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【Original investigation】

Gender difference in neuromuscular hip and knee joint control during single-leg landing.

Issei Ogasawara[1], Shumpei Miyakawa[2], Shigeyuki Wakitani[1]

Abstract

Background: Female's altered lower limb control has been considered as a possible factor for their high anterior cruciate ligament injury rate. However, the detailed gender difference in lower limb control during impact activity is not well studied. The purpose of this study was to investigate the gender difference in neuromuscular hip and knee joint control during single-leg landing motion. It was hypothesized that the male and female subjects show different electromyographic activity patterns for controlling hip and knee motion and female characteristics would elevate their risk for anterior cruciate ligament injury.

Methods: Ten male and 8 female subjects took part in this study. Electromyographic activities of quadriceps, hamstrings and gluteus medius, and three-dimensional kinematic data of hip and knee were measured during single-leg landing task. Peak activities and timing of peak activities of each muscle were compared between genders. Gender differences in the hip and knee kinematics were also investigated.

Results: Female subjects showed significantly greater peak activities in vastus medialis, vastus lateralis and gluteus medius compared to male subjects ($p < 0.05$). Semitendinosus and biceps femoris in females peaked significantly earlier compared to male ST, BF and female quadriceps (ST; $p < 0.05$, BF; $p < 0.05$). Female subjects simultaneously exhibited greater knee valgus ($p < 0.05$) and hip adduction ($p < 0.05$).

Conclusion: Higher vastus medialis activities found in females were considered a strategy to resist knee valgus motion. However, higher vastus medialis activities may cause tibial anterior shear force but resist knee valgus unless synchronized hamstrings activities. Females showed greater hip adduction despite of high gluteus medius activities. This suggests that hip muscle weakness leads to failed hip control.

Keywords： neuromuscular control; hip; knee; anterior cruciate ligament

1) Mukogawa Women's University, Department of Health and Sports Sciences. *6-46 Ikebiraki, Nishinomiya, Hyogo 663-8558, JAPAN.*

2) University of Tsukuba, Comprehensive Human Sciences. *1-1-1 Tennodai, Tsukuba, Ibaraki 305-8577 JAPAN.*

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Ⅰ. Introduction

Anterior cruciate ligament (ACL) injury is a common and serious athletic injury. If performance level remains decreased, athletes will require surgical treatment and long-term rehabilitation to regain previous competitive level. Studies have repeatedly demonstrated that female athletes exhibit a 2- to 8-fold greater likelihood of ACL injury compared to their male counterparts[1, 2], clearly representing a serious issue for female athletes.

Several factors such as anthropometrics[3] and hormonal factors[4] have been reported as reasons for this high injury rate in females. Various risk factors have been investigated, and decreased neuromuscular control of females has recently become a great concern as a possible factor elevating the incidence of ACL injury. This decade, studies aimed at characterizing dynamic neuromuscular control in females by comparing gender differences in lower limb kinematics and/or muscular activities during athletic tasks have been conducted, resulting in beneficial insights[5-10]. Cowling et al.[5] examined gender differences in lower limb muscle activities for single-leg landing with regard to the temporal pattern of muscle activities. In that research, female subjects exhibited less-synchronized hamstring muscle activity with respect to time of foot contact, suggesting that females could not utilize the hamstrings effectively due to decreased muscular control. Malinzak et al.[6] compared thigh muscle activities between genders and found that females often jump and land with much higher quadriceps activities and smaller angles of knee flexion compared to males. Increased quadriceps activity with small angle of knee flexion has been considered as an inappropriate muscle activity because excessive quadriceps contraction has the potential to induce anterior tibial translation. This is one of the typical mechanisms working in the sagittal plane to increase stress on the ACL. These reports have thus revealed that female athletes usually conduct athletic activities under conditions of inadequate muscular control.

Hip motion also influences knee kinematics. Video analysis of non-contact ACL injuries has demonstrated that abnormal hip joint motions such as excessive hip adduction and/or internal rotation are concomitantly observed at the moment of valgus knee collapse[11, 12]. Laboratory controlled studies have demonstrated that dynamic hip adduction and hip internal rotation, which are similar to the actual injury posture, are often seen in female populations[10, 13]. These facts indicate that female athletes may tend to exhibit abnormal hip positions such as adduction and internal rotation, indirectly placing the knee joint toward the midline of the body and increasing knee valgus angle. To clarify the reasons why females exhibit such risk-elevating limb positions, various studies have been conducted. Zazulak et al.[9] examined hip muscular control during single-leg landing, and found lower gluteus maximus (GMAX) activities in female subjects and concomitant dynamic hip adduction during landing. Zeller et al.[10] also investigated gender differences in gluteal muscle activities using a single-leg squat task, and although no significant gender differences were found in their study, females exhibited a tendency toward higher GMAX and lower gluteus medius (GMED) activities. While showing conflicting results, these two reports showed gender characteristics of hip muscle activity which may affect hip motion. Additionally, dynamic knee valgus was observed in both studies; this finding may suggest that the knee alignment was affected by hip joint control. Successful control of hip joint is very important for proper knee alignment. The knee joint has small range of motion (ROM) in the frontal plane. Co-contraction of knee extensor/flexor muscles help to stabilize the joint by increasing joint stiffness; however, there are limited muscular options that directly resist valgus knee stress. The hip joint is much less-constrained but has musculature to control frontal plane motion, thus it can indirectly control the frontal plane knee alignment in weight acceptance phase. Thus

acquisition of a proper hip joint control is quite important for dynamic knee control.

The importance of hip function is becoming recognized, but the literature remains limited. In addition, due to conflicting results among limited studies, reasonable agreement has not been obtained on how female characteristic of hip control influences the knee joint biomechanics. Furthermore, specifically focusing on the frontal plane, the relationship between hip/knee dynamic alignment and muscular control of females has not been well-studied. The purpose of this study was thus to clarify the female characteristics of hip and knee joint control through evaluating gender differences in lower limb kinematics in the frontal plane and electromyographic (EMG) activities of knee extensor/flexors and hip abductors.

Ⅱ. Methods

A. Subjects

Subjects comprised 10 healthy male subjects (mean age, 24.8 ± 4.3 years; height, 172.7 ± 6.5 cm; weight, 70.9 ± 9.1 kg) and 8 healthy female subjects (mean age, 23.0±1.0 years; height, 161.3 ±4.2 cm; weight, 53.1±7.4 kg). The subjects did not participate in the any kind of competitive sports or trainings during the research period, and no history of lower extremity injury or surgery was present in any subject. Physical characteristics of participants are shown in Table 1. Prior to participation, written informed consent was obtained from all subjects and all study protocols were approved by the local ethics committee.

Table 1 . Physical properties of subjects (n=18)

Characteristics	Male n=10(Mean±SD)	Female n=8(Mean±SD)	*p* value
Age (y.o.)	24.8 ± 4.3	23.0 ± 1.0	0.15
Height (cm)	172.7 ± 6.5	161.3 ± 4.2	<0.05*
Weight (kg)	70.9 ± 9.1	53.1 ± 7.4	<0.05*

*denotes statistically significant difference between men and women

B. Experimental Settings

The dominant leg of each subject was determined to be the leg that the subject usually uses to kick a ball for the consistency with the previous studies 14-16. For the preparation of motion capture, 10 markers were attached bilaterally to the anterior superior iliac spine (ASIS) and great trochanter (GT), and to the dominant-side patella, medial and lateral femoral epicondyles, medial and lateral malleoli and great toe (Fig. 1). All markers were placed by the same examiner to maintain consistency.

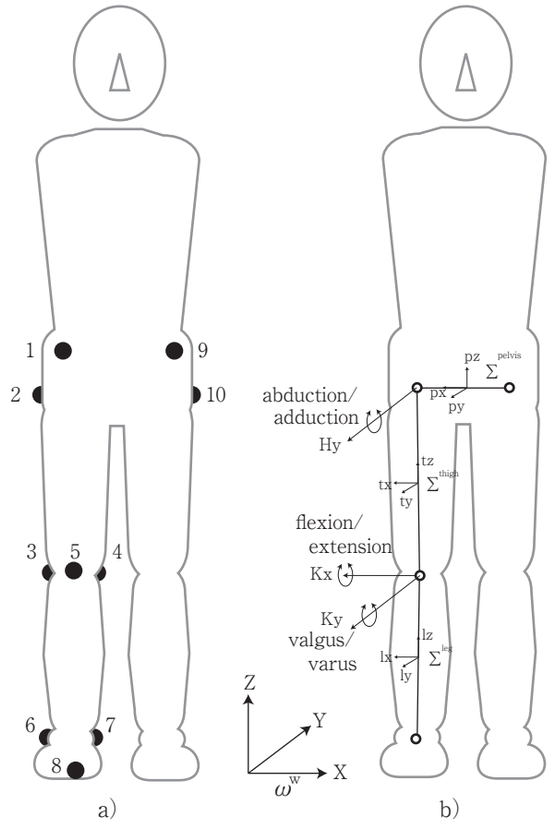

Figure 1 : a) Marker positions to obtain 3-dimensional position data for pelvis, thigh and leg segments.
b) Based on these position data, local coordinate systems Σ^{pelvis}, Σ^{thigh} and Σ^{leg} were attached to calculate kinematic values.

Surface EMG activities of 5 muscles, i.e., vastus medialis (VM), vastus lateralis (VL), semitendinosus (ST), biceps femoris (BF) and GMED, on the dominant side of each subject were recorded using Ag-AgCl bipolar surface electrodes (NT-511G; Nihon Kohden, Tokyo, Japan). EMG electrodes were placed parallel to the direction of muscle fibers

with spacing of approximately 20 mm. A reference electrode was placed on the sternum. Electrode locations were determined by palpating the muscles to maximize the signal from a particular muscle while subjects were contracting their muscles. To decrease skin impedance, the surface of the skin was shaved, scrubbed using skin preparation gel (YZ-0019; Nihon Kohden) to exfoliate epidermal debris and cleaned with alcohol. To minimize the influence of movement artifacts, the electrode cables were firmly taped down to the subjects' body. Surface EMG signals were amplified using a multi-telemeter (WEB-5000; Nihon Kohden) and sampled by an A/D converter (MP100WSW; Biopac Systems, Goleta, USA) at a sampling frequency of 1,000 Hz for each muscle.

C. Task Protocol

1. Maximal voluntary contraction (MVC) trials

After application of surface electrodes, MVC of each muscle was acquired. Subjects were asked to sit in a chair with hips at 90° flexion and knees at 60° flexion during MVC tests for the VM and VL. For ST and BF, subjects took a prone position on a bed with knees at 60° flexion and hip joint at neutral position. For GMED, subjects took a side lying position on a bed with 0° hip adduction/abduction. EMG was recorded for 5 s on 3 occasions, with verbal encouragement.

2. Landing Task

Subjects were required to perform a single-leg landing task from a 32-cm high platform using the dominant leg. Motion in each landing trial was captured using 3 digital video cameras (DCR-TRV17; Sony, Tokyo, Japan) at 60 Hz in synchronization with EMG recording using a custom-made device. If subject could not stay in single leg stance and utilized the contralateral leg just after landing, that trial was considered a failure. Additionally, if more than two of three video cameras could not capture the ASIS and GT markers, by torso rotation or inclination, that trial was also regarded as failure. Based on these criteria, we acquired 10 successful trials from each subjects. Video data of each trial were converted into AVI format file using Premiere Pro 1.5 software (Adobe Systems, San Jose, USA) and stored on a personal computer.

D. Data analysis

To remove movement artifacts, raw EMG signals were filtered using a high-pass fourth-order zero-lag Butterworth filter with cut-off frequency of 10 Hz. After full-wave rectification, signals were smoothed using a low-pass Butterworth filter (fourth-order, zero-lag, 10 Hz cut-off) to create the linear envelope. Choice of cut-off frequencies was done by two of authors. Before processing the EMG data, we tested 8 patterns of different frequency settings (5Hz-40Hz, each 5Hz step). There were less-drastic changes from raw signal form and proper noise reduction, we decided to use 10 Hz for smoothing. Preprocessed EMG signals from each muscle recorded in landing trials were normalized by a reference value defined for each muscle, which was calculated based on EMG signals recorded in an MVC test as follows. First, the EMG signal with the largest maximum value was selected from signals obtained in the 3 MVC trials. The reference value was then defined as the average of 100 samples around the maximum point. We calculated the following 2 parameters from normalized EMG signals: 1) peak %MVC (peak value), representing the activation level of muscle activity with respect to MVC activity; and 2) time of peak EMG activity (peak time), to examine temporal profiles of muscle activities. In this analysis, time of initial foot contact was set as 0 ms.

The 2-dimensional (2D) trajectories of markers were traced in every video of each landing instant using FrameDias2 software (DKH, Tokyo, Japan). A trajectory in 3-dimensional (3D) coordinates was then constructed from three 2D trajectories using DLT methods, and filtered using a fourth-order low-pass Butterworth filter (cut-off frequency, 6 Hz). To calculate joint angles, we defined 3 segments (pelvis, thigh and leg) and 3 local coordi-

nate systems (Σ^{pelvis}, Σ^{thigh} and Σ^{leg}) (Fig. 1). Each local coordinate system has three orthogonal unit vectors, i.e., pelvis: px,py,pz, thigh: tx,ty,tz, leg: lx,ly,lz. Hy in Figure 1 represents the sagittal axis of the hip joint, which was a cross product of tz and px[17]. Kx shows the transverse axis of the knee joint, which was equal to tx. Ky indicates the sagittal axis of the knee joint. This axis was a cross product of the lz and tx[18]. Hip adduction/abduction was defined as thigh segment rotation occurring about the Hy. Knee flexion/extension was defined as leg segment rotation occurring about Kx. Knee valgus/varus was calculated as leg segment rotation occurring about Ky. Angles representing hip adduction, knee flexion and knee valgus were considered positive, with other angles considered negative.

E. Statistical Analysis

Two-way analysis of variance (ANOVA) (2 genders by 5 muscle differences) was performed to check the significant effect of genders and muscle difference for EMG peak time and peak value. When a main effect was noted, a post-hoc Tukey-Kramer test was conducted. To check the significant gender effect on kinematics data as a function of time, two-way ANOVA (2 genders by time) was used. All statistical analyses were performed using StatView version 5.0 software (SAS Institute, Cary, USA). In all cases, values of $p < 0.05$ were considered statistically significant.

