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Normal coronal kinematics of dynamic alignment and bony positions relative to the ground in three-dimensional motion analysis during gait: A preliminary study

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Abstract.
BACKGROUND: During gait, healthy knee coronal kinematics of each bony axis and lower extremity alignment are important because they could be useful as reference data for several surgeries and provide clarification of the etiology of diseases around the knee in healthy participants; however, it remains unknown.
OBJECTIVE: The objective of this study was to clarify the kinematics of lower extremity alignment and the bony axes relative to the ground during gait, focused on the coronal plane, in healthy individuals by applying our unique three-dimensional (3D) motion analysis.
METHODS: The study included 21 healthy individuals, including 9 healthy females and 12 healthy males with an average age of 36 ± 17 years. Knee kinematics were calculated in a gait analysis by combining the data from a motion-capture system and a 3D lower-extremity alignment assessment system on biplanar long-leg radiographs by using a 3D-2D registration technique. The main kinematic parameters were the dynamic position change relative to the ground, applying the femoral anatomical axis (FAA), tibial anatomical axis (TAA), and dynamic alignment in the coronal plane during the stance phase of gait.

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RESULTS: The average changes in FAA, TAA, and dynamic varus alignment were 3.7° ± 1.2°, 3.5° ± 0.8°, and 3.0° ± 1.2°, respectively. The TAA tilted laterally during the loading response and a plateau area appeared afterwards; the FAA gradually inclined laterally until the terminal stance phase, and the dynamic alignment showed varus angular change during the loading response.

CONCLUSIONS: The tibia and femur were found to change approximately 2–5° of the position of the bony axes relative to the ground. In terms of clinical relevance, our findings can be used to clarify the etiology of diseases around the knee joint and as reference data for surgeries.

Keywords: Normal knee kinematics, gait analysis, bony axis, dynamic alignment

1. Introduction

The inclination of the proximal tibia has attracted attention in the field of knee osteoarthritis (OA) as well as in surgeries such as high tibial osteotomy and knee arthroplasty [1–3]. Healthy individuals show medial inclination of the proximal tibia [4], which has been reported to become more parallel relative to the ground (tibial parallel phenomenon) in research on static alignment [5,6]. The tibial parallel phenomenon is associated with alignment changes of the lower-extremity from the non-weight-bearing condition to the weight-bearing condition [5]. During the alignment change, in addition to the tibial motion, the femoral motion is assumingly associated with varus alignment changes, especially in dynamic motions such as free gait. However, the accurate kinematics of femoral and tibial anatomical axes, as well as the dynamic alignment changes during gait, remain unknown.

Knee kinematics have been reported in many studies, which have had specific strengths and limitations [7,8]. The three-dimensional (3D) to two-dimensional (2D) registration technique [8] is accurate but can obtain measurements in a limited space and cannot analyze dynamic motions such as free gait motion. Motion analysis by the motion-capture system can evaluate the motion of the upper and lower extremities and trunk [7]; however, accurate information on the bone in terms of the dynamic motion during gait cannot be detected by commonly used systems. Our group recently developed a new method for 3D evaluation of mechanical factors in combination with a motion-capture system (VICON612; Vicon Motion Systems Ltd., UK) and a 3D lower-extremity alignment assessment system (Knee CAS; LEXI Inc., Tokyo, Japan) on biplanar long-leg radiographs by using a 3D-2D registration technique [9,10]. This system can evaluate the dynamic angular changes of the lower-extremity alignment and bony positions relative to the ground based on the coordinate system during gait.

The objective of this study was to clarify the kinematics of lower-extremity alignment and the bony axes relative to the ground in the coronal plane during gait in healthy individuals. The hypothesis was that the kinematics in the coronal plane would show changes in the varus alignment and femoral and tibial positions relative to the ground in the stance phase. For clinical relevance, if the kinematics in the coronal plane would be identified, the dynamic angle of the bony axes throughout gait for healthy participants would clarify the etiology of diseases around the knee and could also be used as reference data for several surgeries.

