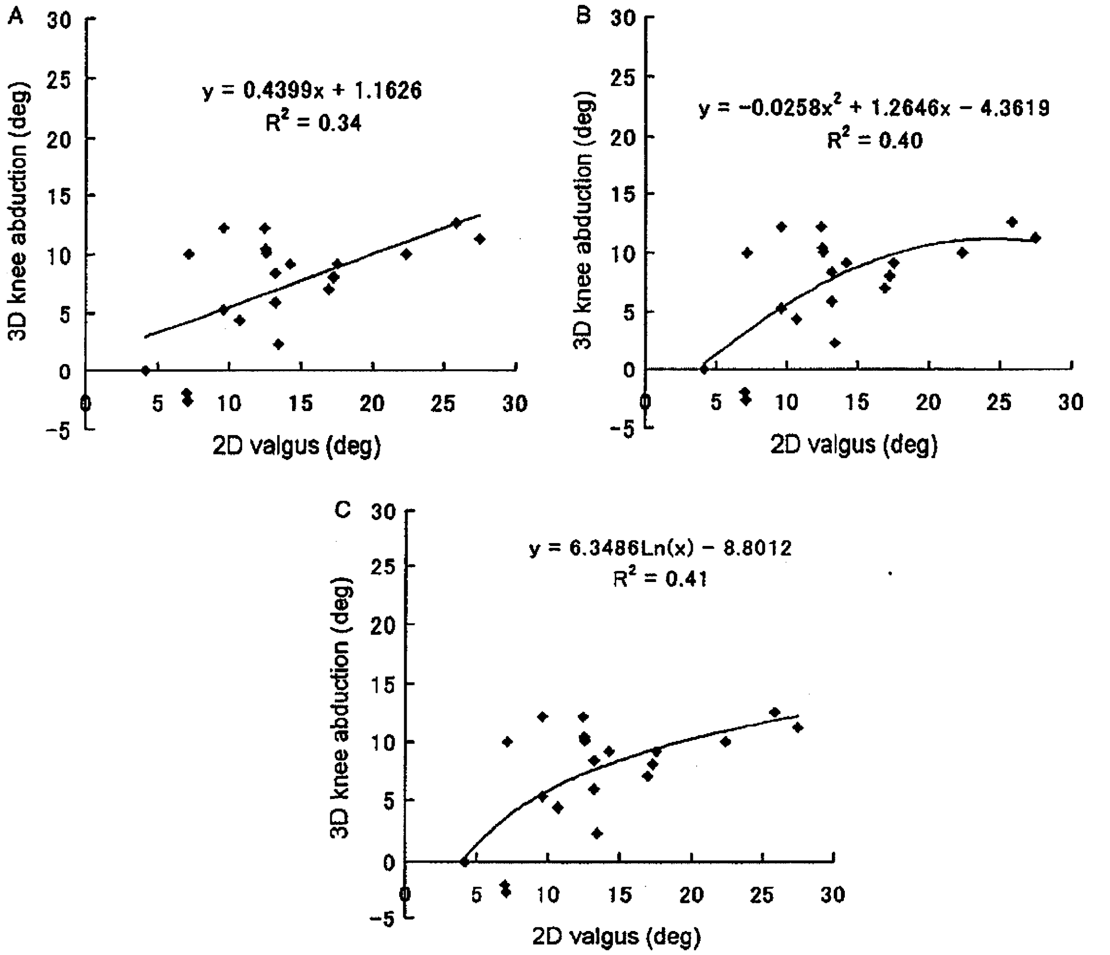

Figure 3. Associations between 2D valgus and 3D knee abduction during the continuous jump test for the linear model (A), the quadratic model (B), and the logarithmic model (C). The R^2 values of all models between 2D valgus and 3D knee abduction were significantly different from zero.