Ⅲ. Results

A. EMG results, Peak time

Figure 2 shows the smoothed and normalized EMG signals observed 300 ms before and after initial foot contact for both genders. Asterisks denote the peak activity of each trial. The activities of VM and VL started before the initial foot contact and the peak activities concentrated after landing, but those activity levels quickly decreased after those peak activations. This trend for VM and VL was consistent among the genders. The activities of ST and BF also gradually increased before the initial foot contact, and those peak activities appeared before and after the contact. The peak activities of females' BF occurred relatively earlier than those of females' ST. GMED started to activate before the initial foot contact as well, and those activity level were maintained throughout the single-legged standing phase.

ANOVA revealed that the genders and muscle difference had a significant effect for EMG peak time (genders; $p < 0.05$, muscle difference; $p < 0.05$ respectively). There was no significant genders-by-muscle difference interaction (p=0.50). Post-hoc test showed that female ST and BF peaked significantly earlier compared to male ST and BF (ST; $p < 0.05$, BF; $p < 0.05$) (Fig. 3). There was no gender difference in peak time of VM, VL and GM. Table 2 shows a within-gender, inter-muscle comparison of peak time. Female ST and BF peaked significantly earlier compared to female VM, VL and GMED (ST vs VM; $p < 0.05$,

Table 2. Within-gender differences in mean peak time of %MVC during single-leg landing.

Gender	Peak time (ms) (Mean ± SD)				
	VM	VL	ST	BF	GMED
Male	115.9 ± 38.4[d]	105.3 ± 46.1[d]	64.2 ± 41.3	8.8 ± 45.1[a,b]	69.9 ± 49.5
Female	93.7 ± 38.6[c,d]	97.8 ± 42.0[c,d]	12.9 ± 48.7[a,b,e]	-32.8 ± 28.0[a,b,e]	56.8 ± 55.3[c,d]

a denotes significant difference vs VM ($p<0.05$)
b denotes significant difference vs VL ($p<0.05$)
c denotes significant difference vs ST ($p<0.05$)
d denotes significant difference vs BF ($p<0.05$)
e denotes significant difference vs GMED ($p<0.05$)

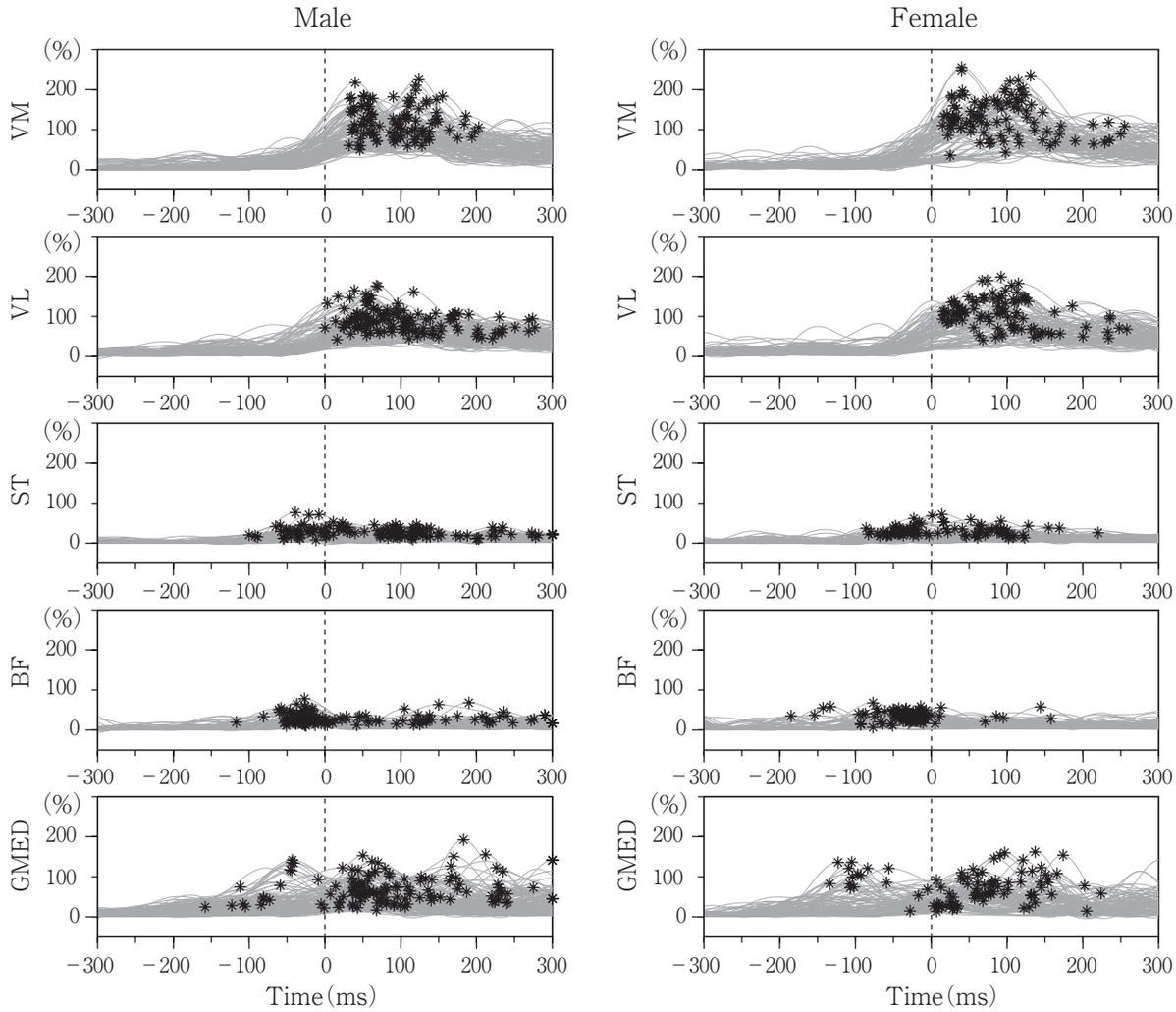

Figure 2 : Normalized EMG signals for each muscle from both genders. The asterisks denote the peak values of each signal. The vertical dashed line indicates the time of initial foot contact.

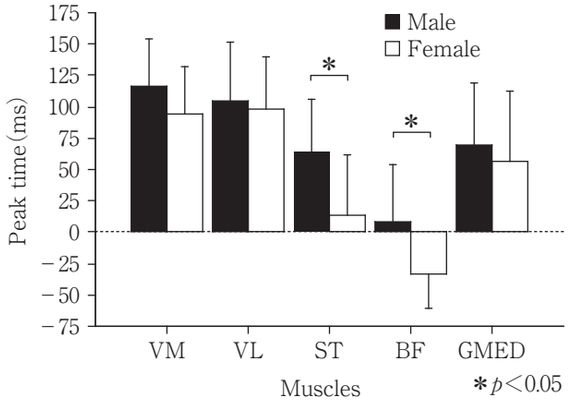

Figure 3 : Gender comparison of mean peak time of %MVC during single-leg landing. The transverse dashed line represents initial foot contact. The asterisk indicates a statistically significant difference between male and female subjects. In ST and BF, female subjects showed significantly earlier peak time compared to male subjects. In this study, the time of initial foot contact was defined as 0 ms.

ST vs VL; $p < 0.05$, ST vs GMED; $p < 0.05$, BF vs VM; $p < 0.05$, BF vs VL; $p < 0.05$, BF vs GMED; $p < 0.05$). In particular, female BF peaked before foot contact even though all the other muscles peaked after landing (Table 2). On the other hand, in male subjects, only BF peaked earlier than VM and VL did (BF vs VM; $p < 0.05$. BF vs VL; $p < 0.05$). A significant peak time difference between ST and the other muscles, observed in female ST, did not observed in male subjects.

B. Peak value

Genders and muscle difference had a significant effect for EMG peak value (gender; $p < 0.05$, muscle difference; $p < 0.05$). There was a significant

genders-by-muscle difference interaction ($p < 0.05$). Post-hoc test showed that female VM, VL and GM activities were significantly higher than those of male subjects (VM; $p < 0.05$, VL; $p < 0.05$, GM; $p < 0.05$) (Fig. 4). Hamstring muscle ST and BF in both genders activated to almost the same levels and there were no significant differences. Table 3 shows a within-gender, inter-muscle comparison of EMG peak value. Both male and female subjects showed in peak value as higher activities in knee extensor VM and VL, and lower activities in ST and BF, however, female VM showed especially higher activities and its peak value was significantly larger than all the other muscles (VM vs VL; $p < 0.05$, VM vs ST; $p < 0.05$, VM vs BF; $p < 0.05$, VM vs GM; $p < 0.05$). In male subject, VM and VL showed significantly higher activities than those of male ST, BF and GM (VM vs ST; $p < 0.05$, VM vs BF; $p < 0.05$, VM vs GMED; $p < 0.05$, VL vs ST; $p < 0.05$, VL vs BF; $p < 0.05$, VL vs GMED; $p < 0.05$), however, there was no significant difference between VM and VL peak value.

C. Kinematic results

A significant gender effect was found in frontal plane knee and hip kinematics ($p < 0.05$). Female subjects demonstrated significantly greater hip adduction and knee valgus motion ($p < 0.05$) compared to male subjects (Fig. 5, Fig. 6). A significant genders-by-time interaction was found in knee valgus angle ($p < 0.05$)(Fig. 6). No significant gender effect was detected in knee flexion (Fig. 7).

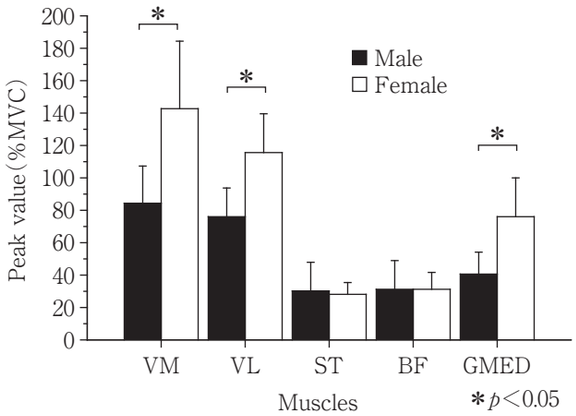

Figure 4 : Gender comparison of mean peak value of %MVC during single-leg landing. The asterisk indicates a statistically significant difference between male and female subjects. In VM, VL and GMED, female subjects exhibited significantly higher peak values of %MVC than those of male subjects.

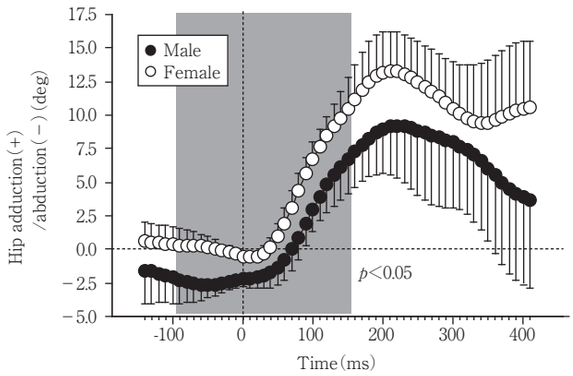

Figure 5 : Male and female mean hip adduction/abduction angles as a function of time. Two-way ANOVA test showed that there was a significant main effect of genders in hip adduction/abduction angles. Gray-colored area represents the period when the gender differences exit.

Table 3 . Within-gender differences in mean peak value of %MVC during single-leg landing.

Gender	Peak value (%MVC) (Mean ± SD)				
	VM	VL	ST	BF	GMED
Male	84.2 ± 22.8 [c,d,e]	76.4 ± 17.7 [c,d,e]	30.2 ± 17.4 [a,b]	30.8 ± 17.9 [a,b]	40.3 ± 13.4 [a,b]
Female	142.5 ± 42.3 [b,c,d,e]	115.6 ± 23.6 [a,c,d,e]	27.7 ± 7.7 [a,b,e]	31.6 ± 10.4 [a,b,e]	76.4 ± 23.4 [a,b,c,d]

a denotes significant difference vs VM ($p<0.05$)
b denotes significant difference vs VL ($p<0.05$)
c denotes significant difference vs ST ($p<0.05$)
d denotes significant difference vs BF ($p<0.05$)
e denotes significant difference vs GMED ($p<0.05$)

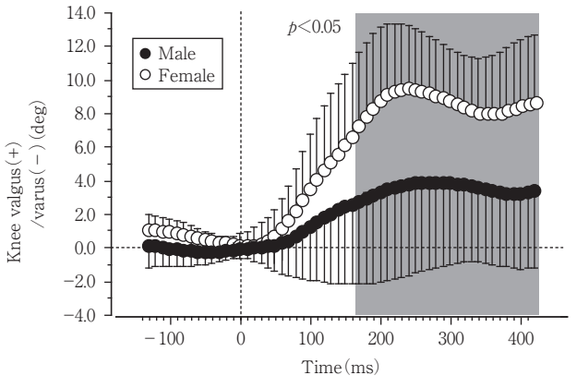

Figure 6 : Male and female mean knee valgus/varus angles as a function of time. Two-way ANOVA test showed that there was a significant main effect of genders in knee valgus/varus angles. A significant gender-by-time interaction was also found. Gray-colored area represents the period when the gender differences exit.

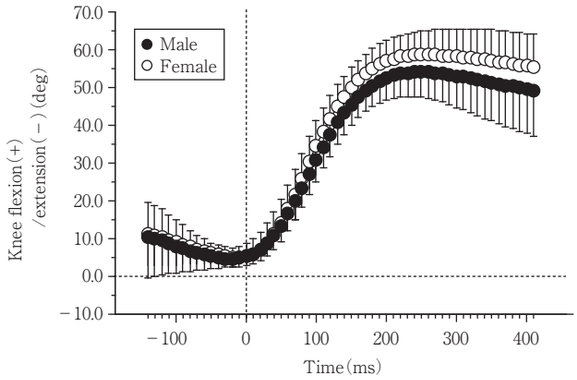

Figure 7 : Male and female mean knee flexion/extension angles as a function of time. Two-way ANOVA test showed that there was no significant main effect of genders.

Ⅳ. Discussion

The purpose of this study was to investigate the gender difference in neuromuscular hip and knee control during single-leg landing task. In agreement with previous studies[5–10], this study found that the female subjects exhibited the higher VM, VL, and GMED activities than those of male counterparts. ANOVA test for EMG peak value also found a significant gender-by-muscle difference interaction. This interaction clearly indicates that EMG peak values were influenced not only by muscle role difference but also gender difference, and suggests that male and female subjects utilized different neuromuscular strategy for hip and knee joint control. Females took more hip adducted and knee abducted positions after foot impact comparing to male subjects (Fig. 5, Fig. 6). Such lower limb orientations had reported as risk-elevated kinematics of ACL injury[11, 19–21]. Female subjects tried to control these frontal plane joint motion by increasing their VM and GMED, however, kinematic results demonstrated that they couldn't successfully keep their dynamic alignment after landing.