2. Methods

2.1. Participants

This study was approved by the ethics review board of the Niigata Institute for Health and Sports Medicine (IRB number: 22). All participants provided informed consent.

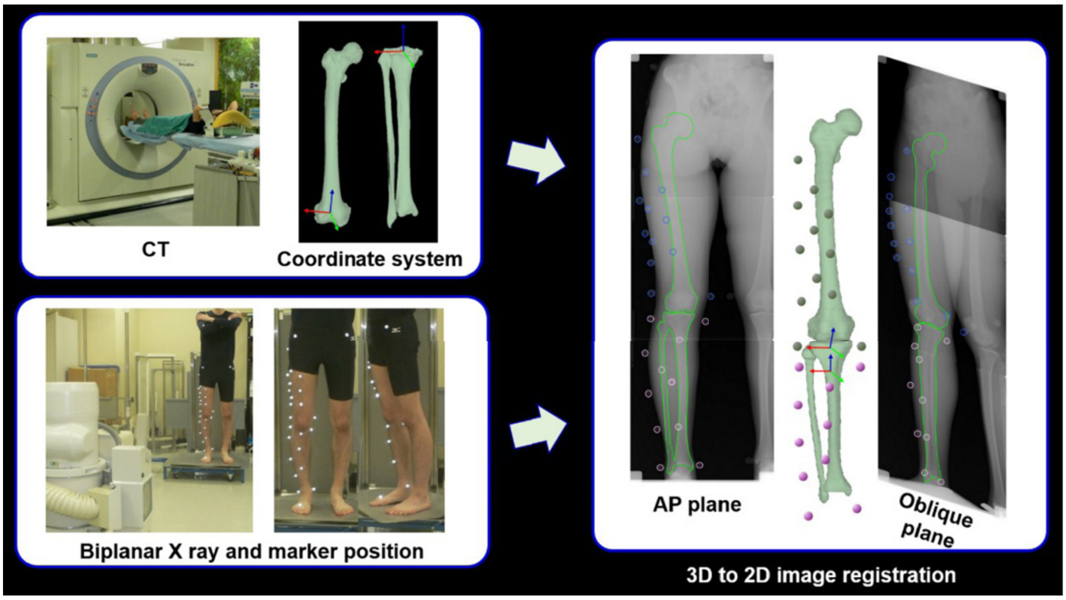

Fig. 1. Combination of motion capture system and three-dimensional lower extremity alignment assessment system.

Among the 3505 subjects who visited the clinic between September 2008 and August 2013, participants with no knee complaints and no other diseases were randomly selected. Participants with grades 0 or 1 using the Kellgren-Lawrence (K-L) classification [11] in radiography were defined as healthy participants. The study involved 21 healthy individuals, including 9 healthy females and 12 healthy males. The average age was 36 ± 17 years (95% confidence interval (CI), 28–43 years), and the average body mass index (BMI) was 22.7 ± 3.1 kg/m^2 (95% CI, 21.3–24.1 kg/m^2).

2.2. Three-dimensional lower extremity alignment assessment system

A 3D lower-extremity alignment assessment system (Knee CAS; LEXI Inc., Tokyo, Japan) using biplanar long-leg radiographs was developed to evaluate lower-extremity alignment and bony morphology. This system uses the 3D-to-2D image-registration technique and enables automatic, strict measurement of all parameters under weight-bearing conditions with high accuracy in a 3D space [12] (Fig. 1). A stereophotogrammetric X-ray apparatus, which consisted of a 0–60° turn stage and a cassette holder with three vertically arranged X-ray imaging plates, was applied. The participant stood on the turning stage, and the entire lower extremity was imaged in the 0° and 60° oblique directions using computed radiography (FCR CAPSULA, Fujifilm Co., Japan). The 3D position of the femoral and tibial bones can be estimated by superimposing the 3D skeletal models onto the bony outline of the lower extremity under weight-bearing conditions (Fig. 1) [12]. A 3D skeletal model was obtained from computed tomography (CT) scans of the entire lower extremity. The femoral and tibial coordinate systems in the 3D skeletal model were determined as described by Sato et al. [13]. The matching error of the 3D-to-2D image-registration technique was within a range of 0.68 mm in rotation and 0.5 mm in translation [12].

