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Effect of altered sagittal-plane knee kinematics on loading during the early stance phase of gait

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ABSTRACT

Background: Individuals with knee osteoarthritis (OA) show various dynamic sagittal-plane changes during the early stance phase of gait. However, the effect of these kinematic alterations on knee load during the early stance remains poorly understood. Research question: The purpose of this study was to examine the effect of altered sagittal- plane knee kinematics on knee load during the early stance.

Methods: A total of 13 healthy adult men underwent gait analysis trials using four conditions (baseline and three altered conditions). The three altered conditions were defined as follows:

1) Less flexion (LF): a gait that decreased knee flexion excursion (KFE) owing to a reduced peak knee flexion angle compared to baseline.

2) Initial flexion (IF): a gait with decreased KFE owing to an increased knee flexion angle at initial contact, during which the peak knee flexion angle did not differ from baseline.

3) Flexion gait (FG): a gait that increased the knee flexion angle at initial contact but did not reduce KFE compared with the baseline.

Data analyzed included peak external knee flexion moment (KFM), KFM impulse (impulse was an integral value from initial contact to peak value), peak vertical ground reaction force (VGRF), and maximum loading rate.

Results: Both LF and IF conditions significantly decreased peak VGRF ($p < 0.05$) compared with the baseline. Peak KFM decreased in the LF condition and increased in the FG condition versus baseline ($p < 0.05$). A significantly increased KFM impulse was found in both IF and FG conditions when compared with baseline ($p < 0.05$).

Significance: An increase in knee flexion angle during early stance increased knee loading. Interventions are likely required for improving excessive knee flexion during early stance phase of gait in individuals with knee OA.

1. Introduction

Knee osteoarthritis (OA) is a common degenerative joint disease causing pain, deformity, and functional disability. Although its causes are unknown, biomechanical factors play an important role. Particularly, previous studies indicated that abnormal joint loading related to gait alterations affects disease progression [1–3].

Knee OA is more common in the medial compartment of the tibiofemoral joint [4]. Individuals with medial-compartment knee OA exhibit varus malalignment [5], which changes load distribution over medial and lateral compartments of the knee joint [6]. Thus, most studies on biomechanical characteristics of gait alterations in individuals with knee OA examined knee loading in the frontal plane. Increased external knee adduction moment (KAM) has been consistently observed in the stance phase of gait [7,8]. Moreover, increased KAM—increase in medial compressive load of the knee—contributes to progressive destruction of the articular cartilage [1].

Recently, researchers have focused on biomechanical changes of the knee in the sagittal plane during gait in individuals with knee OA [3,9]. Most studies have reported kinematic knee alterations during the early stance phase in the sagittal plane [10–12]. However, findings vary between studies. Some studies reported greater knee flexion at initial contact compared to asymptomatic individuals, although peak knee flexion at loading response did not differ [9,11]. Other studies reported

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that individuals with knee OA walk with knee flexion angle similar to asymptomatic individuals at initial contact but with less peak knee flexion on loading response [13,14]. In almost all studies, knee flexion excursion (KFE) during the early stance (range of knee flexion from initial contact to peak flexion) decreased in individuals with knee OA compared to asymptomatic individuals [8–10,12,13,15–17]. In contrast, one study suggested that KFE does not differ compared to asymptomatic individuals, despite increased knee flexion angle at initial contact [18].

Such kinematic changes of the knee during the early stance influence joint loading in the sagittal plane. However, kinetic studies on these changes have been conflicting. Some have reported that individuals with knee OA walk with a smaller peak external knee flexion moment (KFM) during the early stance [13,19], whereas others report a similar [7] or greater [20] peak KFM compared to normal. These discrepancies may be due to kinematic difference of the knee during the early stance. Baliunas et al. [7] assessed kinetics during gait in individuals with knee OA and reported that those who walked with a smaller peak knee flexion angle demonstrated decreased peak KFM during the early stance. Creaby et al. [21] also found correlation between peak KFM and peak knee flexion angle in individuals with knee OA. Thus, although previous studies have demonstrated an association between the peak knee flexion angle and KFM during the early stance, effects of different knee dynamics (i.e., different kinematic patterns from initial contact to peak flexion angle) on knee loads during gait remain unclear. In the early stance, nearly 60% of one's body weight is transferred abruptly onto the new stance limb. Consequently, different knee kinematic patterns may influence biomechanical characteristics of knee load. Therefore, effects of knee kinematic alterations on knee load during the early stance should be clarified.