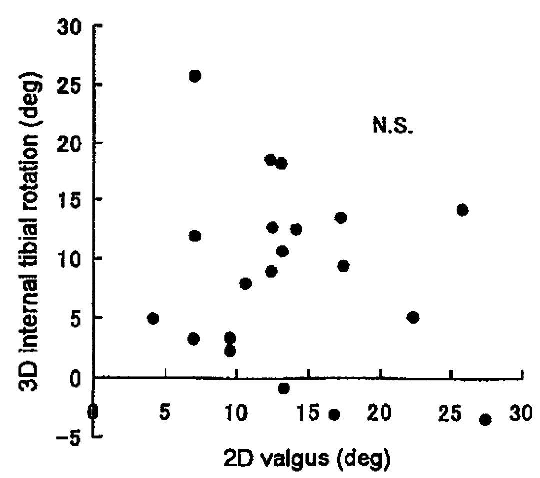

Figure 4. Associations between 2D valgus and 3D internal tibial rotation during the continuous jump test. The R^2 values of all models between 2D valgus and 3D internal tibial rotation were not significantly different from zero.

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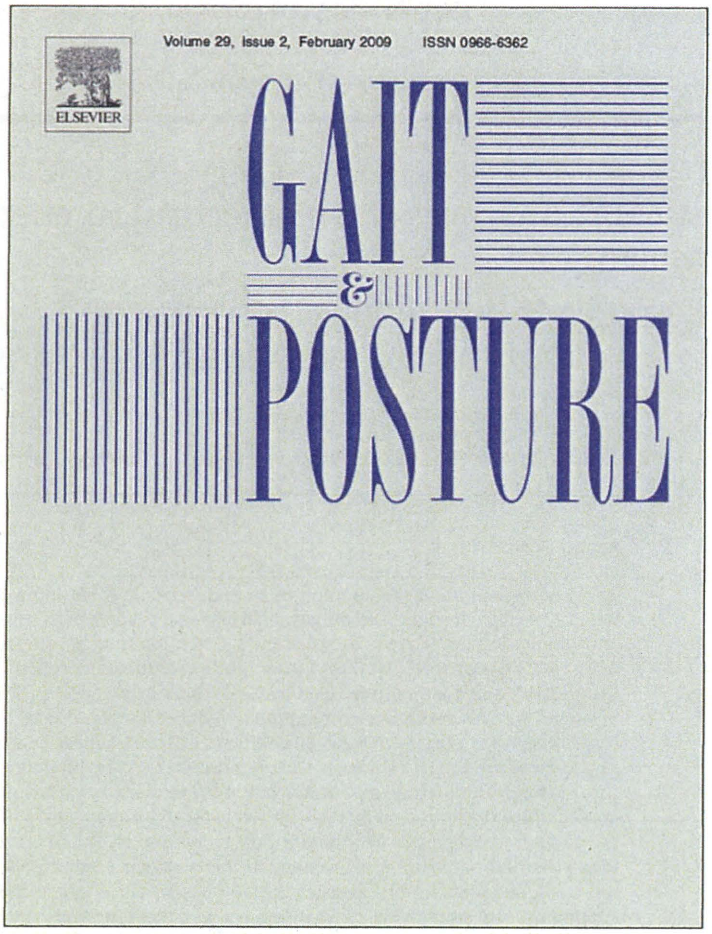
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Addition of an arch support improves the biomechanical effect of a laterally wedged insole

Kohei Nakajima [a,b], Wataru Kakihana [c], Takumi Nakagawa [a], Hiroyuki Mitomi [b], Atsuhiko Hikita [b], Ryuji Suzuki [b], Masami Akai [d], Tsutomu Iwaya [d], Kozo Nakamura [a], Naoshi Fukui [b,*]