2.3. Gait capture and analysis

The basic methodology of this study has been reported previously [9,10]. Thirty-four skin markers were attached to each participant. For the 12 thigh markers and 10 shank markers, the original marker possessed a steel ball to detect its 2D position on X-ray images (Fig. 1). An experienced investigator attached the thigh markers to the great trochanter, medial, and lateral femoral epicondyles, and around the femoral shaft. The shank markers were placed at the medial and lateral tibial condyles, fibula head, medial and lateral malleoli, tibial tuberosity, and around the tibial shaft. The participants walked along a 12.0 m flat lane at their preferred speed. The stride time (duration of one gait cycle), stride length, and the ratio of stride length standardized by body height are presented in Table 1. A world coordinate motion-capture system (VICON612; Vicon Motion Systems Ltd., UK) was set at the center of the lane where the participants reached a constant walking speed. Their gaits were captured at a sampling rate of 120.0 Hz. The 3D positions of the markers were determined as reported previously [9,10]. For accuracy, the detection error of the 3D position of the marker was less than 0.7 mm in static conditions under a capture space of 3.0 m × 4.0 m × 2.5 m with a retroreflective skin marker with a diameter of 12.0 mm and a sampling rate of 120 Hz. The gait cycle was classified using the five time points: initial contact, foot flat, heel-rise, opposing initial contact, and toe-off. The stance phase was defined as the time from initial contact to toe-off. In the stance phase, the initial contact to foot flat was defined as the loading response phase (0–20% of the stance phase); the foot flat to heel-rise was defined as the mid-stance phase (20–50% of the stance phase); the heel-rise to opposing initial contact was defined as the terminal stance phase (50–83% of the stance phase); and the opposing initial contact to toe-off was defined as the pre-swing phase (83–100% of the stance phase). Biplanar X-ray images of the lower-extremity with the markers were obtained immediately after gait capture, and the 3D position was estimated from the 2D data of the markers on the X-ray image [9,10] (Fig. 2).

In terms of the kinematics parameters, for the femur (femoral anatomical axis: FAA), a point group centroid was automatically calculated for each of the 10 respective cross-sectional planes that divided the femoral diaphysis into 11 equal sections in the femoral coordinate system. For the tibia (tibial anatomical axis: TAA), the same calculation was performed for each of the 12 cross-sectional planes that divide the tibial diaphysis into 13 equal sections in the tibial coordinate system. The anatomical axes were determined as a regression line obtained by approximating distances from these 10 centroids in the femur and 12 in the tibia by using the least squares method. The dynamic positions of the “FAA” and “TAA” during the stance phase of the gait cycle were calculated. Each bony position relative to the ground was defined as the position of the FAA and TAA relative to the z-axis of the world coordinate system in the coronal plane (lateral inclination: +). The world coordinate system was defined as follows: the y-axis was gait direction, the z-axis was gravity direction, and the x-axis was the cross product of the y- and z- axes. The kinematics parameters of the lower-extremity alignment were presented as the “dynamic alignment”, which was determined by evaluating the association between the FAA and TAA in the coronal plane of the world coordinate system. Positive values of dynamic alignment indicated varus alignment (Fig. 3). The kinematics of each parameter were assessed by applying the location and the translation defined as the change from the initial position.

The static femorotibial angle (FTA) was defined as the angle between the FAA and the TAA projected onto the coronal plane in the femoral coordinate system (Fig. 3). A larger FTA indicates a larger varus alignment.