We examined the effect of altered knee sagittal kinematics on knee load during the early stance phase of gait and hypothesized that knee load would be increased in gait patterns with greater knee flexion at initial contact or more flexed knee throughout the early stance.

2. Methods

2.1. Participants

Sample size was determined based on results of G*power software (Version 3.1.9.2, Heinrich Heine University, Düsseldorf, Germany). The program was set to statistical test of one-way analysis of variance with repeated measures, an alpha of 5%, beta of 80%, and an effect size of 0.25 at comparison of each variables. We determined the effect size based on Cohen's d (the 0.25 corresponded to a medium effect size). The necessary sample size was 13 volunteers. We recruited 16 healthy adult men, 3 of whom did not complete the experimental task and were excluded; finally, 13 men were included. Participants' mean age was 20.9 ± 0.9 years, height was 1.7 ± 0.1 m, and weight was 60.4 ± 6.4 kg. Exclusion criteria included lower extremity pain, nervous system and musculoskeletal disorders, and history of orthopedic surgery. All participants provided written informed consent prior to study initiation. Ethical approval was obtained from the Health Sciences University of Hokkaido (18R077070).

2.2. Gait analysis

Data collection included kinematics and kinetics of the lower limb during gait. Kinematic data were collected at 100 Hz using an 8-camera motion analysis system (MAC-3D system, Motion Analysis Co., CA, USA). We used the Helen Hayes markers set to collect kinematic data. Twenty-nine reflective markers were attached with double-sided tape to each participant's body. Kinetic data were collected at 1000 Hz using three force plates (AMTI Inc., MA, USA). Kinematic and kinetic data were time-synchronized.

2.3. Experimental protocol

Participants performed gait trials under four conditions (baseline and three altered conditions designed to mimic OA-related gait changes). Participants controlled only gait speed in the baseline condition and both gait speed and knee flexion motion during the early stance on the right side in the altered conditions. Details of altered conditions are described below.

1) Less-flexion condition (LF)

LF was defined as gait with decreased KFE created by a reduction of the peak knee flexion angle to below baseline. The knee flexion angle at initial contact was maintained unchanged from baseline.

2) Initial flexion condition (IF)

IF was defined as gait with decreased KFE created by increased knee flexion at initial contact. The peak knee flexion was maintained unchanged from baseline.

3) Flexion gait condition (FG)

FG was defined as gait with increased knee flexion angle at initial contact. KFE did not reduce from baseline. The knee flexion angle at initial contact was identical to that of the IF condition. The peak knee flexion angle increased to more than that of the baseline condition, which was necessary to prevent KFE from decreasing compared to baseline.

In the LF and IF conditions, KFE was set at 32% of baseline, established using the mean value from previous studies [8–10,12,13,15–17]. In each altered condition, participants walked while controlling knee motion on the right side using an auditory feedback system. The system comprised a feedback logger (PTS-2050, DKH Co., Tokyo, Japan) and flexible goniometer (SG150, Biometric Ltd., Newport, UK). The flexible goniometer was attached with double-sided tape to the lateral aspect of the right thigh and leg. The feedback logger was fixed to the lower back (Fig. 1). The system provided real-time auditory signal feedback by setting two threshold angles. The system emitted an auditory signal when an angle measured by the flexible goniometer was within the range of two threshold angles and stopped the signal when the angle exceeded the range of two threshold angles. We set knee flexion angle at initial contact and peak knee flexion angle during the early stance as the threshold angles in each altered condition. Details of each threshold setting are shown in Fig. 2.