[a] Department of Orthopaedic Surgery, University of Tokyo, Bunkyo-ku, Tokyo, Japan
[b] Clinical Research Center, National Hospital Organization Sagamihara Hospital, Sagamihara, Kanagawa, Japan
[c] Department of Nursing, Ishikawa Prefectural Nursing University, Kahoku, Ishikawa, Japan
[d] Department of Rehabilitation for Movement Functions, National Rehabilitation Center for Persons with Disabilities, Tokorozawa, Saitama, Japan

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Gait analysis

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

* Corresponding author at: Department of Etiology, Clinical Research Center, National Hospital Organization Sagamihara Hospital, Sakuradai 18-1, Sagamihara, Kanagawa 228-8522, Japan. Tel.: +81 42 742 8311; fax: +81 42 742 7990.
E-mail address: n-fukui@sagamihara-hosp.gr.jp (N. Fukui).

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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

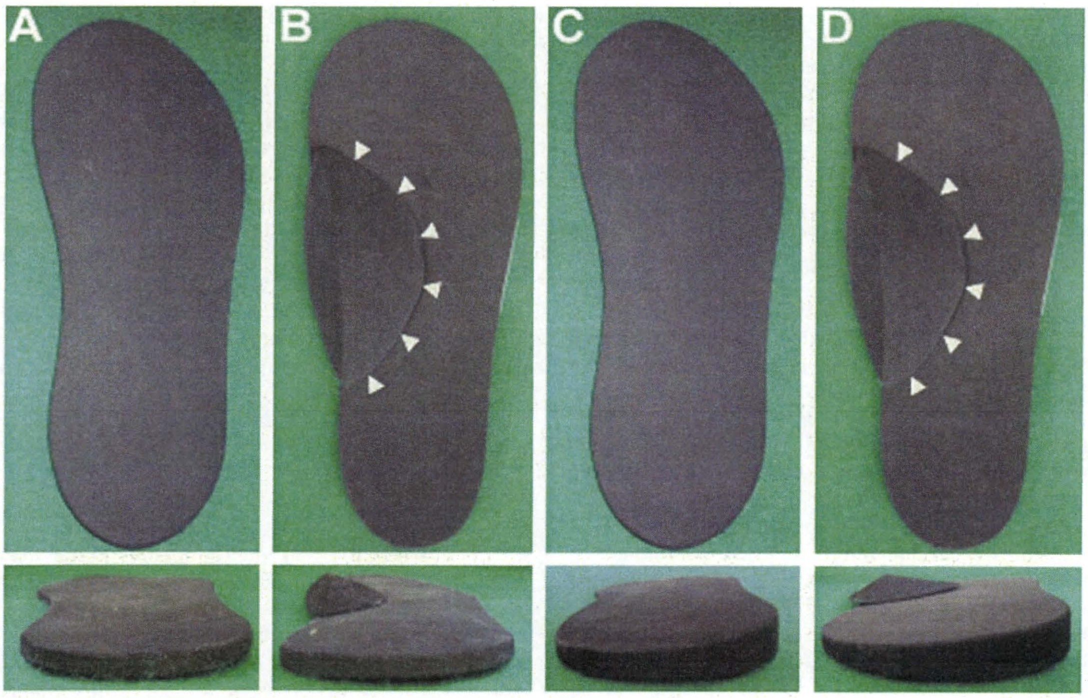

Fig. 1. Insoles used for this study. Upper (upper panels) and posterior views (lower panels) of FLAT (A), AS (B), LW (C), and LWAS (D) are shown. In (B) and (D), white arrowheads indicate arch supports added to the insoles.