A. Strategy to control frontal plane knee kinematics in female subjects.

In this section, we discuss knee joint control in female subjects. Within females, VM activity was significantly higher than that of VL (Table 3). This difference between VM and VL was not found in male subjects. Additionally, from kinematic results, females showed greater knee valgus angle after landing (Fig. 6). These results suggest that females try to resist knee valgus by increasing VM activity[10]. Buchanan et al.[22] reported that VM, ST and gracilis muscles show higher activities against knee valgus load than against varus knee load. Dhaher et al.[23] also found that as valgus angle increased, VM activity increased accordingly. These studies suggest that the muscles located on the medial side of the knee specifically respond to knee valgus stress. However, the concern is that whether this VM activity is appropriate or not to avoid ACL injury because the aggressive activity of the knee extensors without proper hamstring activity would elicit the anterior translation of the tibia, resulting in a high load on ACL. The hamstrings of female subjects peaked much earlier than VM and VL (Table 2). BF in females especially peaked before initial foot contact. Male hamstring also tended to peak earlier (Table 2), but it was not as marked as female subjects showed (Fig. 3). These results indicate that, in females,

the peak activity of the quadriceps and hamstrings were not well synchronized as compared to male subjects. Cowling et al.[5] examined the gender differences in thigh muscle activities during landing focusing timing characteristics. They found that female ST peaked earlier than males. This finding was partially consistent with our own. Our results and previous report suggest that females are not good at synchronizing their quadriceps and hamstrings during landing. As many studies have reported, isolated quadriceps contraction with lack of hamstring co-contraction causes anterior tibial translation[24, 25]. For these reasons, VM activity, with which females tried to resist knee valgus, may pull the tibia anteriorly rather than protect the knee joint from valgus stress. Control strategy of knee joint in the frontal plane, using aggressive VM activity, is thus considered less-appropriate for avoiding ACL injury unless proper hamstring synchronization.

B. Female characteristics in hip joint control.

Although female subjects showed much higher GMED activities than male subjects, the hip joint displayed greater adduction (Fig. 5). These results indicate that females experienced difficulties in controlling the hip motion despite the higher GMED activities. Some interpretations have been suggested for failed stabilization of the hip joint from the perspective of muscular control. Zeller et al.[10] investigated gender differences in hip muscle (GMED and GMAX) activities and lower limb kinematics during single-leg squats. In their kinematic results, similar to our own results, female subjects showed greater hip adduction than male subjects. In EMG results, although no significant gender differences were identified, the female subjects tended to display higher GMAX and lower GMED activities. These data suggest that women may have difficulty activating GMED muscles. Another study using a single-leg landing task was conducted by Zazulak et al.[9]. They investigated gender differences in EMG activities of hip muscles and found that female GMAX exhibited significantly lower muscle activities as compared with male controls. No gender differences in GMED activity were seen. Zazulak et al. did not check kinematic data, but dynamic knee valgus and hip adduction were observed on visual observation. Given these findings, lower activities of GMAX were suggested to cause internal rotation of the femur and lead to decreased hip stabilization. As these reports explained, lower hip muscle activities would hardly generate sufficient joint stiffness to stabilize the hip joint. However, the same interpretations cannot be applied to our findings, as we identified higher GMED activities in female subjects. Higher GMED activities found in female subjects may represent a compensation for weakness of hip abductors. To achieve adequate force generation, female GMED required almost 80% of maximum activation level. However, the female population is generally recognized as displaying hip muscle weakness[26]. Thus, in spite of higher GMED activities, female subjects could not generate sufficient joint torque enough to stabilize the hip joint. Another possible reason for higher GMED activities in female subjects was miss collecting of MVC signal which potentially results in over estimation of %MVC values. However, this technical mistake hardly occurred because a great care was taken to measure MVC signal as detailed in method section. Moreover, it is hard to consider that all the eight female subjects were over estimated by the same technical mistake. There are of cause difficulties in evaluating hip joint stability only with the EMG data, as EMG signals do not necessarily correlate with muscle force output. In this study, we utilized the relative value %MVC to enable comparisons between genders. As %MVC represents only the amount of muscle activity relative to own MVC level. For example, if the %MVC is identical between female and male subjects, the generated joint torque may differ between genders. Thus it is difficult to discuss joint stiffness generation using EMG data alone. To solve this issue, kinetics analy-

sis may be helpful. Although the present study did not include any kinetic parameters, a detailed investigation of the relationship between EMG activities and joint torque would provide greater understanding of joint stability.

In conclusion, this study found that the female subjects exhibited much higher VM, VL, and GMED activities with more hip adducted and knee abducted position than those of the male subjects during single-leg landing motion. Higher VM activities without proper hamstring synchronization may cause anterior tibial translation rather than to resist the knee valgus angulation. Female GMED could not resist hip adduction, despite of its higher %MVC values. Much greater hip abductor strength might decrease the female's hip adduction and indirectly prevent knee valgus shift in the single-leg landing motion.

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【Original investigation】

The external force model for determining the frontal plane knee loading pattern –Implication for the mechanism of non-contact anterior cruciate ligament injury –

Issei Ogasawara[1], Shumpei Miyakawa[2], Shigeyuki Wakitani[1]

Abstract

This study proposes a simple model for predicting the external knee adduction-abduction moment, which is a key mechanism of anterior cruciate ligament (ACL) injury. We simplified the Newton-Euler's equation of motion by omitting its dynamic terms, since the experimental trial revealed that the contribution of the dynamic terms became negligible relative to the external force term during landing impact phase. The experimental data also showed that the external force term precisely predicted the knee adduction-abduction moment which was calculated by the Newton-Euler's equation of motion. This result means that the knee loading pattern during impact activity is largely determined by the external force and if the lower limb orientation with respect to the ground reaction force (GRF) is inappropriate, knee would experience a large abduction loading. Next, we estimated GRF and its acting point from the measured kinematic data aimed at predicting knee loads without using a force plate data. The result indicated that the moment calculated by the external force model using estimated GRF broadly predicted the profile of the Newton–Euler method, but was less precise during impact phase. As an implication for the mechanism of non contact ACL injury, the specific landing motions which can especially increase the knee abduction loading were introduced through model consideration.

key words: External force term; Anterior cruciate ligament injury; Risk evaluation; Knee abduction moment; Zero-moment point

Ⅰ. Introduction

The anterior cruciate ligament (ACL) is frequently injured during impact activities such as jump landing or side cut motion.[1,2] The forcefully abducted knee position at the moment of injury[3] implies that a large knee abduction moment is one of a main mechanism of this injury. An epidemiological study revealed that the population exhibiting increased knee abduction moment has a higher rate of ACL injury in comparison with the normal population.[4] Previous cadaveric data supports that knee abduction loading increases in situ force of the ACL.[5] This indicates that the knee adduction-abduction moment reflects the ligament's stress and it can be used as a predictor of the athlete's risk for ACL injury.[4]

A common way to calculate the knee adduction-abduction moment is to use Newton-Euler's equation of motion. The equation consists of dynamic terms (inertia, Coriolis, and centrifugal force) and the external force term (moment of external force). In the landing motion that a large ground reaction force (GRF) suddenly applies at the impact foot, the contribution of the dynamic terms are considered to be negligible relative to the external force term. This suggests that the knee adduction-abduction moment can be approximated using only the external force term, i.e., the moment caused by the GRF. We call this simplified equa-

1) Mukogawa Women's University, Department of Health and Sports Sciences. *6-46 Ikebiraki, Nishinomiya, Hyogo 663-8558, JAPAN.*
2) University of Tsukuba, Comprehensive Human Sciences. *1-1-1 Tennodai, Tsukuba, Ibaraki 305-8577 JAPAN.*
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tion of motion as "the external force model". The simplification of the equation of motion not only saves a computational cost to calculate dynamic terms but also provides a convenient estimate of the risk of the ACL injury.

One possible application of external force model would be a video analysis of ACL injury. Previous video analysis studies have mainly investigated the kinematic pattern of the injured knee joint.[3, 6-8] However, to know the true injury mechanism, it is essential to quantify the knee loads which actually caused the abnormal knee kinematics. If the GRF and its acting point are determined based on the kinematic data, the external force model will be able to extract the knee loading pattern from the video data.

The primary purpose of this study is to propose a external force model to predict the knee adduction-abduction moment aimed at establishing a convenient tool to determine the risk of ACL injury. The next objective is to verify the accuracy of the knee loading prediction based only on the kinematic data.

Ⅱ. Method

A. Model

1. Simplification of the equation of motion

Figure 1 (A) illustrates a right side leg model which involves the global Σ^O and shank Σ^S coordinate systems. All vectors in Fig. 1 (A) are represented in the global coordinate system Σ^O. The rotational equation of motion :

$$\boldsymbol{I}\ddot{\boldsymbol{\theta}} + \dot{\boldsymbol{\theta}} \times \boldsymbol{I}\dot{\boldsymbol{\theta}} = \boldsymbol{\tau} - \boldsymbol{J}^T\boldsymbol{f} \tag{1}$$

is usually used to analyze the dynamics of a link system, where $\boldsymbol{I}$ is the inertia matrix, $\boldsymbol{\theta}$ is the attitude of the segment, $\boldsymbol{\tau} = [\tau_x, \tau_y, \tau_z]^T$ is the joint moment, $\boldsymbol{J}$ is the Jacobian matrix and $\boldsymbol{f} = [f_x, f_y, f_z]^T$ is the GRF. The inertia term $\boldsymbol{I}\ddot{\boldsymbol{\theta}}$ and the gyroscopic torque $\dot{\boldsymbol{\theta}} \times \boldsymbol{I}\dot{\boldsymbol{\theta}}$ are moments derived from a rotational movement of a segment, and the external force term $\boldsymbol{J}^T\boldsymbol{f}$ is the moment caused by GRF. When the foot impacts the ground, the moment of GRF $\boldsymbol{J}^T\boldsymbol{f}$ becomes dominant compared to both the inertia term $\boldsymbol{I}\ddot{\boldsymbol{\theta}}$ and gyroscopic torque $\dot{\boldsymbol{\theta}} \times \boldsymbol{I}\dot{\boldsymbol{\theta}}$ as

$$\|\boldsymbol{I}\ddot{\boldsymbol{\theta}}\| \ll \|\boldsymbol{J}^T\boldsymbol{f}\|, \quad \|\dot{\boldsymbol{\theta}} \times \boldsymbol{I}\dot{\boldsymbol{\theta}}\| \ll \|\boldsymbol{J}^T\boldsymbol{f}\|. \tag{2}$$

We thus neglect the dynamic terms and obtain the externally applied knee moment $\hat{\boldsymbol{\tau}} = [\hat{\tau}_x, \hat{\tau}_y, \hat{\tau}_z]^T$ as

$$\hat{\boldsymbol{\tau}} = \boldsymbol{J}^T\boldsymbol{f}. \tag{3}$$

The Jacobian matrix is defined as

$$\boldsymbol{J} = [\boldsymbol{e}_x \times \boldsymbol{p}, \ \boldsymbol{e}_y \times \boldsymbol{p}, \ \boldsymbol{e}_z \times \boldsymbol{p}], \tag{4}$$

where unit vectors $\boldsymbol{e}_i\,(i = x, y, z)$ are the bases of the shank coordinate system Σ^S, which represent the knee rotation axes; $\boldsymbol{e}_x$ is the adduction-abduction axis pointing forward, $\boldsymbol{e}_y$ is the flexion-extension axis pointing medial, and $\boldsymbol{e}_z$ is the internal-external rotation axis pointing upward, respectively. The moment arm vector

$$\boldsymbol{p} = \boldsymbol{r}_c - \boldsymbol{r}_k \tag{5}$$

is the vector from knee joint center $\boldsymbol{r}_k = [r_{kx}, r_{ky}, r_{kz}]^T$ to the center of pressure (CoP) $\boldsymbol{r}_c = [r_{cx}, r_{cy}, r_{cz}]^T$. The knee adduction-abduction moment $\hat{\tau}_x$ can be extracted from Eq. (3) as

$$\begin{aligned} \hat{\tau}_x &= (\boldsymbol{e}_x \times \boldsymbol{p})^T \boldsymbol{f} \\ &= \|\boldsymbol{p}\|\boldsymbol{u}^T\boldsymbol{f}, \end{aligned} \tag{6}$$

where $\|\boldsymbol{p}\|$ is the length of the moment arm vector $\boldsymbol{p}$ and

$$\boldsymbol{u} = \frac{\boldsymbol{e}_x \times \boldsymbol{p}}{\|\boldsymbol{e}_x \times \boldsymbol{p}\|} \tag{7}$$

is the unit vector pointing medial and mutually perpendicular to $\boldsymbol{e}_x$ and $\boldsymbol{p}$. $\hat{\tau}_x$ in Eq. (6) is the knee adduction-abduction moment expressed in the shank coordinate system Σ^S, and a negative $\hat{\tau}_x$ denotes the knee abduction moment. The external force model (Eq. (6)) predicts the knee moment without angular accelerations, segmental masses, inertial moments and, moreover, segment by segment computations.

In addition, the external force model helps in understanding the principle of knee abduction moment generation. Since $\|\boldsymbol{p}\|$ is non-negative and nearly constant, $\boldsymbol{u}^T\boldsymbol{f}$ determines the direction and magnitude of knee adduction-abduction moment.