Prior to data collection, participants practiced walking while controlling their knee motion in response to the auditory signal feedback during the early stance. The practice continued for at least 15 min in each altered condition. Furthermore, a physiotherapist observed the controlled knee motion and provided additional verbal feedback. Trials were measured until three successful cycles of complete foot contact at the force plate with controlled gait speed and knee motion were achieved. First, participants performed the baseline condition, followed by the three altered conditions. In each trial, participants walked at 1.0 ± 0.05 m/s. Walking speed was determined by measuring the transit time between two reference points using a stopwatch. Measurement order of altered conditions was randomized.

2.4. Data analysis

Kinematic and kinetic data were processed with Visual 3D software (C-motion Inc., MD, USA). Data from the right lower limb of all participants were analyzed. Kinematic data were applied through a low-pass filter at 6 Hz using a 4th-order Butterworth filter. For analysis, the knee flexion angle at initial contact and peak knee flexion angle during the early stance were identified, and KFE was calculated by subtracting the

Fig. 1. Experimental set up including the auditory feedback device. The flexible goniometer was attached to the lateral aspect of the right thigh and shank, and the feedback logger was fixed to the lower back. Reflective markers were also attached to the participant's body.

knee flexion angle at initial contact from the peak knee flexion angle during the early stance. The initial contact was identified as the time when magnitude of vertical ground reaction force (VGRF) exceeded 20 N. KFM and impact loading were used as surrogate measures of joint load on the knee during gait based on evidence that these measures are associated with the pathology of knee OA [3,22]. KFM was calculated using inverse dynamics and normalized to body weight and height. Peak KFM was identified during the first 50% of the stance phase. KFM impulse was calculated as the integral from initial contact to peak value. Impact loading was evaluated using peak VGRF and rate of VGRF increase (loading rate) [23]. Data for VGRF were obtained from the first half of the stance phase. Loading rate was calculated as the first-time derivative of the VGRF [23]. The maximum loading rate was used for analysis. Peak VGRF and maximum loading rates were normalized according to body weight.

2.5. Statistical analysis

Mean values of three trials for each condition were analyzed using R (Version 2.8.1, R Foundation, Vienna, Austria). Shapiro-Wilk test was used to confirm data normality. Based on normality test results, we used one-way analysis of variance with repeated measures or Friedman test. Post-hoc tests were performed using Bonferroni correction. The alpha level was set at 0.05.

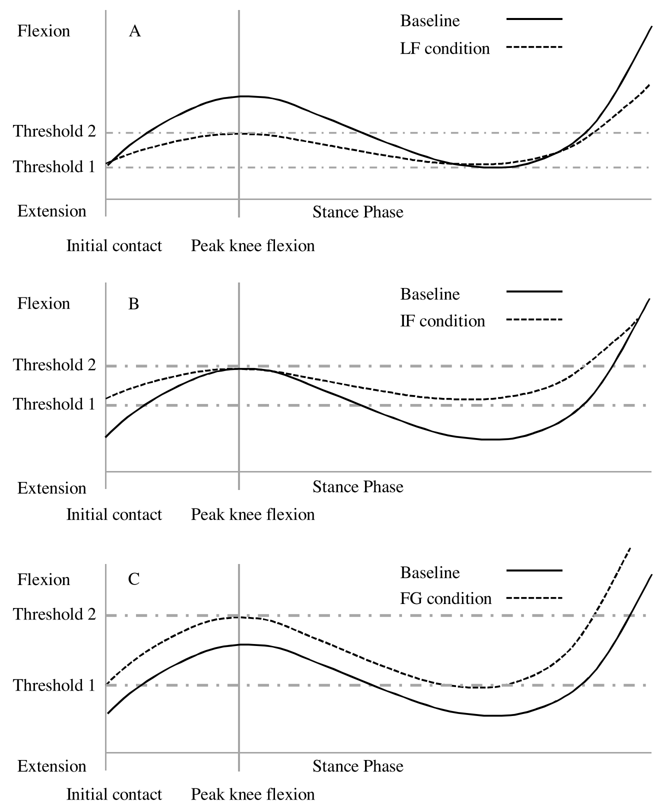

Fig. 2. Real-time auditory signal feedback at the upper and lower limits of knee flexion excursion for the knee joint angle. Knee joint kinematics in the sagittal plane and details of each threshold set in (A) less flexion condition, (B) initial flexion condition, and (C) flexion gait condition.