Table 1
Demographic data

Variables	All (n = 21)			Female (n = 9)			Male (n = 12)			Female vs Male
	mean ± SD	95%CI		mean ± SD	95%CI		mean ± SD	95%CI		p value
Age, years	36 ± 17	28	43	40 ± 14	30	51	32 ± 18	20	43	.247
Body weight, kg	63.0 ± 13.7	56.8	69.3	50.8 ± 4.1	47.6	53.9	72.2 ± 10.6	65.5	79.0	<.001*
Body height, cm	165.9 ± 9.6	161.5	170.2	156.2 ± 4.5	152.7	159.6	173.1 ± 4.3	170.4	175.9	<.001*
Body mass index, kg/m^2	22.7 ± 3.1	21.3	24.1	20.8 ± 1.6	19.6	22.1	24.1 ± 3.3	22.0	26.2	.002*
Stride time, s	1.0 ± 0.1	1.0	1.0	1.0 ± 0.1	0.9	1.0	1.0 ± 0.1	1.0	1.1	.178
Stride length, cm	138.1 ± 15.7	130.9	145.2	136.4 ± 10.1	128.7	144.1	139.3 ± 19.2	127.1	151.5	.656
Ratio of stride length, %	83.4 ± 9.0	79.3	87.5	87.3 ± 5.0	83.4	91.2	80.5 ± 10.4	73.9	87.1	.065
FTA, °	178.0 ± 3.6	176.4	179.7	175.9 ± 2.2	174.3	177.6	179.6 ± 3.7	177.2	181.9	.017*

n = knees; SD = standard deviation; 95%CI = 95% confidence interval; Stride time = time in one gait cycle; Ratio of stride length = stride length/body height; FTA = femorotibial angle; * = ≤0.05.

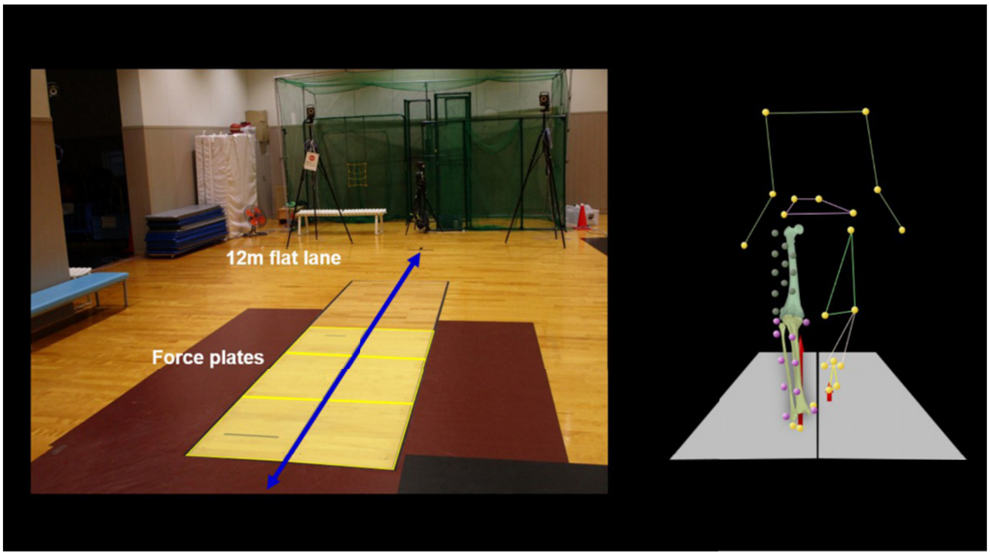

Fig. 2. Gait capture and analysis.