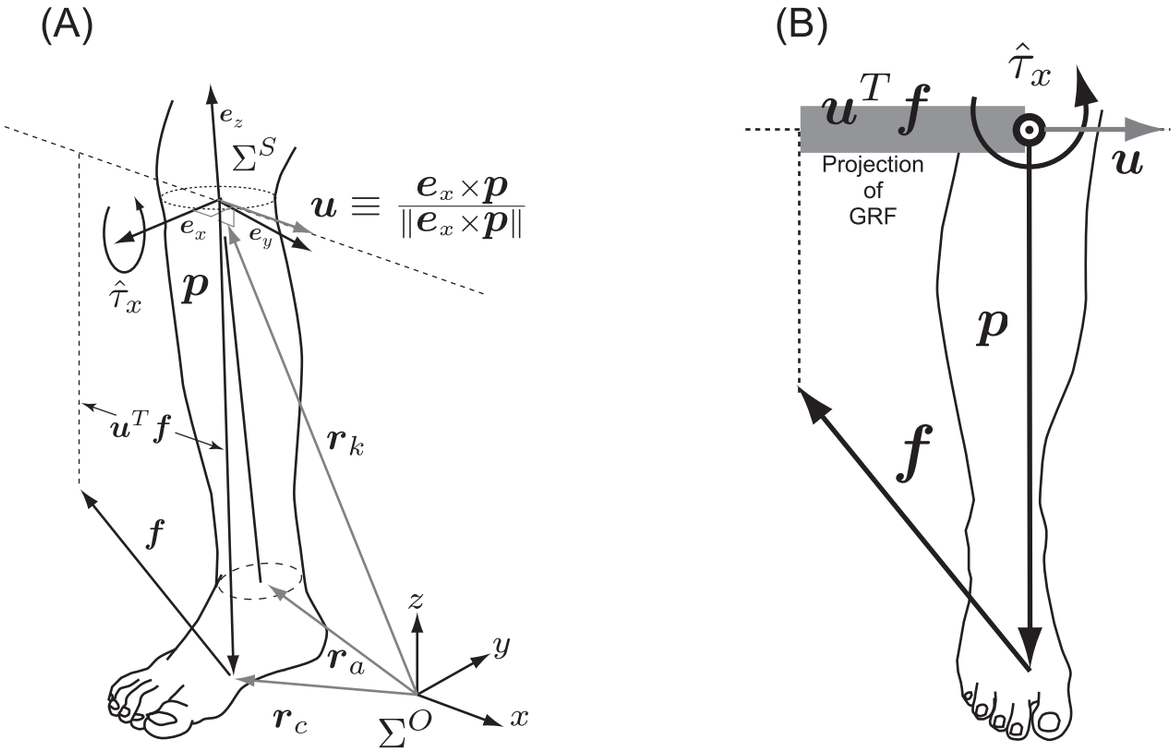

Figure 1: Schematic representation of the external force model

(A) The right side lower leg model. Σ^O and Σ^S denote the global and shank coordinates, respectively. $\hat{\tau}_x$ is the knee adduction-abduction moment acting about knee sagittal axis $\boldsymbol{e}_x$. The vector $\boldsymbol{p}$, which goes from knee joint center to center of pressure, represents moment arm. The vector $\boldsymbol{f}$ denotes ground reaction force. The vector $\boldsymbol{u}$ is the unit vector that is determined by orientations of both vector $\boldsymbol{p}$ and vector $\boldsymbol{e}_x$. The direction and the magnitude of the knee adduction-abduction moment is determined by $\boldsymbol{u}^T\boldsymbol{f}$; a projection of ground reaction force vector $\boldsymbol{f}$ onto the vector $\boldsymbol{u}$. The position vector of the center of ankle joint $\boldsymbol{r}_a$ was defined as the midpoint of the medial and lateral malleolus markers. The position vector of the center of knee joint $\boldsymbol{r}_k$ was defined as the midpoint of the medial and lateral femoral epicondyle markers. The internal-external rotation axis $\boldsymbol{e}_z$ was defined as a unit vector from the ankle joint center $\boldsymbol{r}_a$ to the knee joint center $\boldsymbol{r}_c$. The flexionextension axis $\boldsymbol{e}_y$ was defined as a unit vector which is medially pointing and mutually perpendicular to both $\boldsymbol{e}_z$ and a supplemental vector from the knee joint center $\boldsymbol{r}_c$ to the position of the marker on tibial tuberosity. Then the adduction-abduction axis $\boldsymbol{e}_x$ was defined by a cross product between $\boldsymbol{e}_y$ and $\boldsymbol{e}_z$.

(B) When GRF vector $\boldsymbol{f}$ directs laterally with respect to the moment arm vector $\boldsymbol{p}$, the knee moment becomes abduction.

$\boldsymbol{u}^T\boldsymbol{f}$ represents a projection of the GRF vector $\boldsymbol{f}$ onto the vector $\boldsymbol{u}$. When the acting line of the GRF $\boldsymbol{f}$ vector directs laterally with respect to the moment vector $\boldsymbol{p}$, then $\boldsymbol{u}^T\boldsymbol{f}$ increases in the negative direction and the knee moment becomes an abduction moment (Fig. 1 (B)). We suggest that the laterally directed GRF with respect to the moment arm vector $\boldsymbol{p}$ is the mechanism of knee abduction moment generation.

2. Estimation of GRF and its acting point from kinematic data

This section proposes models to estimate GRF and its acting point using measured kinematic data in the event that force plate data is unavailable. Assuming the body segments are rigid bodies, net force acting on the body is obtained by summing up translational equation of motion of all segments as

$$\hat{\boldsymbol{f}} = M(\ddot{\boldsymbol{c}} - \boldsymbol{g}), \tag{8}$$

where $M = \sum_{i=1}^{n} m_i$ is the total of each segmental mass m_i,

$$\boldsymbol{c} = \frac{1}{M}\sum_{i=1}^{n} m_i \boldsymbol{r}_i \tag{9}$$

is the position vector of body center of mass (CoM), $\boldsymbol{g} = [0, 0, -9.8]^T$ is the gravitational acceleration vector and $\boldsymbol{r}_i$ is the position vector of ith segment's CoM. While standing with a single leg, if no other forces except for GRF are acting, then $\hat{\boldsymbol{f}}$ is assumed to be the GRF acting on stance foot. The acting point of GRF (CoP) was calculated by referring to the idea of zero-moment point (ZMP).[9] This idea suggests that ZMP coincides with CoP when the ground reaction forces and moments balance all the other forces and moments acting on the body. It also suggests that the horizontal components of the ground reaction moment vector $\boldsymbol{\tau}_p = [\tau_{px}, \tau_{py}, \tau_{pz}]^T$ acting at ZMP are always zero as

$$\boldsymbol{\tau}_p = [0, 0, \tau_{pz}]^T. \tag{10}$$

We thus obtain the position vector of ZMP $\boldsymbol{r}_z = [r_{zx}, r_{zy}, r_{zz}]^T$ from an equation of moment equilibrium by selecting a point at which horizontal components of the ground reaction moment vector, τ_{px} and τ_{py}, will be zero. The whole moment of the body $\boldsymbol{\tau}_{all}$ can be obtained by differentiating the angular momentum of the body $\boldsymbol{H}$ as

$$\boldsymbol{\tau}_{all} = \frac{d}{dt}\boldsymbol{H} = \boldsymbol{c} \times M\ddot{\boldsymbol{c}}. \tag{11}$$

The whole moment $\boldsymbol{\tau}_{all}$ is also expressed by sum of all the moments acting on the body as

$$\boldsymbol{\tau}_{all} = \boldsymbol{\tau}_g + \boldsymbol{\tau}_f, \tag{12}$$

where

$$\boldsymbol{\tau}_g = \boldsymbol{c} \times M\boldsymbol{g} \tag{13}$$

is the moment of gravity, and

$$\boldsymbol{\tau}_f = \boldsymbol{r}_z \times \boldsymbol{f} + \boldsymbol{\tau}_p \tag{14}$$

is the moment of GRF $\boldsymbol{f}$. Substituting Eq. (11), (13), and (14) into Eq. (12) gives

$$\boldsymbol{\tau}_p = \boldsymbol{c} \times M\ddot{\boldsymbol{c}} - \boldsymbol{c} \times M\boldsymbol{g} - \boldsymbol{r}_z \times \boldsymbol{f}. \tag{15}$$

We substitute the estimated GRF vector $\hat{\boldsymbol{f}}$ (Eq. (8)) into GRF vector $\boldsymbol{f}$ in Eq. (15) to obtain position vector of ZMP as

$$\boldsymbol{r}_z = \begin{bmatrix} c_x - \frac{c_z \ddot{c}_x}{\ddot{c}_z + g} \\ c_y - \frac{c_z \ddot{c}_y}{\ddot{c}_z + g} \\ 0 \end{bmatrix}. \tag{16}$$

3. The external force model with estimated GRF

Substituting estimated GRF $\hat{\boldsymbol{f}}$ (Eq. (8)) and ZMP $\boldsymbol{r}_z$ (Eq. (16)) into $\boldsymbol{f}$ in Eq. (6) and $\boldsymbol{r}_c$ in Eq. (5) respectively, we have the external force model with estimated GRF as

$$\begin{aligned} \check{\tau}_x &= \|\boldsymbol{r}_z - \boldsymbol{r}_k\| \hat{\boldsymbol{u}}^T \hat{\boldsymbol{f}} \\ &= M\|\hat{\boldsymbol{p}}\| \hat{\boldsymbol{u}}^T (\ddot{\boldsymbol{c}} - \boldsymbol{g}), \end{aligned} \tag{17}$$

where

$$\hat{\boldsymbol{p}} = \boldsymbol{r}_z - \boldsymbol{r}_k \tag{18}$$

is the moment arm vector taking account of ZMP and

$$\hat{\boldsymbol{u}} = \frac{\boldsymbol{e}_x \times \hat{\boldsymbol{p}}}{\|\boldsymbol{e}_x \times \hat{\boldsymbol{p}}\|} \tag{19}$$

is the unit vector that is pointing medial and perpendicular to both $\boldsymbol{e}_x$ and $\hat{\boldsymbol{p}}$.

B. Experiment

To verify the accuracy of the two external force models (Eq. (6) and Eq. (17)), we conducted a single-legged landing experiment and compared the results of these two solutions with that of the Newton–Euler method.

1. Subjects and protocol

Seven healthy adults (3 men: 25.0 ± 1.0 yr, 172.3 ± 11.0cm, 68.2 ± 8.8kg, 4 women: 24.2 ± 0.5 yr, 161.7 ± 3.7cm, 56.2 ± 5.2kg), with no history of lower limb injuries, participated in this experiment. We explained the purpose of this research to the subjects and obtained written informed consent that was approved by the ethics committee of Japan Institute of Sports Sciences. After a 10 min warm up, we placed reflective markers on the subjects as shown in Fig. 2 and explained to them the experimental protocol that is detailed below.

Subjects were asked to fall from a 0.3 m high box and land on the force plate with their dominant leg. The dominant leg for each subject was

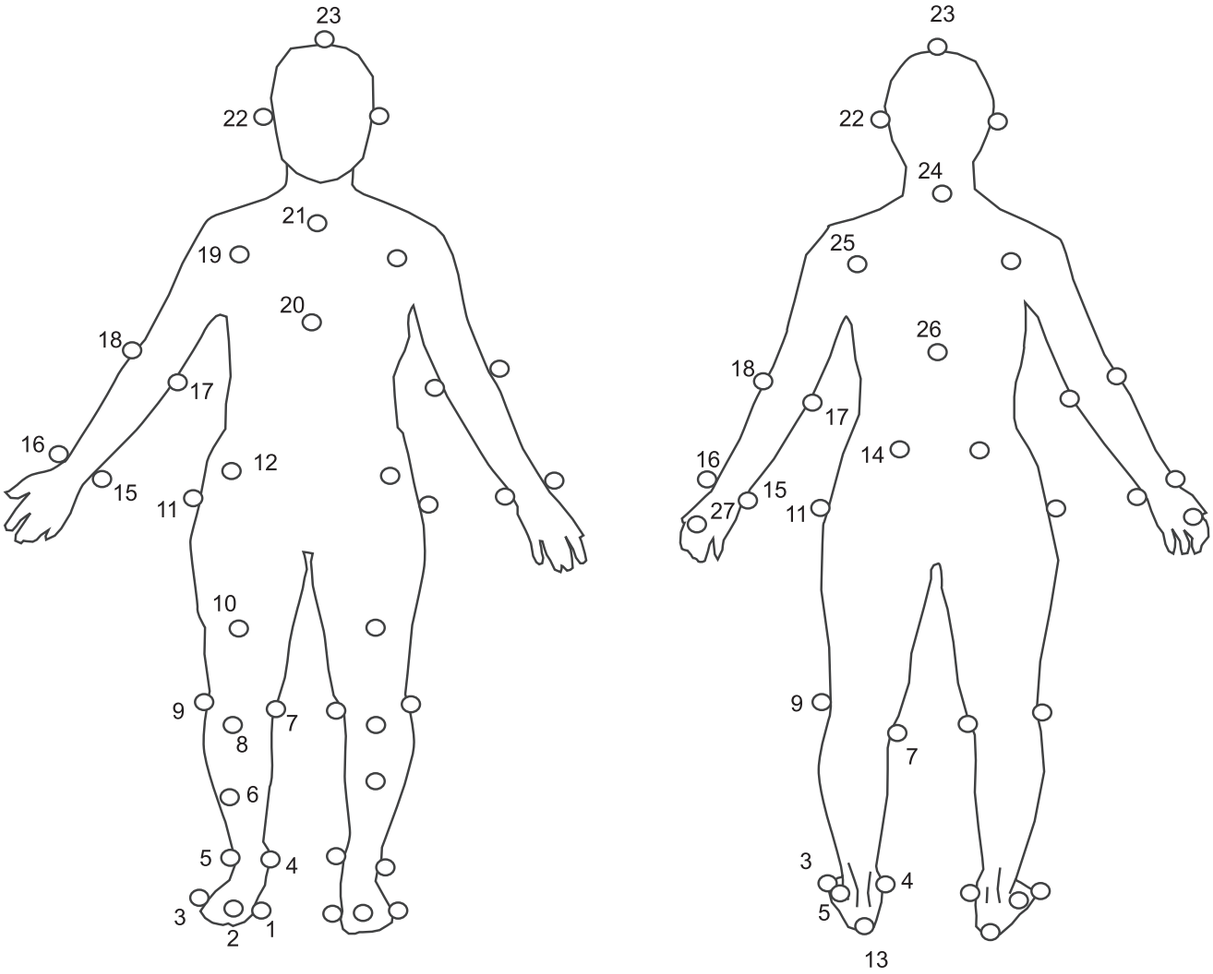

Figure 2: Position of the body markers

Reflective markers placed on following bony landmarks. (1) first metatarsophalangeal (MTP) joint, (2) second MTP joint, (3) fifth MTP joint, (4) tip of medial malleolus, (5) tip of lateral malleolus, (6) anterior aspect of shank, (7) the most medial point of the border of the medial femoral epicondyle, (8) the center of tibial tuberosity, (9) the most lateral point of the border of the lateral femoral epicondyle, (10) anterior aspect of thigh, (11) tip of great trochanter, (12) anterior superior iliac spine, (13) the most posterior point of the heel, (14) posterior superior iliac spine, (15) ulnar styloid process, (16) radial styloid process, (17) medial epicondyle of humerus, (18) lateral epicondyle of humerus, (19) anterior aspect of the shoulder joint, (20) the most inferior edge of the sternum, (21) mid point of bilateral sternoclavicular joint, (22) in front of the ear, (23) tip of the head, (24) C7 spinous process, (25) posterior aspect of the shoulder joint, (26) T10 spinous process and (27) mid point of the third metacarpal bone.

determined as the leg that was usually used to kick a ball. No further instructions were provided. Trials in which a subject was able to maintain a single leg stance for two seconds were regarded as successful. A maximum of 56 trials per subject were carried out, with 3 minutes of rest after every 10 trials to avoid fatigue.