3. Results

3.1. Gait speed and knee kinematics

Gait speed and knee joint kinematics for each condition are represented in Table 1. Fig. 3 illustrates angular changes of the knee joint in each condition. There was no significant difference in gait speed between conditions. The LF and IF conditions significantly reduced the KFE from baseline ($p < 0.05$). In the LF condition, the peak knee flexion angle during the early stance was decreased ($p < 0.05$), although the knee flexion angle at initial contact did not differ from baseline. In the IF condition, the knee flexion angle at initial contact was significantly increased ($p < 0.05$), although the peak knee flexion angle during the early stance did not differ from baseline. No significant difference was observed in KFE between the LF and IF conditions. In the FG condition, the knee flexion angle at initial contact, peak flexion angle during the early stance, and KFE were significantly greater than those in the other conditions ($p < 0.05$).

3.2. Knee joint kinetics

Table 2 shows kinetic variables in each condition. Average VGRF and KFM patterns for the stance phase in each condition are presented in Fig. 4. Compared with the baseline condition, the LF and IF conditions significantly decreased the peak VGRF ($p < 0.05$). Although the peak VGRF increased in the FG condition, no significant difference was found from baseline. There was no significant difference between conditions at the maximum loading rate.

The LF condition significantly decreased the peak KFM from baseline ($p < 0.05$), while the FG condition significantly increased it ($p < 0.05$). However, the IF condition did not change the peak KFM from baseline. A significantly increased KFM impulse was found in the

Table 1
Kinematic variables and gait speed for each gait condition.

	Baseline (n = 13)	LF (n = 13)	IF (n = 13)	FG (n = 13)
Knee flexion angle at initial contact (°)	4.0 (2.6)‡,§	4.1 (2.8)‡,§	7.9 (3.3)*,†,§	10.3 (4.2)*,†,‡
Peak knee flexion angle (°)	14.9 (3.4)†,§	10.3 (3.4)*,‡,§	14.5 (3.8)†,§	29.7 (4.9)*,†,‡
Knee flexion excursion (°)	10.9 (1.6)†,‡,§	6.2 (1.7)*,§	6.7 (1.3)*,§	19.4 (3.9)*,†,‡
Gait speed (m/s)	1.04 (0.03)	1.03 (0.04)	1.02 (0.02)	1.02 (0.03)

Data presented as mean (standard deviation).
* Significant difference compared with a baseline.
† Significant difference compared with LF condition.
‡ Significant difference compared with IF condition.
§ Significant difference compared with FG condition.

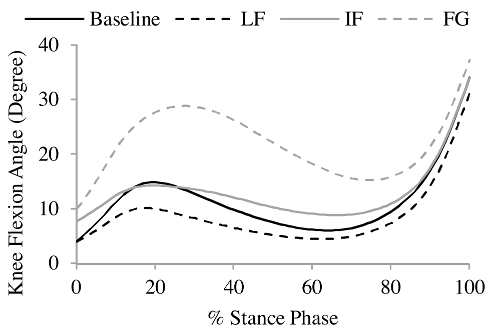

Fig. 3. The angular changes of the knee joint kinematics in each gait condition during the stance phase.

IF and FG conditions ($p < 0.05$), but in the LF condition, it remained unchanged.

4. Discussion

Although some studies have examined the relationship between knee kinematics and knee load in the sagittal plane during gait [7,13,18], none have focused on different knee dynamics during the early stance. Therefore, this study examined the effect of kinematic alternations during the early stance on knee load. As hypothesized, the knee load increased in gait patterns with greater knee flexion at initial contact or more flexed knee throughout the early stance.