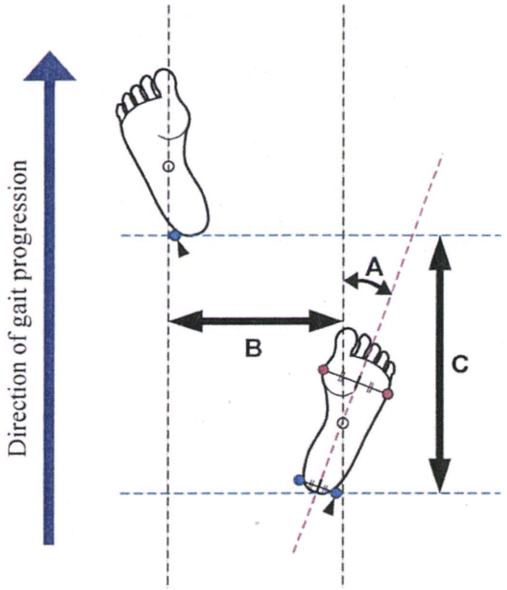

Fig. 2. Three kinematic parameters evaluated. Progression angle (A) is the angle formed by the direction of gait progression and the foot axis, which is an assumptive line passing through the middle of the first and fifth metatarsal heads and the middle of the calcaneal tuberosity at the midstance (red 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

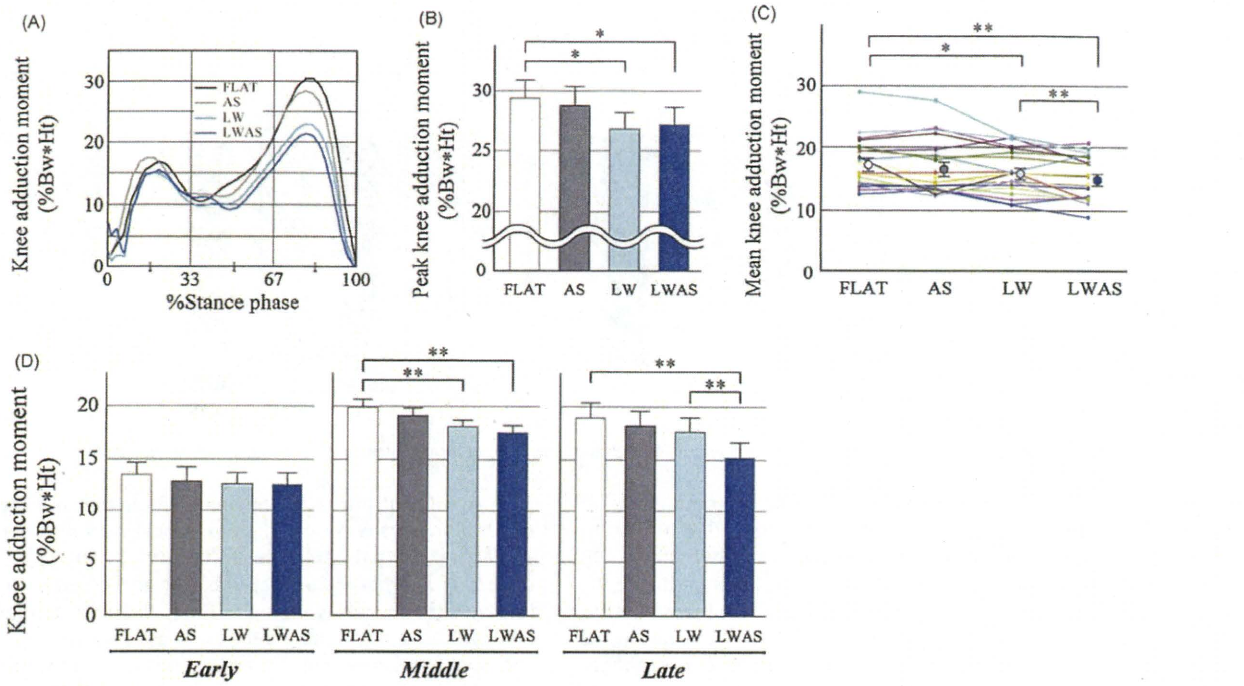

Fig. 3. External adduction moment of the knee joint with the four types of insoles. (A) Knee adduction moment with the four types of insoles during the stance phase. Representative result of a single subject is shown. Data are averages of five measurements. (B) Peak values of knee adduction moment with 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