C. Data acquisition and analysis

The positions of the reflective markers were captured with a Vicon 624 system (Oxford Metrics, Oxford, UK) at a sampling frequency of 120 Hz. The GRF data was measured by the force plate (9287B, Kistler, Winterthur, Switzerland) synchronously with the kinematic data. Trials with more than 20 consecutive frame gaps were excluded. The position data of the reflective markers

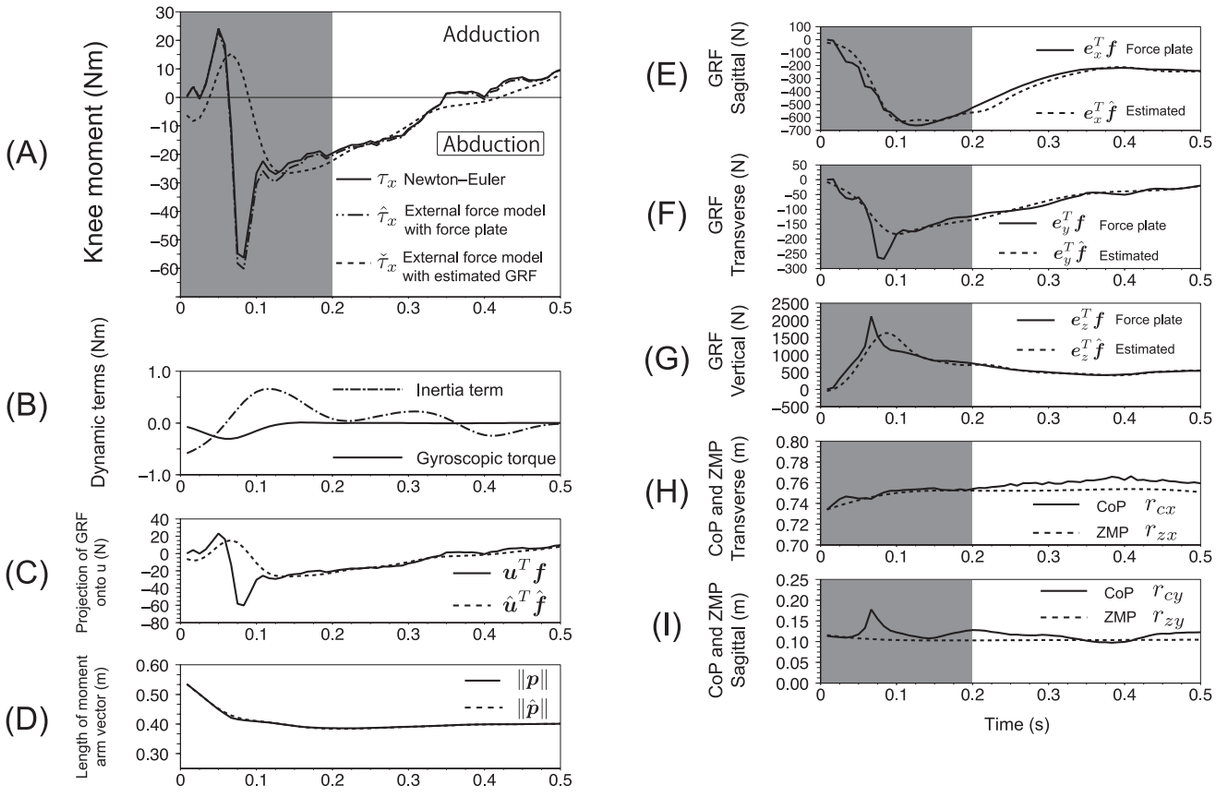

Figure 3: Time histories of knee moment, dynamic terms, external force term, moment arm length, estimated GRF, and ZMP.

A time history of the knee joint moment, GRF and ZMP from a representative subject. (A) Comparison of the knee adduction-abduction moment (Newton–Euler method τ_x: solid line, the external force model using force plate data (Eq. (6)) $\hat{\tau}_x$: dashed spaced line, and the external force model using estimated GRF (Eq. (17)) $\check{\tau}_x$: dashed line). (B) Dynamic terms of equation of motion. Both torques are acting about the knee adduction-abduction axis $\boldsymbol{e}_x$ (C) Projection of the GRF vector onto the vector $\boldsymbol{u}$. (D) Length of the moment arm vector. (E–G) Comparison of the force plate (solid line) and estimated GRF (dashed line), expressed in the shank coordinate system Σ^S. (H, I) Comparison of measured CoP by force plate (solid line) and calculated ZMP from kinematic data (dashed line). The gray shaded areas represent the periods analyzed with RMSE and % RMSE.

were smoothed using a low-pass second order zero lag Butterworth digital filter at cut-off frequencies of 10 Hz for the transverse and sagittal components and 18 Hz for the vertical component. Cut-off frequencies were determined to minimize the root mean square error (RMSE) between the estimated GRF (Eq. (8)) and force plate data.[10] Using the measured kinematic and force plate data, the knee adduction-abduction moments based on different solutions (τ_x: Newton–Euler method, $\hat{\tau}_x$: the external force model with force plate data (Eq. (6)) and $\check{\tau}_x$: the external force model with estimated GRF (Eq. (17))) were calculated. The anthropometric parame- ters were estimated based on[11] for the inverse dynamics calculation with the Newton–Euler method. $\hat{\tau}_x$ and $\check{\tau}_x$ were compared with τ_x to verify the accuracies of each external force model. The estimated GRF $\hat{\boldsymbol{f}}$ and ZMP $\boldsymbol{r}_z$ were compared with the force plate data. The errors between different solutions were quantified using the RMSE and relative RMSE (% RMSE).[12] The time window of interest was defined from 0 to 0.2 s after the initial foot impact.

Ⅲ. Results

A. Accuracy of the external force model

Figure 3 (A) compares the moment of the Newton-Euler method (solid line) and those of two external force models; one using the force plate data (dashed spaced line) and the other using the estimated GRF (dashed line). The external force model with force plate data showed a good prediction with small % RMSE of 2.12±1.1 % and RMSE of 1.71±1.0 Nm. Figure 4 (A) shows the relationship between $\hat{\tau}_x$ and τ_x from all 295 trials at each 0.05 s steps from 0.05 to 0.2 s after foot impact. This figure also shows that $\hat{\tau}_x$ linearly correlates with τ_x throughout the investigated period. Figure 3 (B) illustrates the time histories of the dynamic terms of the same trial. The small amplitudes of these dynamic terms indicate that the external force term of the equation of motion mainly determines the magnitude of the knee adduction-abduction moment. Figure 3 (C) shows the time profile of $\boldsymbol{u}^T\boldsymbol{f}$ and $\hat{\boldsymbol{u}}^T\hat{\boldsymbol{f}}$, and Fig. 3 (D) represents the length of the moment arm vectors $\boldsymbol{p}$ and $\hat{\boldsymbol{p}}$. Both $\boldsymbol{u}^T\boldsymbol{f}$ and $\hat{\boldsymbol{u}}^T\hat{\boldsymbol{f}}$ show profiles similar to those of τ_x and $\hat{\tau}_x$, respectively. The length of the moment arm vectors was nearly constant throughout the period of interest, except for a small shortening due to ankle dorsiflexion in the early phase. These results indicated that the direction and magnitude of each knee adduction-abduction moment (τ_x and $\hat{\tau}_x$) are mainly determined by the external forces $\boldsymbol{u}^T\boldsymbol{f}$ and $\hat{\boldsymbol{u}}^T\hat{\boldsymbol{f}}$, respectively. The moment calculated by the external force model using the estimated GRF (Eq. (17)) indicated less accurate results (% RMSE of 24.1±7.6 % and RMSE of 19.3±8.4 Nm) than when using the force plate data. The time pattern of moment $\check{\tau}_x$ did not follow the sudden change of the moment τ_x calculated by the Newton-Euler method in the early phase (Fig. 3 (A)). The poor accuracy in the early phase is also obvious in the correlation coefficients between $\check{\tau}_x$ and τ_x that were the smallest at 0.05 s (R = 0.90) and were gradually increased as time passed by (Fig. 4 (B)).

B. GRF and ZMP calculation from kinematic data

Figure 3 (E) - (G) compares the estimated GRF $\hat{\boldsymbol{f}}$ with the force plate data $\boldsymbol{f}$. The general trend of each component broadly agreed with the force plate data; however, the impact force in the early phase cannot be estimated in every component and large errors were obtained (RMSE

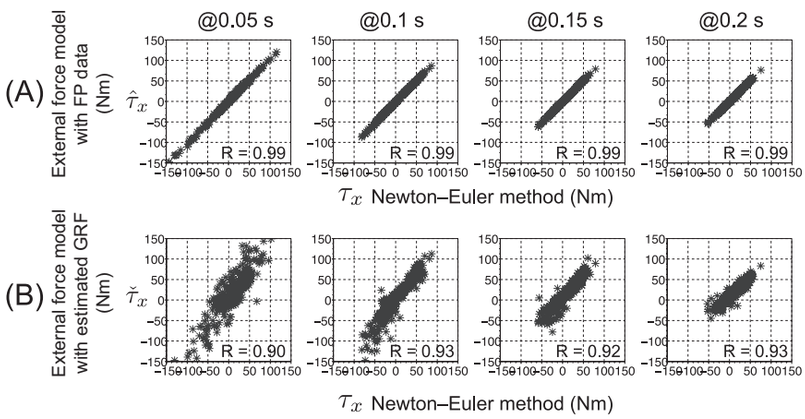

Figure 4: Association between each external force model and the Newton-Euler method.

(A) Association between the knee moment calculated by the external force model with force plate data and that of the Newton-Euler method. (B) Association between the knee moment calculated by the external force model with estimated GRF and that of the Newton-Euler method. Every plot was based on data from all 295 trials of every subject at each 0.05 s steps from 0 to 0.2 s after foot impact. The external force model showed a good prediction throughout the investigated period; however, when using the estimated GRF, less accurate estimates were obtained in the early phase.

Transverse: 26.6±17.5 N, Sagittal: 66.3±19.4 N, and Vertical: 149.2±64.5 N, % RMSE Transverse: 19.7±3.7 %, Sagittal: 19.9±3.1 %, and Vertical: 8.3±2.1%).

Figures 3 (H) and (I) compare the patterns of ZMP and CoP during the same trial. Although the resulting % RMSEs values were large (36.4±14.8 % for transverse and 61.5±28.1 % for sagittal component) due to ZMP's small value, the absolute errors (0.01±0.01 m for transverse and 0.04±0.02 m for sagittal) were reasonably small to allow substitution of the CoP.

Ⅳ. Discussion

A. Accuracy of the external force models

The purpose of this study was to propose a simple method to evaluate the knee loading pattern during impact phase using the external force model. The experimental results indicated that the moment calculated by the external force model using the force plate data (Eq. (6)) strongly agreed with the moment calculated by the Newton–Euler method, despite the fact that the external force model neglects the dynamic terms of the equation of motion. This indicates that the knee adduction-abduction moment is largely determined by the external force (GRF), which is highly increased just after foot impact.

The benefits of the external force model include not only reducing computational costs or avoiding recursive calculations but also eliminating the influence of numerical noises which are accumulated in the differential process.

Moreover, this model provide a insight into decreasing the risk of ACL injury. As Fig. 1 (B) and Fig 3 (A) shows, the knee adduction-abduction is increased as the magnitude of $\boldsymbol{u}^T\boldsymbol{f}$ increased. Therefore, to minimize knee abduction loading and

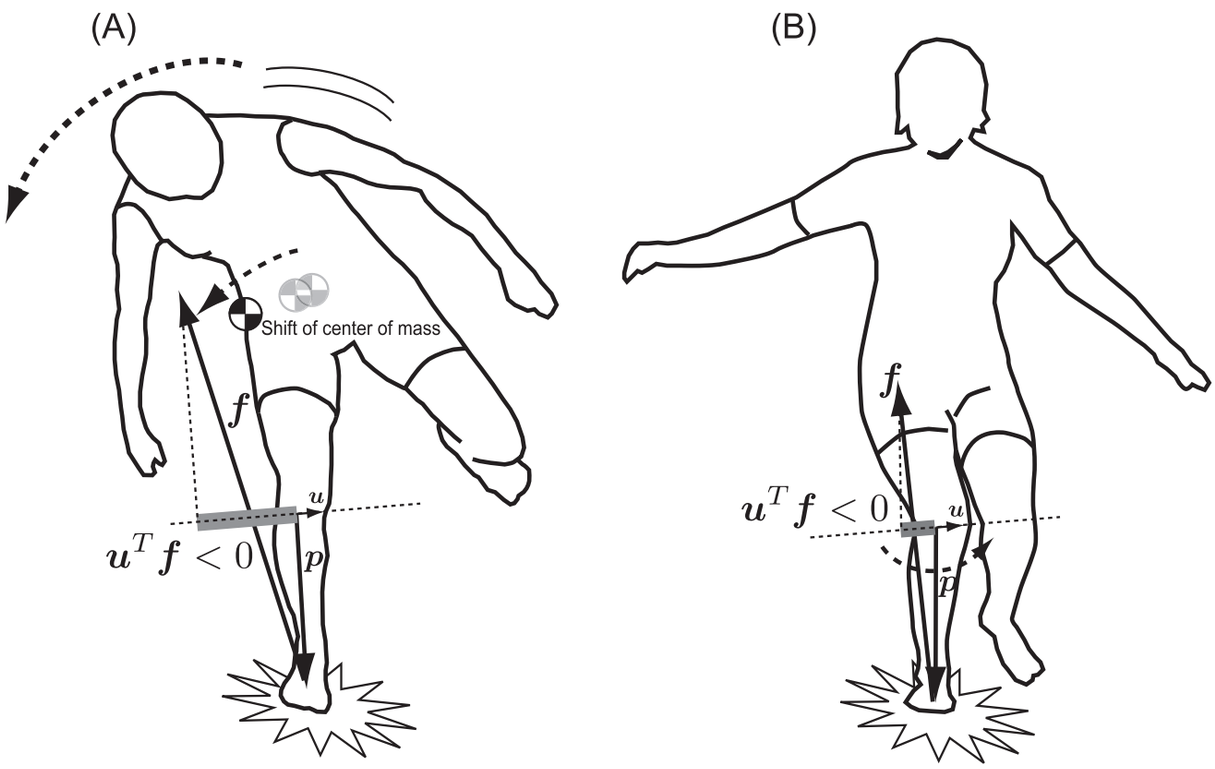

Figure 5: Risk elevating landing postures

Examples of landing postures that potentially increase the knee abduction moment. The lateral shift of the body CoM toward the landing limb (A) or medially positioned lower shank (B) experience increased knee abduction moment. To reduce the risk of ACL injury, one should avoid these landing postures.

to prevent ACL injuries, a great care should be taken to align landing leg orientation relative to the direction of GRF so as to minimize the magnitude of $\boldsymbol{u}^T\boldsymbol{f}$.