Healthy adults produced OA-related alterations of knee kinematics during the early stance phase of gait. Results of kinematic variables for the altered conditions were similar to those in individuals with knee OA [9,13,18]. Unlike healthy participants, individuals with knee OA have impairments, such as muscle weakness, limited joint range of motion, and pain. However, altered gait conditions that were adopted in this study can result from these impairments. For example, limited knee extension range of motion can cause increased knee flexion during the early stance [16] and weakness of quadriceps muscle and joint instability can cause reduced KFE [10]. Therefore, our findings can help better understand the influence of altered knee flexion motion on knee load during the early stance phase of gait in individuals with knee OA.

This study showed lesser peak VGRF during the early stance in the two reduced KFE gait patterns. During the early stance, knee flexion controls vertical displacement of the body's center of mass (COM). As the VGRF comprises vertical acceleration of the body's COM, knee flexion dynamics are reflected in VGRF magnitude during this period. Greater KFE is associated with higher peak VGRF [21,24]. Our results are similar to those of previous studies. Reduced KFE probably restricts displacement of the body's COM, thereby reducing peak VGRF during the early stance. VGRF refers to impact loading on the lower extremity during gait. Peak VGRF is reduced by experimental-induced knee pain in healthy individuals [25]; hence, KFE reduction may be used as strategy to decrease impact loading to avoid pain.

Different knee flexion kinematics affected peak KFM during the early stance. Particular, peak KFM significantly decreased during LF and increased during FG, which may be associated with peak knee flexion angle during the early stance. External joint moment is determined by magnitude of resultant ground reaction force (GRF) and perpendicular distance from GRF to the joint's center of rotation, or the moment arm [26]. When the knee flexes more, the moment arm in the sagittal plane lengthens so that KFM increases. Peak knee flexion angle is positively correlated with peak KFM [7,21]. Consistent with these findings, our findings support the possibility that increasing peak knee flexion angle increases the peak KFM during the early stance.

Different knee flexion kinematics also affected KFM impulse during the early stance. Our results showed that IF and FG conditions significantly increased KFM impulse, which reflects the magnitude and duration of KFM during the early stance. In normal gait, the vector of GRF is anterior to the knee at initial contact. Next, as the knee flexes, the vector moves posterior to the knee. The external joint moment results in a change from the extension to the flexion moment during the early stance. In our study, the IF condition was set to flex the knee more at initial contact, though peak knee flexion did not differ from baseline. When knee flexion was increased at initial contact, the vector of GRF was positioned posterior to the knee earlier. Therefore, the IF condition may cause the KFM to begin earlier, thereby increase KFM duration.

Table 2
Kinetic variables for each gait condition.

	Baseline (n = 13)	LF (n = 13)	IF (n = 13)	FG (n = 13)
Peak VGRF (%BW)	104.9 (2.4)†,‡	100.0 (4.1)*,§	101.5 (3.6)*,§	111.1 (10.3)†,‡
Maximum loading rate (%BW/s)	1473.2 (245.0)	1704.2 (394.9)	1528.7 (247.6)	1494.7 (263.7)
Peak KFM (%BW × Height)	3.0 (0.7)†,§	2.2 (0.7)*,‡,§	3.1 (0.7)†,§	6.1 (1.8)*,†,‡
KFM Impulse (sec × %BW × Height)	0.2 (0.1)‡,§	0.2 (0.1)‡,§	0.3 (0.1)*,†,§	0.6 (0.2)*,†,‡

Data presented as mean (standard deviation).
VGRF, vertical ground reaction force; BW, body weight; KFM, knee flexion moment.
* Significant difference compared with a baseline.
† Significant difference compared with LF condition.
‡ Significant difference compared with IF condition.
§ Significant difference compared with FG condition.