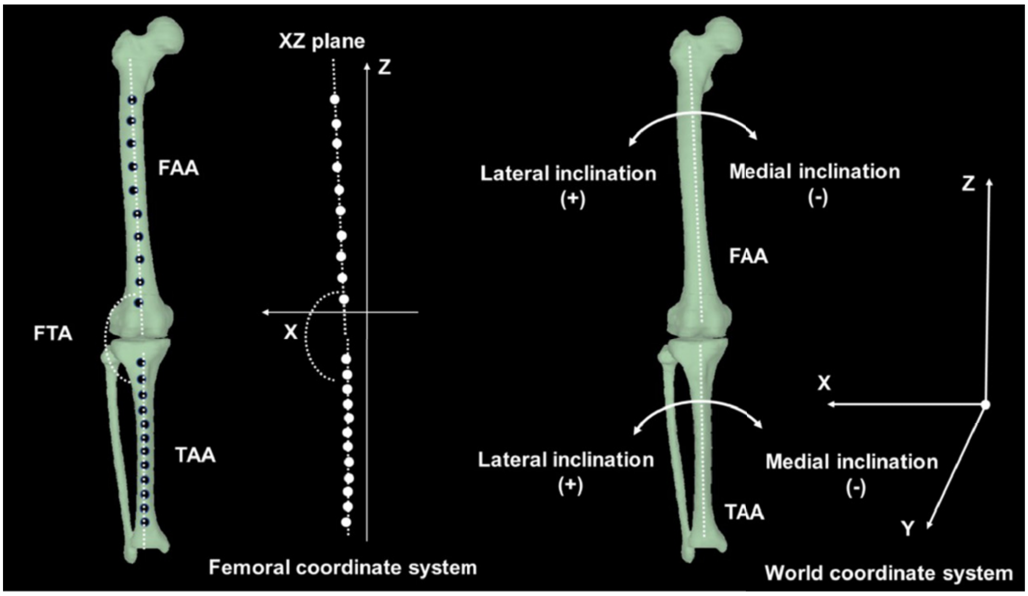

Fig. 3. Assessment parameters. FTA = femorotibial angle; FAA = famoral anatomical axis; TAA = tibial anatomical axis.

2.4. Statistical analysis

The Shapiro-Wilk test was used to examine the normality of the data. In the comparison of the difference between the females and males (Tables 1 and 2), the two-sample t-test was applied for data with a normal distribution and equal variance. The Welch test was applied for data with a normal distribution without equal variance, and the Mann-Whitney U test was used for data without a normal distribution.

Table 2
Kinematic parameters

Variables	All (n = 21)			Female (n = 9)			Male (n = 12)			Female vs Male
	mean ± SD	95%CI		mean ± SD	95%CI		mean ± SD	95%CI		p value
Minimum FAA, °	−0.9 ± 2.1	−1.8	0.1	−0.3 ± 1.9	−1.7	1.2	−1.3 ± 2.1	−2.7	0.0	.242
Maximum FAA, °	2.8 ± 1.7	2.0	3.6	3.0 ± 1.4	2.0	4.1	2.6 ± 2.0	1.3	3.9	.575
Change of FAA, °	3.7 ± 1.2	3.1	4.2	3.3 ± 1.0	2.5	4.1	3.9 ± 1.3	3.1	4.8	.236
Minimum TAA, °	2.3 ± 1.8	1.5	3.1	1.7 ± 1.1	0.8	2.5	2.8 ± 2.0	1.5	4.1	.154
Maximum TAA, °	5.8 ± 2.0	4.9	6.7	5.0 ± 0.9	4.4	5.7	6.3 ± 2.4	4.8	7.9	.099
Change of TAA, °	3.5 ± 0.8	3.1	3.8	3.3 ± 0.8	2.8	3.9	3.5 ± 0.8	3.0	4.0	.917
Minimum dynamic alignment, °	1.7 ± 3.0	0.4	3.1	0.6 ± 2.4	-1.2	2.4	2.6 ± 3.3	0.5	4.7	.141
Maximum dynamic alignment, °	4.7 ± 2.9	3.4	6.1	3.6 ± 2.7	1.5	5.7	5.6 ± 2.8	3.8	7.4	.132
Change of dynamic alignment, °	3.0 ± 1.2	2.4	3.6	3.0 ± 0.9	2.3	3.7	3.0 ± 1.5	2.0	3.9	.945

SD = standard deviation; 95%CI = 95% confidence interval; FAA = femoral anatomical axis; TAA = tibial anatomical axis; * = ≤0.05.