One possible application of the external force model is to analyze the injury video data, and determine how large the knee abduction moment occurred at the moment of injury. However, the force plate data is usually unavailable in the actual sports games. This study hence attempted to estimate GRF using kinematic data aimed at future application of the external force model to injury video analysis. The results indicated that the knee adduction-abduction moment calculated by the external force model using estimated GRF (Eq. (17)) broadly agreed with that calculated by the Newton–Euler method. However it was less accurate than that calculated using force plate data because the GRF model (Eq. (8)) cannot predict impact force in the early phase. This result suggested that it is difficult to estimate impact force only with the measured kinematic data. The ACL injury is reported to occur 13–105 ms after impact[2] and its time duration overlaps with that of the impact force. The precise estimates of GRF in impact phase is therefore required to yield accurate risk evaluation. The impact force, i.e., high frequency force in time histories in GRF[13] is generated by the foot collision with the ground; previous studies have estimated the time pattern of impact force using mass-spring-damper models.[14–16] At this point, our model does not include such mathematical approach and this is recognized as a substantial limitation for the current model which are going to analyze the impact phase of ACL injury. Future studies should add a spring and damper terms into current model to achieve more sensitive prediction of the impact induced knee moment.

Although large RMSEs were expected in predicting ZMP due to the acceleration of the body CoM, a practical accuracy was obtained. One limitation of the current ZMP model is that it cannot distinguish between more than two force acting points. Our approach cannot investigate the ACL injury during bilateral landing or that involving body contacts from other players because the acting point of GRF on injured leg becomes indeterminable. The contact force cannot be obtained as well in the latter case. Hence it is noted that the current external force model using ZMP is feasible only for single stance and non-contact ACL injury.

B. Clinical implications

In addition to the aspect of risk screening, the external force model is helpful to understand how the knee abduction moment occurs. We can suggest that the following landings increase the knee abduction moment. First, a lateral shift of the body CoM toward the landing limb, which allows the GRF to direct laterally with respect to the vector $\boldsymbol{p}$ (Fig. 5 (A)). This kind of landing posture occurs due to the trunk leaning toward the landing limb.[3] Secondly, the medially positioned shank with respect to the acting line of the GRF will also increase the knee abduction moment (Fig. 5 (B)). Such a limb position is observed in real injury situations.[3] Our model suggests that these kinds of motions and limb positions should be avoided to reduce the risk of ACL injury.

Conflict of interest

None of the authors have conflict of interests.

Acknowledgements

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【Short communication】

Measurement Accuracy of Hand Dynamometers Used for Physical Fitness Testing

Kanji Watanabe[1], Kaori Watanabe[2], Masami Yoshioka[3]

Abstract

The present study was conducted as a field survey to determine the actual degree of hand dynamometer validation performed in schools. The investigation consisted of a questionnaire survey of 43 schools (12 elementary, 6 junior high, and 25 senior high schools) that participated in a workshop for faculty members in the departments of health and physical education, together with actual validation tests for hand dynamometers in 7 of 43 schools (2 public elementary schools and 5 junior high schools), all 7 of which agreed to the testing. The dynamometer validation device of Takei Scientific Instruments Co., Ltd., was used for this study. The questionnaire responses indicated that 5 (12%) of 43 schools had performed hand dynamometer validation and 38 (88%) had not. Although this survey was small in scale, this finding suggests that many schools in Japan do not perform hand dynamometer validation, for various reasons. In the present study, in which an error of ±2.5 kg or more was considered large, the maximum deviation from the reference value was +8.0 kg in a hand dynamometer that showed a mean error of +6.5 kg in the test range of 10-70 kg. The present study has highlighted the issues associated with hand dynamometers used in physical fitness tests. It will be necessary in the future for teachers to raise consciousness about the validation testing of hand dynamometers in schools.

key words：hand dynamometer; measurement accuracy; validation

Purpose

The report on physical fitness of Japanese youth (6 to 19 years of age), issued by the Ministry of Education, Culture, Sports, Science and Technology (MEXT) in 2009, indicated the emergence of a slightly increasing trend in physical fitness on a nationwide level, although it was still below the peaks recorded in 1985 for boys and girls[1]. The report also showed that physical fitness differed greatly depending on the presence or absence of regular exercise, which is also known to have a strong influence on the occurrence of obesity[2-5].

In the test methodology for comprehensive evaluation of physical fitness adopted in Japan in 1999, and in various other tests, the devices used for direct measurements include stopwatches, seated body anteflexion meters, vertical jump meters, and hand dynamometers.

The accuracy of hand dynamometers in particular is known to decline with frequent use, and in the national survey on physical fitness, MEXT directed that measuring devices be properly adjusted and measurement methods be properly implemented and accurately performed[1]. The same concept applies to schools not subject to the national survey guidelines. It is not clear, however, whether validation of hand dynamometers used in the physical fitness tests is regularly performed at general elementary, secondary, and tertiary

1) Mukogawa Women's University, School of Health and Sports Sciences, Department of Health and Sports Sciences
2) The University of Shiga Prefecture, School of human nursing
3) Shinden elementary school, Toyonaka city
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schools.

The present study was conducted as a field survey to determine the actual degree of hand dynamometer validation performed at public elementary and junior high schools.

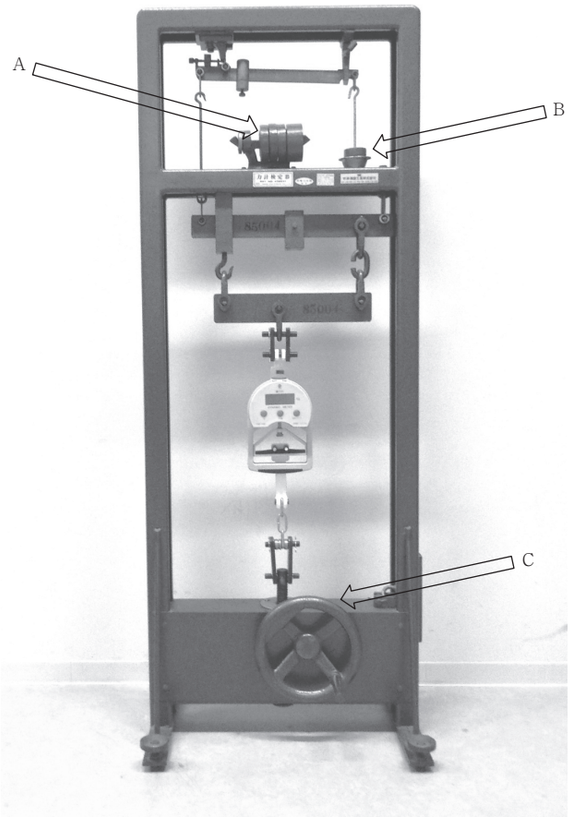

Figure 1 Calibration device

Methods

1. Objects of investigation

The investigation consisted of a questionnaire survey of 43 schools (12 elementary, 6 junior high, and 25 senior high schools) that participated in a workshop for faculty members in the departments of health and physical education, together with actual validation tests for hand dynamometers in 7 of 43 schools (2 public elementary schools and 5 junior high schools), all 7 of which agreed to the testing. Thus, we chose only the schools from which we had obtained agreement.

2. Questionnaire survey on hand dynamometers

The questionnaire items in this survey centered on: 1) whether physical fitness tests were done; and 2) the time of purchase of hand dynamometers, their state of validation, and reasons for non-validation. Faculty members in the departments of health and physical education were asked to complete the questionnaire.

3. Hand dynamometer validation

Validation testing was performed for 30 hand dynamometers in 2 public elementary schools and 5 junior high schools that had not themselves implemented validation of their hand dynamometers but agreed to its performance in this study. The dynamometer validation device of Takei Scientific Instruments Co., Ltd., was used for this purpose (Figure 1).

In the validation test, we first placed weights (Figure 1, A) with certified values of 10 kg to 70 kg in 10-kg increments on the scale pan of the dynamometer validation device as the reference values, then rotated its hand wheel (Figure 1, C) to apply force to the dynamometer hand grip, and determined the difference from the reference value.

At least 2 or 3 trials were performed at each reference level. The mean of 2 stable trials was used as the measured value, and its divergence from the reference value was used as the measurement error. The generally recognized accuracy of spring-type hand dynamometers is ±2.0 kg. All but 2 (thus, 28) of the hand dynamometers tested in this study were spring-type, and any divergence of ± 2.5 kg or more from the reference value was therefore deemed to be a large error.

Figure 2 shows a typical correlation between calibration test values obtained in 2 trials (the first and second) for one of the hand dynamometers. The high reliability of the test is shown by its correlation coefficient of $r^2 = 0.998$ ($p < 0.01$). The reliability of all test values obtained in this study was the same as shown in Figure 2.

4. Measurement period

May 2009 to August 2009

5. Statistical analysis

The reliability of the 2 stable measurement values was tested by Pearson's product-moment correlation coefficient. The compensation formula was constructed with Y as the reference value (10-70 kg) and X as the measured value. The statistical level of significance was less than 5%.

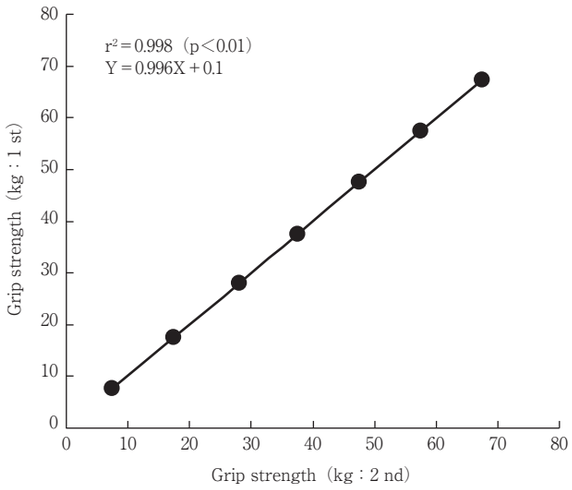

Figure 2 Reliability of certification values

Results

The questionnaire responses indicated that 39 (91%) of 43 surveyed schools conducted physical fitness testing and the remaining 4 (9%) did not. The reasons given for non-testing included problems such as 1) an inability to provide a proper measurement environment due to insufficient budget allocations and 2) avoidance of an influence of such testing on the progress of course work under the course revisions necessitated by changes in government curriculum guidelines.

Only 5 of 43 schools (12%) indicated in the questionnaire responses that they conducted validation testing of their hand dynamometers, and 38 schools (88%) indicated that they did not. The reasons given for non-performance included problems such as 1) lack of budget allocation, 2) insufficient time for preparation and performance due to busy schedules despite awareness of the need, and 3) insufficient knowledge of the method of validation despite awareness of the need.

Table 1 shows the differences found between the reference values and the measured values of 30 hand dynamometers at 7 schools. At one school (School F), the No. 1 and No. 2 hand dynamometers were used for young children, and the validation was therefore performed only up to 60 kg. Manufacturers present the hand dynamometers as uniformly accurate to within 2.0 kg. In the present study, a difference of 2.5 kg or more was assumed to be an error, as is emphasized in gray in Table 1. The errors found in the validation tests ranged up to a maximum of 8.0 kg (error rate: 40%, E-1). Three different tendencies were found in hand dynamometer errors: 1) non-uniform variation in error size among the reference values, 2) emergence of errors only above a certain reference value, and 3) uniform errors over a certain range of reference values. Within the uniform-error range, the error rate tended to increase with decreasing standard weights (Table 1, E-1). Only 16 (53%) of the hand dynamometers consistently showed errors of 2.0 kg or less, and thus were not in need of adjustment.

Discussion

The reports by MEXT on its surveys of physical fitness and exercise capability include classification of results by student age and sex and changes in results as students get older[1]. These reports enable comparisons of changes in physical fitness and motor ability with the progression of student age, comparisons of results for a given age group in different periods, and determination of differences between girls and boys. The reports provide basic information related to the state of student health and the formulation of exercise programs.

It has been reported that, in the results of the physical fitness and motor ability tests, grip strength shows a close correlation with VO_2max, maximum anaerobic power, leg extension power[6], muscle strength other than grip strength[7], and limb muscle mass[6,8]. In short, it is recognized that

Table 1 Difference between criterion value and hand dynamometer certification value.