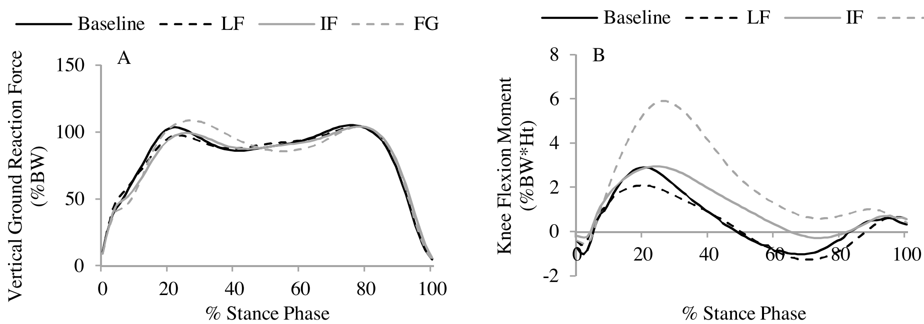

Fig. 4. Average patterns of vertical ground reaction force (A) and knee flexion moment (B) shown as waveforms, for the stance phase in each gait condition.

Increased KFM impulse under the FG condition would be due to both increase of the KFM itself and increased KFM duration (since the FG condition was set to increase knee flexion throughout the early stance).

Reduced KFE is coupled with decreasing KFM in individuals with knee OA [13,19]. Our study showed that reduced KFE decreased peak VGRF during early stance. However, peak KFM and KFM impulse changed depending on knee kinematic patterns associated with reduced KFE. Peak KFM was decreased along with reduced KFE, showing less peak knee flexion at loading response, but KFM impulse was increased alongside reduced KFE with increased knee flexion at initial contact. Thus, increased knee flexion angle during early stance may increase joint load although KFE is reduced. In addition, the greater knee flexion angle throughout the early stance phase in the FG condition resulted in both peak KFM and KFM impulse increases. Manal et al. [27] reported that KFM was positively associated with contact force of the medial tibiofemoral compartment. Chehab et al. [3] demonstrated that 1% BW*Ht increase in the baseline peak KFM caused 0.15-mm loss of medial tibial cartilage thickness over 5 years. Favre et al. [9] showed radiographically that more severe individuals with knee OA have greater knee flexion angle at initial contact than controls. Thus, greater knee flexion angle during early stance may cause increased knee load and contribute to disease progression. To prevent disease progression, interventions are needed to improve excessive knee flexion during the early stance phase of gait in clinical practice. Further longitudinal studies focusing on the effects of knee kinematic changes on disease progression are also needed.

This study had some limitations. First, healthy adults produced OA-related alterations of knee kinematics during the early stance phase of gait. It might be desirable to recruit individuals with knee OA to examine the effects of kinematic alternations on knee load during gait. However, they often use strategies to compensate for joint dysfunctions during gait [28], which can have confounding factors because they may affect knee load. Furthermore, kinetic consequences of changing knee kinematics at sagittal plane may be similar between those with and without knee OA. Hence, healthy adults were recruited. Second, the real-time auditory signal feedback system was used to control knee motion. Feedback systems include wearable sensor [29] and three-dimensional motion capture system [30]; of these, the system used in this study could easily set multiple kinematic variables simultaneously using the feedback, which is why it was adopted. However, participants found it difficult to walk while controlling knee motion at a specific value based on the auditory feedback, so standard deviations of kinematic variables were large in the altered conditions. Moreover, KFE in LF and IF conditions showed greater values compared to the set value of 32%. However, it may be applicable because variables were within those reported for OA-related alterations of knee kinematics [8–10,12,13,15–17].

5. Conclusions

This study revealed the effects of altered sagittal knee kinematics during the early stance phase of gait on knee load. Our study can aid in identifying changes in joint load associated with gait alterations and developing effective interventions to improve excessive joint load during gait in individuals with knee OA.

CRediT authorship contribution statement

Hayato Kawaji: Conceptualization, Investigation, Writing - original draft. **Satoru Kojima:** Project administration, Writing - review & editing.

Declaration of Competing Interest

None.

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