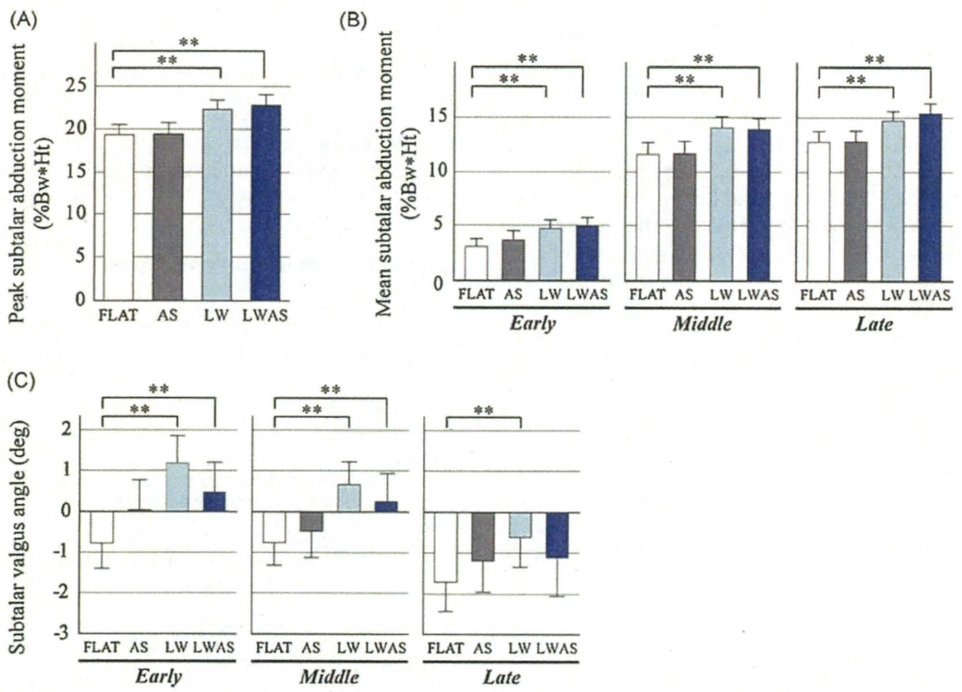

Fig. 4. Abduction moment and valgus angle at a subtalar joint with the four types of insoles. (A) Peak values of abduction moment with respective insoles. (B) Abduction moment averaged in respective sections of the stance phase. (C) Valgus angle averaged in respective sections of the stance phase. Values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

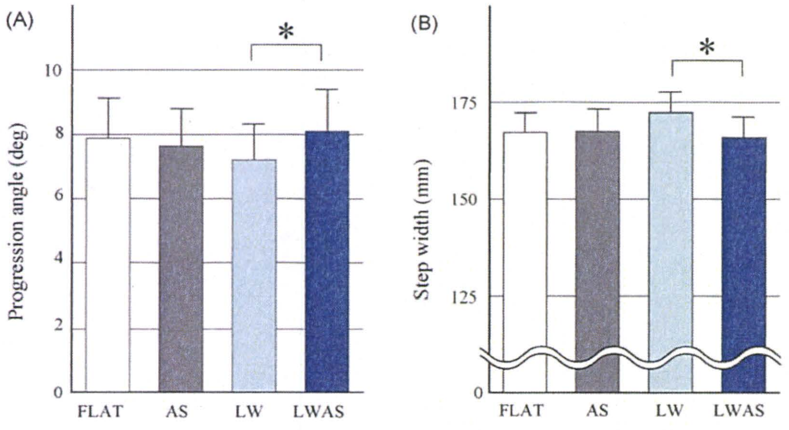

Fig. 5. Progression angle (A) and step width (B) with the four types of insoles. Values are the mean + S.E.M. $^{*}p < 0.05$.