Table 3
Comparison of overall kinematics using repeated measures ANOVA

Variables	Female vs Male
FAA (Graph 4)	
Overall kinematics (Location)	0.611
Overall kinematics of the change from the initial position (Translation)	0.030*
TAA (Graph 5)	
Overall kinematics (Location)	0.168
Overall kinematics of the change from the initial position (Translation)	0.947
Dynamic alignment (Graph 6)	
Overall kinematics (Position)	0.220
Overall kinematics of the change from the initial position (Change)	0.050*

ANOVA = analysis of variance; FAA = femoral anatomical axis; TAA = tibial anatomical axis; * = ≤0.05; Graphs are shown in figures.

The differences in overall kinematics were assessed by repeated measures analysis of variance (ANOVA) (Table 3). Statistical significance was set at $p \leq 0.05$, using SPSS software (version 27; IBM Corp., Armonk, NY, USA).

3. Results

In assessments of the demographic data (Table 1), body weight, body height, BMI, and FTA were different between the female and male participants. The gait parameters of stride time, stride length, and stride length ratio were not different between the sexes.

The average changes in the kinematic data for the overall study population, females, and males were 3.7° ± 1.2°, 3.3° ± 1.0°, and 3.9° ± 1.3° of lateral inclination in the FAA; 3.5° ± 0.8°, 3.3° ± 0.8°, and 3.5° ± 0.8° of lateral inclination in the TAA; and 3.0° ± 1.2°, 3.0° ± 0.9°, and 3.0° ± 1.5° of varus change in dynamic alignment, respectively, with no differences between the sexes, 2–5° relative to the ground (Table 2).

In the assessment of femoral kinematics (Fig. 4), the translation demonstrated a similar pattern, but showed differences in magnitude between the sexes ($p = 0.030$) (Table 3). The femur gradually inclined laterally until the terminal stance phase and subsequently inclined medially in both groups.

In the assessment of tibial kinematics (Fig. 5), there were no differences in the location and translation between females and males (location, $p = 0.168$; translation, $p = 0.947$). The TAA in both sexes showed the greatest changes in the lateral inclination at the loading response and the plateau area in the subsequent phase (Table 3).

The dynamic alignment exhibited a similar pattern but different magnitudes of translation ($p = 0.050$) (Table 3) (Fig. 6).

4. Discussion

The most important findings of this study in relation to the kinematics in the coronal plane during gait were as follows: (1) the tibia was acutely inclined laterally at the loading response and the plateau area at the subsequent phase, the femur was gradually inclined laterally until the terminal stance phase, and

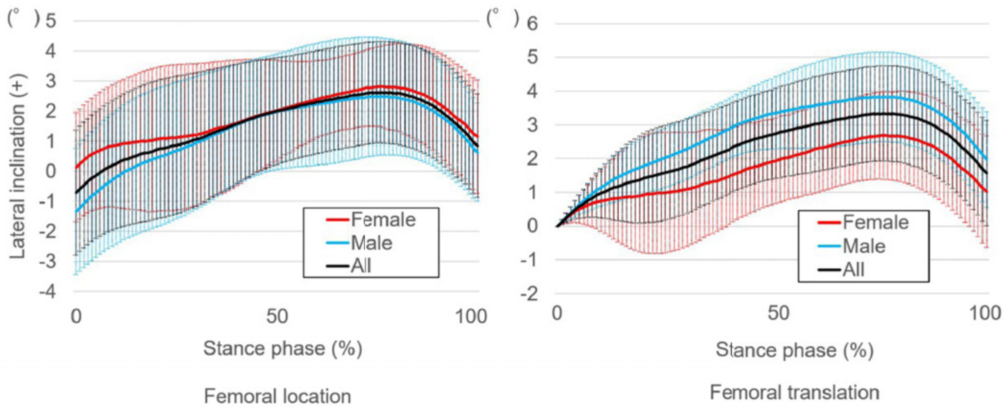

Fig. 4. Gait cycle (stance phase) – femoral anatomical axis (FAA) angle graphs. Femoral location meant the angle of the FAA relative to the ground that was not standardized by the initial position, while femoral translation meant the angle standardized by the initial position in the world coordinate system.