School	No.	Criterion weight														Measurement system/ indication
		10 kg		20 kg		30 kg		40 kg		50 kg		60 kg		70 kg		
		Error		Error		Error		Error		Error		Error		Error		
		kg	%	kg	%	kg	%	kg	%	kg	%	kg	%	kg	%	
A	1	−0.5	−5.0	0.5	2.5	0.5	1.7	1.0	2.5	1.0	2.0	1.0	1.7	0.5	0.7	SP/A
	2	−3.0	−30.0	−3.0	−15.0	−2.5	−8.3	−2.5	−6.3	−1.5	−3.0	−3.5	−5.8	−1.0	−1.4	SP/A
B	1	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.5	1.0	0.0	0.0	0.0	0.0	SP/A
	2	0.0	0.0	1.0	5.0	1.5	5.0	1.0	2.5	1.5	3.0	1.5	2.5	2.0	2.9	SP/A
C	1	3.0	30.0	1.8	9.0	1.3	4.3	1.0	2.5	1.0	2.0	0.8	1.3	0.3	0.4	SP/A
	2	2.5	25.0	1.8	9.0	1.0	3.3	0.5	1.3	0.0	0.0	−0.5	−0.5	−1.5	−2.1	SP/A
	3	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	−0.2	−0.3	SP/A
	4	2.5	25.0	2.5	12.5	2.5	8.3	1.5	3.8	1.3	2.6	0.3	0.5	0.5	0.7	SP/A
	5	1.0	10.0	1.0	5.0	1.0	3.3	1.0	2.5	0.5	1.0	−0.2	−0.3	−1.0	−1.4	SP/A
	6	2.0	20.0	2.0	10.0	3.0	10.0	3.0	7.5	3.0	6.0	3.0	5.0	3.0	4.3	SP/A
	7	1.0	10.0	1.0	5.0	1.3	4.3	1.3	3.3	1.3	2.6	1.0	1.7	1.0	1.4	SP/A
	8	1.8	18.0	1.6	8.0	1.8	6.0	1.7	4.3	1.2	2.4	0.7	1.2	0.0	0.0	SG/D
D	1	0.5	5.0	0.5	2.5	0.5	1.7	0.3	0.8	0.0	0.0	−0.5	−0.8	−0.5	−0.7	SP/A
	2	0.8	8.0	1.0	5.0	0.8	2.7	1.0	2.5	1.0	2.0	0.5	0.5	1.0	1.4	SP/A
	3	2.0	20.0	2.0	10.0	2.0	6.7	3.0	7.5	2.0	4.0	3.0	5.0	3.0	4.3	SP/A
	4	0.5	5.0	0.5	2.5	0.5	1.7	0.3	0.8	0.0	0.0	0.5	0.8	0.5	0.7	SP/A
E	1	7.0	70.0	8.0	40.0	7.0	23.3	6.2	15.5	6.4	12.8	4.3	7.2	4.3	6.1	SP/A
	2	1.0	10.0	1.0	5.0	1.0	3.3	2.0	5.0	2.0	4.0	2.0	3.3	3.0	4.3	SP/A
	3	0.2	2.0	0.0	0.0	−0.4	−1.3	−1.0	−2.5	−1.0	−2.0	−1.0	−1.7	−3.0	−4.3	SP/A
F	1	−1.0	−10.0	−1.7	−8.5	−1.5	−5.0	−3.0	−7.5	−3.0	−6.0	−4.0	−6.7	—		SP/A
	2	−1.0	−10.0	−0.5	−2.5	−1.2	−4.0	−3.0	−7.5	−3.0	−6.0	−4.0	−6.7	—		SP/A
	3	0.8	8.0	1.0	5.0	0.0	0.0	−0.7	−1.8	−0.7	−1.4	−2.5	−4.2	−3.0	−4.3	SP/A
	4	−1.5	−15.0	−1.5	−7.5	−0.7	−2.3	−1.0	−2.5	−1.7	−3.4	−0.5	−0.8	−1.0	−1.4	SP/A
	5	−0.2	−2.0	−0.2	−1.0	−0.2	−0.7	−0.2	−0.5	−0.5	−1.0	−1.0	−1.7	−1.2	−1.7	SG/D
G	1	1.0	10.0	1.5	7.5	−1.5	−5.0	−1.0	−2.5	−1.5	−3.0	−2.0	−3.3	−2.0	−2.9	SP/A
	2	−2.0	−20.0	−2.0	−10.0	−2.0	−6.7	−3.0	−7.5	−3.0	−6.0	−3.0	−5.0	−3.3	−4.7	SP/A
	3	2.0	20.0	2.0	10.0	3.0	10.0	2.0	5.0	3.0	6.0	3.0	5.0	3.0	4.3	SP/A
	4	2.0	20.0	2.0	10.0	2.0	6.7	2.0	5.0	2.0	4.0	2.0	3.3	2.0	2.9	SP/A
	5	2.0	20.0	1.0	5.0	2.0	6.7	1.0	2.5	0.0	0.0	1.0	1.7	1.0	1.4	SP/A
	6	2.0	20.0	2.0	10.0	2.0	6.7	2.0	5.0	2.0	4.0	1.0	1.7	1.0	1.4	SP/A

The value indicates difference with the criterion value.
□ The value indicates the error beyond 2.5 kg for each criterion weight
※The accuracy that the manufacturer shows is ±2 kg or less.
SP/A : Spring system/Analog
SG/D : Strain gauge system/Digital

grip strength is closely related to other muscle strength, physical fitness, and muscle mass. Grip strength is also deemed important for assessment of physical fitness in the elderly, because of its functional role in the performance of activities of daily living[9]. In a related trend for fall prevention, a toe dynamometer has recently been developed to measure toe muscle strength[10,11].

Various factors are being investigated for their influence on grip strength measurements, including measurement posture[9], grip width[12], and number of measurement repetitions[13]. Hand dynamometer validation also warrants consideration. Errors in measurement may result in significant underestimation or overestimation of grip strength. Advance validation before testing and adjustment of hand dynamometers is therefore essential.

The extent of measuring device validation and

adjustment for physical fitness tests at many elementary, secondary, and tertiary schools in Japan is not known. The present study may provide some insight into this question. It began with a questionnaire survey of 43 public and private elementary, junior high, and senior high schools concerning their implementation of hand dynamometer validation. The questionnaire responses indicated that 5 (12%) of 43 schools had performed hand dynamometer validation and 38 (88%) had not. Although this survey was small in scale, this finding suggests that many schools in Japan do not perform hand dynamometer validation, for various reasons.

We also performed validation tests for hand dynamometers at schools that had not performed validation, to investigate their accuracy. Only 2 of 30 devices tested were strain-gauge hand dynamometers, and 28 were spring-loaded. Spring-loaded hand dynamometers are generally considered accurate to ±2.0 kg. In contrast, strain-gauge hand dynamometers can generally be adjusted to provide an accuracy of ±0.1 kg, and they are also considered superior to the spring-loaded type because they can be used for 2 to 3 years, depending on frequency of use. The times of purchase ranged from 1976 to 2008, and some of the older dynamometers had last been inspected some 30 years prior to this study.

In the present study, in which an error of ±2.5 kg or more was considered large, the maximum deviation from the reference value was +8.0 kg in a hand dynamometer that showed a mean error of +6.5 kg in the test range of 10-70 kg. Measurements with this dynamometer would thus inherently yield a two-rank overestimation of grip strength on a 10-rank scale.

It was found that, even among the dynamometers at a given school, the measurement errors tended to include both overestimation and underestimation of grip strength. This may be attributable in part to differences in frequency of use among the dynamometers.

In their investigation on the measurement accuracy of six Jamar dynamometers, Harkonen et al.[14] tested each one at five different handle positions with seven weights of 5 kg to 60 kg and found that none of them exhibited any substantial difference in accuracy with differing grip breadths, but noted a tendency for lower accuracy in older dynamometers. In the present study, no relation was found between the period of dynamometer validation and the degree of error, as may be seen in the finding of 3-kg to 5-kg errors in a dynamometer (G-3) last validated 1 year before the present investigation, in contrast to the finding of no large error (except at 70 kg) in two dynamometers (E-2, 3) last validated 5 years before the present investigation. It was concluded that the large errors in the quite recently validated dynamometer (G-3) was probably the result of its high frequency of use. The oldest dynamometer (B-1), which had been purchased 33 years previously, was a case in which no large error was found. Although no relevant records remained, it appears that this dynamometer had been regularly validated.

The proportion of hand dynamometers at each school with an error of ±2.5 kg or more was at least 0% and at most 100%. Overall, the proportion of hand dynamometers found not to be in need of adjustment was 53% (16 hand dynamometers). Results of this study thus indicate that many schools had been performing measurements with hand dynamometers that resulted in large errors. The hand-written reasons given in the questionnaires for non-implementation of hand dynamometer validation included such problems as 1) insufficient budget allocations; 2) forgetfulness in the midst of busy schedules, despite an understanding of the need for validation; and 3) lack of knowledge concerning the method of validation, despite an awareness of the need for it.

It should be noted that with some of the dynamometers, the error (deviation from the reference value) was fairly uniform throughout the grip-strength test range of 10-70 kg, and with others

the error varied throughout the test range. If the error is uniform throughout this range and the deviation is displayed by the hand dynamometer, then in actual use it is possible to compensate for the error in succeeding measurements. For dynamometers showing non-uniform errors throughout the test range, it is possible to reduce the error size by constructing and applying a compensation formula with X as the measured value and Y as the estimated actual value. Such expediencies, however, are not really desirable, and it is preferable either to have a specialized agency adjust any hand dynamometer that yields large errors or to replace it with a new one.

It may be possible for the local board of education or a local school serving as a representative to set up a dynamometer validation post with a device such as the spring-type and strain gauge used in the present study to facilitate periodic validation, and where necessary, adjustment (or compensation formula derivation) for each school with its own hand dynamometers. Moreover, for cases in which the error is found to be too large, it would be advisable to consider either contracting a specialized agency to implement the hand dynamometer correction or purchasing a new one.

Obligatory record-keeping of the times of dynamometer maintenance and new dynamometer purchase may also be useful for future maintenance scheduling. The present study did not provide information useful for appropriate maintenance scheduling, and further study will be necessary to determine the relation of error occurrence and degree to the time of purchase, time of previous validation, and number of individuals measured per year.

In addition to the heightened understanding of the importance of hand dynamometer validation, one favorable result of the present study has been the decision by some schools to dispose of old instruments showing large errors and replace them with new ones.

Schools in Japan are currently confronted with major problems with budget cutbacks on educational expenditures. The investigation in this study only included a few schools, but there may be many schools that do not carry out the validation testing of hand dynamometers, which can be inferred from comments on the questionnaire. The present study has raised an important issue relating to the practice of health and fitness education. In the on-site measurement of children' s health and physical fitness at schools, proper management and assurance of the accuracy of the measurement devices is unquestionably important. The present study has highlighted the issues associated with hand dynamometers used in physical fitness tests, but this is only one small part of the overall problem. Going forward, it will be necessary to discuss means of further heightening the awareness of teachers in related positions of specialization while discussing methods for effective explanation of the need to secure budgetary allocations that will meet educational costs.

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【Review】

Commotio Cordis: importance of awareness

Shigehiro Tanaka[1), 2)]

Abstract

Commotio Cordis, primarily reported by B. J. Maron since 1990, is one of the causes of sudden cardiac death. In large autopsy-based surveys performed in the United States, hypertrophic cardiomyopathy has consistently been the single most common cardiovascular cause of sudden death, accounting for high incident rates in prior reports. However, it has been suggested that the prevalence of these events is not as high in younger individuals. Commotio cordis is the second most frequent cause of sudden cardiac death, with approximately 10 to 20 cases added to the Commotio Cordis Registry yearly since the late 1990s.

In patients with commotio cordis, ventricular fibrillation or complete A-V block is recognized before cardiac arrest. In recent years, survival from these incidents has increased, likely because of more rapid response times and access to defibrillation devices, as well as greater public awareness of this condition. In addition, automatic external defibrillator availability has contributed to this increase.

Ⅰ Introduction

Athletes are often considered to be some of the healthiest people in society, and sudden cardiac death (SCD) in a young athlete is shocking and profoundly impacts the school or community where it occurs. However, it has been suggested that the prevalence of these events is not as high as it may seem as compared to the considerable media attention that often accompanies SCD in a young individual.[1-3] On the other hand, SCD is the leading medical cause of death in college athletes in the United States, as well as the most common cause of death during sport or exercise participation, and occurs at a much higher rate than previously thought. The American College of Cardiology defines SCD as "nontraumatic and unexpected sudden death that may occur from a cardiac arrest, within 6 hours of a previously normal state of health".[2, 4]

A variety of cardiovascular abnormalities represent the most common causes of sudden death in competitive athletes, such as dilated cardiomyopathy, aortic rupture in the context of Marfan syndrome, myocarditis, valvular disease (aortic stenosis, mitral valve prolapse), and electrical disorders (Wolff-Parkinson-White syndrome, long QT syndrome, Brugada syndrome), as well as commotio cordis (malignant arrhythmia due to blunt chest trauma).[5]

Commotio cordis, a cause of SCD, has primarily been reported by B. J. Maron.[1, 2, 4, 6, 7] In those studies, sudden death from cardiac arrest that occurred during sports play after a blunt blow to the chest in the absence of structural cardiovascular disease or traumatic injury (cardiac concussion or commotio cordis) was studied in young cases, and the clinical features of this apparently uncommon but important phenomenon

1) Department of Health and Sports Sciences, School of Health and Sports Sciences, Mukogawa Women's University, 6-46 Ikebiraki-cho, Nishinomiya, 663-8558, Japan
2) Instisute for Health and Sports Science, Mukogawa Women's University, 6-46 Ikebiraki-cho, Nishinomiya, 663-8558, Japan
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were reported. The global occurrence of commotio cordis is very similar in regard to demographics is very similar in the United States and other countries.[7] However, the frequency of chest blows from soccer balls causing commotio cordis events, particularly during sports events played internationally, seems to contradict the prevailing notion that air-filled projectiles convey less risk for ventricular fibrillation than those with solid cores (e.g., baseballs or lacrosse balls). In Japan, some cases of Commotio cordis have also been reported to be caused by a fall[8, 9] or traffic accident,[10, 11] as well as sports activities.[12] As the reported papers of commotion cordis in Japan on sports activities is only one paper[12], we investigated reports of commotio cordis including those presented in Japan and provide here an overview.

Ⅱ Epidemiology

In large autopsy-based surveys of a total of 387 athletes performed in the United States, hypertrophic cardiomyopathy was consistently the single most common cardiovascular cause of sudden death, accounting for the high incidence noted in prior reports[13-15] (Table 1). However, it has been suggested that the prevalence of these events is not as high as it may seem, because of the considerable media attention that often accompanies SCD in a young individual.[1, 2] In those studies, commotio cordis was the second most common, occurring in 20% of those cases (Table 1).

Commotio cordis is generally seen in younger individuals, though not exclusively.[16-18] In another study, the condition showed a predilection for children and adolescents (mean age, 15 ± 9 years; range, 6 weeks to 50 years).[20] Among 224 cases over the past 15 years,[6, 21, 22] 26% of the affected individuals were younger than 10 years of age, whereas only 9% were 25 years or older. Commotio cordis has rarely been reported in blacks or females, as most victims are male (95%) and white (78%).

Table.1 Cause of sudden death in 387 young athletes[13)-15)]

Cause	No. (%)
Hypertrophic cardiomyopathy (HCM)	102 (26.4)
Commotio cordis	77 (19.9)
Coronary artery anomalies	53 (13.7)
Left ventricular hypertrophy of indeterminate causation (autopsy: suggestive of HCM (nsufficient for diagnosis)	29 (7.5)
Myocarditis	20 (5.2)
Ruptured aortic aneurysm	12 (3.1)
Arrhythmogenic right ventricular cardiomyopathy	11 (2.8)
Tunneled (bridged) coronary artery	11 (2.8)
Aortic-valve stenosis	10 (2.6)
Atherosclerotic coronary artery disease	10 (2.6)
Dilated cardiomyopathy	9 (2.3)
Myxomatous mitral-valve degeneration	9 (2.3)
Asthma (or other pulmonary condition)	8 (2.1)
Heat stroke	6 (1.6)
Drug abuse	4 (1.0)
Other cardiovascular cause	4 (1.0)
Long QT syndrome (documented on clinical evaluation)	3 (0.8)
Cardiac sarcoidosis	3 (0.8)
Trauma involving structural cardiac injury	3 (0.8)
Ruptured cerebral artery	3 (0.8)

Although cardiovascular collapse is virtually instantaneous, 20% of affected individuals remain physically active for a few seconds after the blow (e.g., continuing to walk, run, skate, throw a ball, or even speak), which may reflect individual tolerance for sustained ventricular tachycardia accompanied by arrhythmia. For example, a baseball pitcher struck in the chest by a batted ball was able to retrieve the ball at his feet, successfully complete the play (throwing out the base runner), and then prepare for the next pitch before collapsing. In another instance, a batter playing baseball was struck by a pitch while attempting to bunt and collapsed only after running to first base.[20] In such cases, ventricular tachycardia accompanied by arrhythmia appears

first, after which cardiac arrest occurs, with cerebral and body blood flow continuing for a few or several seconds.