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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The Knee 16 (2009) 153–158

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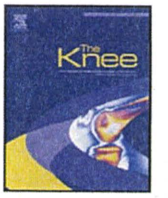

Biomechanical characteristics of the knee joint in female athletes during tasks associated with anterior cruciate ligament injury

Yasuharu Nagano [a,*], Hirofumi Ida [b], Masami Akai [c], Toru Fukubayashi [d]

[a] *Waseda University Graduate School of Sports Sciences, Saitama, Japan*
[b] *Kanagawa Institute of Technology, Kanagawa, Japan*
[c] *National Rehabilitation Center for Persons with Disabilities, Saitama, Japan*
[d] *Waseda University Faculty of Sports Sciences, Saitama, Japan*

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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

* Corresponding author. Yasuharu Nagano, Waseda University Graduate School of Sports Sciences, Sports Orthopedic Lab Mikajima 2-579-15, Tokorozawa, Saitama 359-1192, Japan. Tel.: +81 4 2947 7031; fax: +81 4 2947 6879.
E-mail address: ynema@f2.dion.ne.jp (Y. Nagano).

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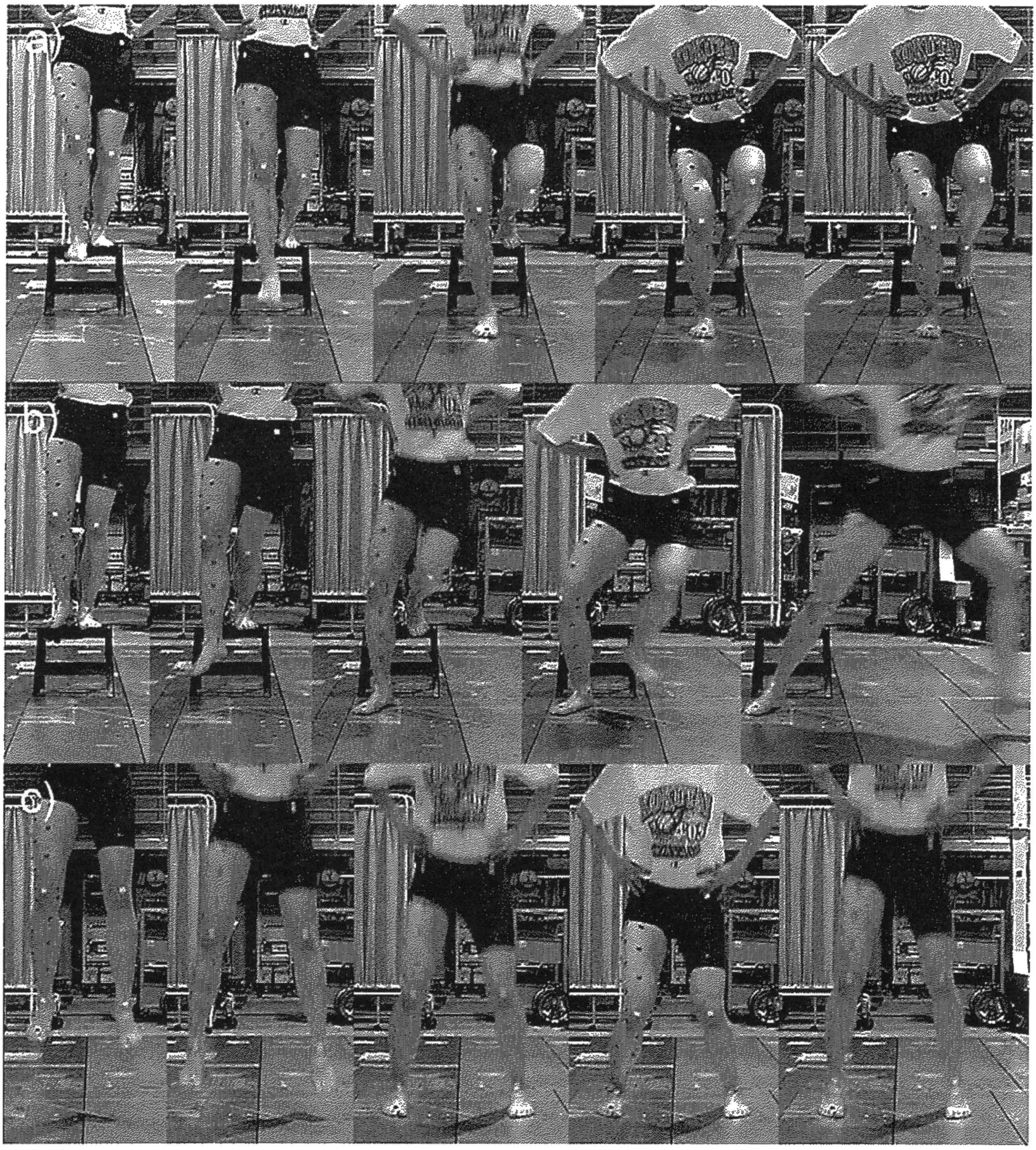