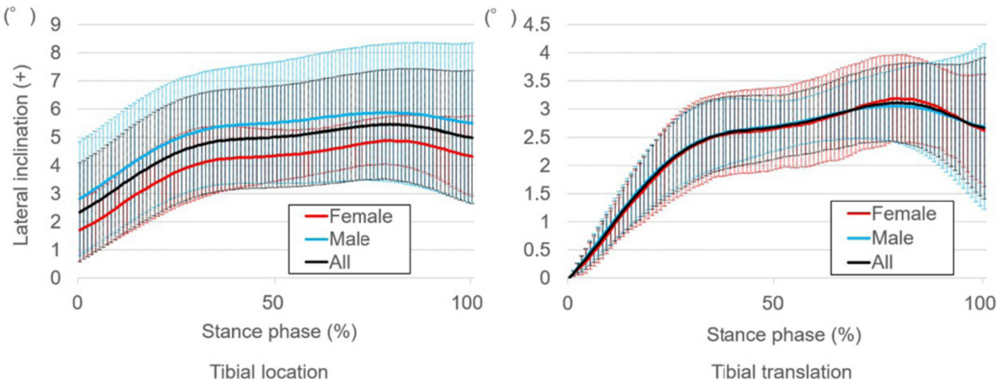

Fig. 5. Gait cycle (stance phase) – tibial anatomical axis (TAA) angle graphs. Tibial location meant the angle of the TAA relative to the ground that was not standardized by the initial position, while tibial translation meant the angle standardized by the initial position in the world coordinate system.

the dynamic varus alignment change occurred at the loading response; (2) the tibia, femur, and dynamic alignment changed around 2–5° relative to the ground; and (3) coronal kinematics, especially in the femur, showed sex-related differences.

The TAA was inclined laterally at the loading response and the plateau area subsequently appeared with no sex-related differences, showing nearly the same motion during the stance phase. This motion is probably produced by the morphology, alignment, ligament laxity, KAM, and several other factors [5,6,9,14–16]. The normal tibia has been reported to have a medial inclination [9], which is approximately 5°, using the condylar plateau angle between the TAA and the line connecting the medial and lateral edges of the proximal tibia [14]. With regard to medial-lateral laxity, in normal knees, lateral and medial ligamentous laxities were not balanced, and more lateral (5°) than medial ligamentous laxity (2°) has been reported [15]. As a dynamic mechanism, KAM [9] has been reported to force the lower-extremity alignment and tibia into varus (tibial lateral inclination) [17,18], the direction of which implies that the

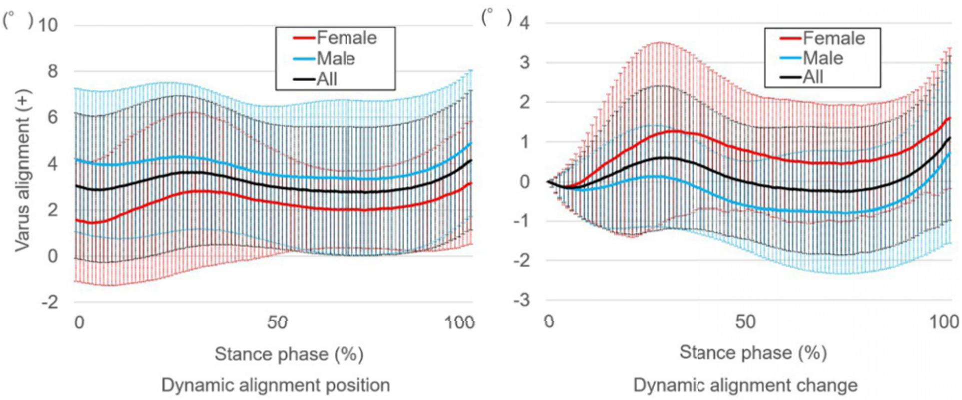

Fig. 6. Gait cycle (stance phase) – dynamic alignment angle graphs. Dynamic alignment position meant the angle that was not standardized by the initial position, while dynamic alignment change meant the angle standardized by the initial position in the world coordinate system.