Fifty percent of reported commotio cordis events have occurred in young competitive athletes participating in a variety of amateur sports after receiving a blow to the chest wall that is usually (but not always) delivered by a projectile used to play the game. In baseball, for example, commotio cordis is often triggered when players are struck in the chest by balls that have been pitched, batted, or thrown in a variety of scenarios[20] (Table 2).

Approximately 10 to 20 cases are added to the Commotio Cordis Registry each year,[20, 23] though it was only rarely reported until the late 1990s. It is thought that this increase in number of cases is not due to an increase in incidence, but rather greater awareness following a *New England Journal of Medicine* report on commotio cordis presented in 1995,[6] with many more cases subsequently recognized. Furthermore, what was once thought to be a uniquely North American phenomenon is increasingly being reported in countries outside the United States.[24] In recent years, survival from commotio cordis has increased, likely owing to more rapid response times and access to defibrillation devices, especially during high-risk sports events, as well as greater public awareness of this condition.[25, 26]

Table.2 Example circumstances in which chest blows have triggered commotio cordis (some parts omitted or reorganized)[20]

Sports
Baseball Softball Cricket Football Soccer Hockey Lacrose
Fights and scuffles, with blow from hand or elbow
Psychiatric aide struck by patient Teacher struck while restraining student during fight Youth struck during play-shadowboxing or roughhousing Youth struck by boxing glove during sparing Child struck by parent or babysitter (with disciplinary intent) Young adult struck during slam dancing Student involved in fist fight at fraternity party Youth hit by snowball Adult struck in prison gang initiation ritual Infant struck with open hand while having diaper changed
Other circumstances
Child kicked by horse Youth hit with recoil of gun butt while deer hunting Child hit with rebound of playground swing Adult thrown against steering wheel during automobile accident Youth hit by tennis ball filled with coins Young adult kicked during cheerleading routine Adult received chest blow by falling into body of water Child received blow from head of 23-kg (50-lb) pet dog Child received blow from falling on playground apparatus Child hit by tossed hollow plastic bat Child hit by plastic sledding saucer Youth received blow intended to terminate hiccups Child hit handlebars while falling off of bicycle

Ⅲ Mechanism

Ventricular fibrillation[11, 12, 24] and complete A-V block[8, 9, 26] have been recognized prior to cardiac arrest in patients with commotio cordis. When a projectile strikes an area near the heart, the initial depolarization trigger is likely a focal phenomenon from the direct impact, similar to a premature ventricular contraction induced by a catheter, as seen in electrophysiology and cardiac catheterization laboratories. Alternatively, it could be occur after depolarization induced by changes in current flow. This 2-step process is similar to the R-on-T phenomenon, in which a premature ventricular contraction that occurs on the upslope of the T wave will cause ventricular fibrillation in acute ischemic conditions, but not in nonischemic situations.[24] In addition, that study noted that impacts occurring throughout the cardiac cycle may cause ST-segment elevation and left bundle-branch block in impact events in which ventricular fibrillation is not induced.[24]

Moderate precordial impact is particularly effective in triggering instant ventricular fibrillation if

delivered during the early T-wave of the ECG. This timing is identical to the 'vulnerable period' for electrical induction of ventricular fibrillation, initially identified in 1936,[27, 28] and raises questions as to what mechanism would: (i) allow near-instantaneous translation of a mechanical stimulus into an electro-physiologically relevant signal, which is (ii) powerful enough to cause serious rhythm disturbances and that is (iii) particularly potent in doing so during the T-wave. Obviously, that set of questions might be inappropriate, as the link between the timing of a mechanical impact and its electrophysiological effect may be primarily related to the mechanical rather than the electrical cycle of the heart.

It is not difficult to imagine that the extent of background tissue strain and/or dimensions of the cardiac chambers determine susceptibility to mechanical induction of ventricular fibrillation. Although maximum filling of the ventricles coincides with the PQ segment, that does not explain the observed peak sensitivity to mechanical induction of ventricular fibrillation during the T-wave.[29]

Causative theories regarding a predisposition to ventricular fibrillation include mechanical electrical coupling. A sudden myocardial stretch can be elicited when an external blow occurs during a vulnerable window that is based on repolarization inhomogeneity and stretch pulses applied during this vulnerable window can lead to nonuniform activation. Repolarization dispersion might play a crucial role in the occurrence of fatal tachyarrhythmia during commotio cordis.[30] In an experimental model of commotio cordis that utilized anesthetized juvenile swine, ventricular fibrillation was produced by a baseball travelling at 30 mph when the strike occurred during the vulnerable period of repolarization, on the upslope of the T-wave (Fig. 1, 2). [29] Energy from the impact object was also found to be a critical variable with baseballs travelling at 40 mph, while ventricular fibrillation was more likely with velocities less or greater than 40 mph. In addition, more rigid impact objects and blows directly over the center of the chest more often caused ventricular fibrillation. Peak left ventricular pressure generated by a chest wall blow was found to be correlated with the risk of ventricular fibrillation, with activation of the K (+) (ATP) channel a likely cause of that fibrillation.[24, 31]

Successful resuscitation is attainable with early defibrillation,[32] and initiation of ventricular fibrillation may be mediated by an abrupt and substantial increase in intracardiac pressure.[33] The energy of the chest-wall impact is an important variable in the generation of ventricular fibrillation and impacts at 40 mph were more likely to produce ventricular fibrillation than those with greater or lesser velocities, suggesting that the predilection for commotio cordis is related in a complex manner to the precise velocity of the chest-wall impact.[34] These swine studies also showed that a rapid rise in left ventricular pressure to between 250 and 450 mm Hg may mediate electrophysiological consequences, as stretch channels may be activated by a specific degree of myocardial stretch.[30-35] On the other hand, in another swine experimental model of commotio cordis, blockade of the K (+) (ATP) channel reduced the incidence of ventricular fibrillation and magnitude of ST-segment elevation, indicating that selective K (+) (ATP) channel activation may be a pivotal mechanism in sudden death resulting from a low-energy chest-wall trauma in young individuals occurring during sporting activities.[24,31]

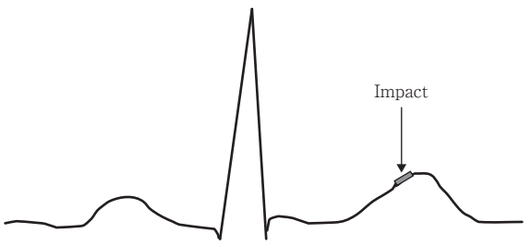

Fig. 1 Schematic representation of effects of pre-cordial mechanical stimulation on cardiac rhythm.
Kohl P et al. Cardiovasc Res 2001;50:280-289

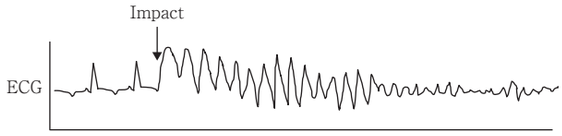

Fig. 2 ECG obtained from an anaesthetised pig subjected to a pre-cordial impact that coincided with the upstroke of the T-wave.
(Kohl P et al. Cardiovasc Res 2001;50:280-289.)

Ⅳ Protection and Prevention

Commotio cordis is a devastating cause of sudden death in young and healthy individuals, in which ventricular fibrillation is the most common arrhythmia.[23] Subsequent survival rates are dismal. Increased awareness of this phenomenon is imperative, especially among those who may be first responders, such as parents, coaches, game officials, and medical personnel. Current commercially available chest barriers are not sufficiently effective in preventing chest-blow-induced sudden cardiac death and, in fact, probably offer only a false sense of security to athletes, families, and the general public.[36] When soft safety baseballs is used, the rate of induction of VF was at its lowest with warring chest protection.[37] However, preventive chest wall official using protectors produced for baseball catchers and lacrosse players are important, though current designs are not impeccable. Safer and more appropriate protectors to be worn while playing sports are needed. We believe that Kevlar body armor with ceramic plate inserts[38] for the partial left chest wall might be able to protect commotio cordis.

Prevention may be enhanced through the use of soft "safety" baseballs and improved chest protector design. Ready availability of automated external defibrillators (AEDs) at youth sport venues may also improve survival rates, as early defibrillation improves outcomes. In 7 cases of commotio cordis reported in Japan, 6 were recovered by utilization of CPR and/or an AED. In addition, cardiopulmonary resuscitation education and first-responder awareness are important factors related to survival.[31]

Ⅴ Conclusive view

In the future perspectives of this research and clinical fields based on the contents of this article, further efforts are needed to prevent avoidable deaths by providing more education concerning commotio cordis, ideal Kevlar body armor with ceramic plate inserts for the partial left chest wall for such as baseball catchers or lacrosse goalies, and wider equipment of AEDs at every organized athletic events.

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健康運動科学「原稿執筆要領」

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【引用文献の記載例】

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5．校正

初校は著者校正とし，印刷上の誤り以外の加筆・修正・削除は認めない。

平成22年６月18日

健康運動科学「投稿規定」

1．健康運動科学について

「健康運動科学（Mukogawa Journal of Health and Exercise Science；MJHES)」（以下，本誌）は，健康運動科学研究所が発刊する科学雑誌であり，健康・スポーツ科学領域，リハビリテーション科学領域をはじめ，広く健康科学に関する研究論文などを掲載し，人々のquality of life（QoL）の向上に資することを目的とする。

2．投稿資格

本誌に投稿できるのは原則として本学教員とするが，編集委員会が必要と認めた場合には，学外からも投稿を依頼することがある。

3．原稿執筆及び種類

本誌の原稿は別掲の執筆要領にしたがって，日本語または英語で執筆する。原稿の種類は「総説」,「原著」,「短報」,「速報」,「実践研究」などとし，いずれも未発表のものに限る。ただし，論文の内容に応じて編集委員会から種類の変更を求める場合がある。

英文論文や英文抄録を含む場合は，必ずネイティブスピーカーの校閲を受けることとする。

種類の概要

総　　説（Review）：特定の研究分野に関する知見を総合的・体系的にまとめた論文。

原　　著（Original investigation）：本誌の趣旨に沿った内容で，新たな知見（独創性）を示した研究であり，なおかつ完成度が高い論文。

短　　報（Research letter）：独創的かつ研究上の価値があると思われる成績が示されており，原著に準じた体裁でまとめた論文。

速　　報（Rapid Communication）：研究上の価値があると思われる成績が示されており，方法論上の独創性を主張するために緊急を要する論文。

実践研究（Practical investigation）：実践現場での指導法・治療法に関する知見や情報をまとめた内容であり，方法・結果考察など適切に記述されている論文。

原著論文はタイトルページ（原稿執筆要項に記載)、英文抄録、Ⅰ　緒言、Ⅱ 研究対象、方法、Ⅲ　結果、Ⅳ 考察、Ⅴ　謝辞、Ⅵ　引用文献などと記載、図、図の説明文の順序で構成する。短報、速報は原則として原著論文に準ずる。

4．査読制度と論文の採否

本誌では査読制度を設ける。編集委員会は投稿された論文の内容に詳しい適任者（査読委員）を２～３名選定し，査読委員の意見を参考に論文の採否を決定する。なお，本誌に掲載された論文原稿は，原則として返却しない。

5．ヒトを対象とする研究及び動物実験に関する研究倫理基準

ヒトを対象とした研究では，「ヒトを対象とする医学研究の倫理的原則」（ヘルシンキ宣言，1964年，2002年追加）の基準に従う。また，動物実験の場合は「大学等における動物実験について」（文学情第141号，1987年）及び本学の「武庫川女子大学動物実験規程」における指針に従う。

6．論文の投稿

論文の投稿に際しては，原本１部とそのコピー（３部）及び共著者全員が投稿に同意することを示した投稿承諾書（別添）を添えて下記編集委員会宛に送付する。編集委員会は投稿原稿を受け付けた後，投稿者に投稿受理通知書を発行する。また，査読の結果，論文が受理された場合は最終の原本（図，表等を含む）１部と共に電子媒体を下記編集委員会宛に送付する。

7．掲載料

掲載料は原則無料とするが，ページの超過分については編集委員会の議を経て定める。また，写真などカラーページは別途実費を徴収する。

8．著作権

本誌に掲載された論文の著作権は，武庫川女子大学に帰属する。ただし，著作者本人は論文を許諾なしに利用することができる。また，論文は武庫川女子大学リポジトリに搭載し，インターネットを通して公開されるものとする。

—原稿の提出先—
〒663-8558　西宮市池開町6-46
武庫川女子大学健康運動科学研究所　編集委員会
Tel：(0798)45-9793

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投　稿　承　諾　書

健康運動科学　編集委員長殿

論文名 ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

上記の論文を「健康運動科学」に投稿いたします。投稿は，共著者全員の承諾の上で行われること，本論文の内容は刊行物として未発表であり，また他誌に投稿中でないこと，本誌に掲載された論文の著作権は武庫川女子大学に帰属すること，さらに論文は武庫川女子大学リポジトリに搭載し，インターネットを通して公開することに同意いたします。

年　　月　　日

筆頭著者氏名（自署） ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

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共著者氏名（自署） ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

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共著者氏名（自署） ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

（共著者が多数の場合，同紙のコピーを使用してください）

編　集　後　記

2013年度（2014年３月発行）は英文の論文のみを掲載いたしました。査読をお願いした先生方には、お忙しい中にも拘らず気持ちよくお引き受け頂きありがとうございました。この場をお借りして、お礼申しあげます。今後も研究成果の発表のため、雑誌「健康運動科学」に投稿していただきますようお願いいたします。

〔田中繁宏〕

第４巻第１号の論文で査読をお願いした先生方（敬称略）

吉　川　貴　仁（大阪市立大学）
新　井　　彩（武庫川女子大学）
伊　東　太　郎（武庫川女子大学）
小　笠　原一生（武庫川女子大学）
坂　井　和　明（武庫川女子大学）
田　中　繁　宏（武庫川女子大学）

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発行所　武庫川女子大学健康運動科学研究所
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