Fig. 1. Sequential photographs of experimental tasks: Single-limb landing (a), plant and cutting (b), and both-limb jump landing.

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-

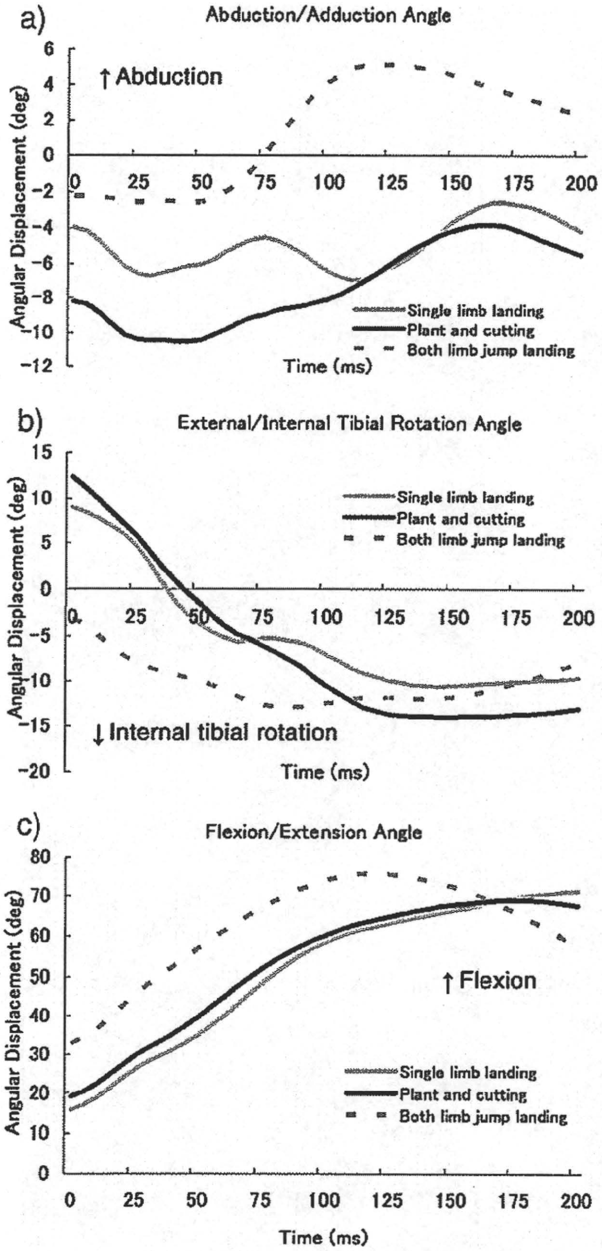

Fig 2. Comparisons of joint motion. Data are presented for knee abduction/adduction (a), external/internal tibial rotation (b), and knee flexion/extension (c).

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

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