tibial articular surface became more parallel [5,6]. The absence of sex-related differences in the dynamic tibial motion of 2–5° in this study can be attributed to the rational angle produced by several factors like those mentioned above because knee motion is assumingly produced within the medial-lateral laxity (around 5°) to make the inclined articular surface (around 5°) more parallel to the ground. To realize stable bipedal locomotion [19], the distal articular surface in the loaded joint holding the proximal articular surface plays the important role of aligning parallel to the ground to achieve its antigravity action. This dynamic motion study suggested that, at the knee joint in normal knees, the tibial mechanism making the articular surface more parallel to the ground is crucial for maintaining balance and realizing bipedalism [5].

In this study, the femur was gradually inclined laterally until the terminal stance phase, and the inclination showed sex-related differences. As a result of the femoral and tibial motion, the dynamic alignment change at the loading response showed sex-related differences. Regarding femoral motion, the proportion of individuals using a postural control strategy that primarily uses the hip joint is reportedly greater than that using the ankle joint strategy [18]. In actual ground contact, the distal part of the lower extremity is first placed on the ground, in the order of the foot, tibia, and femur, so that the proximal part of the leg performs a compensatory movement relative to the distal part of the leg. The dynamic bony change in this study demonstrated that the femur was laterally inclined (hip adduction) to achieve single-leg standing balance relative to the acute tibial medial inclination during the loading response in the stance phase, which showed a sex-related difference in the femoral motion. Many studies have reported sex-related differences in gait analysis [20–24], and larger sex-related differences in the joint motion in the frontal and transverse planes have also been reported [23]. The present study showed larger values for body weight, body height, BMI, and FTA in males than in females. In general, forces acting on the knee joint are higher in men because of the higher body weight [23]. The larger forces in the knee joint as a result of differences in body composition, bony anatomy, alignment, muscle strength, joint laxity, and other factors might result in larger compensatory motion in the femur for males than females.

The dynamic alignment exhibited an acute change at the loading response and plateau area in the subsequent phases of this study. The varus thrust is expressed as a momentary sideways movement of the knee [25,26]. Sharma et al. [27] noted that varus thrust visualized throughout gait is associated with knee OA progression and should be a target of intervention development, but the causes and mechanisms remain unclear. In an actual clinical setting, we also have, on occasion, experienced varus thrust for normal participants during gait [28]. Theoretically, the mechanism of varus thrust can be determined by analyzing the changes in the alignment and relative position of the bones by motion analysis. Since the KAM at the loading response is the largest in the stance phase [9], the varus thrust is more likely to be observed at the loading response. Considering the acute changes in the dynamic alignment at the loading response and the plateau area (tibial parallel area) in the subsequent phases, the varus thrust may itself be a form of the dynamic alignment change with a larger magnitude at the loading response. The previous study showed that the lower-extremity alignment to the ground was varus under weight-bearing conditions, and this phenomenon was produced mainly by the tibia rather than the femur [5]. In this study, the tibia caused knee varus motion at the loading response, whereas the femur gradually showed hip adduction to maintain single-leg standing balance until the terminal stance phase. The tibial varus motion at the loading response may be the main cause of varus thrust, but this is beyond the scope of this study. Further research is required to elucidate this aspect in the future.

This study had several limitations. First, the sample size of this preliminary study was relatively small. In the future, the sex-related characteristics should be clarified by a study with the large sample size in terms of the relationship between the coronal motion and factors as bony morphology and strength, muscle, body composition, and other factors. Second, this study included selection bias. The measurement is time-consuming and participants' availability is limited; hence, informed consent is relatively difficult to obtain.

5. Conclusions

The tibia was acutely inclined laterally at the loading response and the plateau area in the subsequent phase. The femur was gradually inclined laterally until the terminal stance phase and dynamic varus alignment change occurred at the loading response. The tibia, femur, and dynamic alignment changed by approximately 2–5° relative to the ground. Coronal kinematics, especially in the femur, showed sex-related differences.

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Conflict of interest

None of the contributing authors have any conflict of interest to declare.

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