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Biomechanical Characteristics of Long Stair Climbing in Healthy Young Individuals in a Real-World Study Using a Wearable Motion Analysis System

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Abstract: Background: Stair climbing is a part of the basic activities of daily living. Previous biomechanical analyses of stairs have been conducted in the laboratory, resulting in only a few steps. Therefore, the biomechanical characteristics of long stair climbing in the real world remain unclear. The purpose of this study was to identify differences in kinematic and kinetic in the lower limb between the beginning and end phases of long stair climbing in an outdoor environment using a wearable motion analysis system. Eight subjects (four males and four females) were included in the data analysis (age: 23.6 ± 0.5 years). The long stair was 66 consecutive steps out of 202 stone steps. A wearable motion analysis system comprised six inertial measurement units and foot pressure sensors. The maximum ankle joint flexion angle in the end phase was significantly increased more than in the beginning phase ($p < 0.001$). On the other hand, the other kinematic, kinetic, and stair climbing speeds showed no significant difference between the phases. The findings indicated that fatigue during long stair climbing might increase ankle dorsiflexion to compensate for forwarding propulsion.

Keywords: long stair climbing; wearable motion analysis system; real-world motion analysis

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1. Introduction

Stair climbing is one of the most fundamental activities of daily living and is necessary for independent living at home and in the community. Stair climbing is more difficult than level walking, as evidenced by previous biomechanical studies [1,2]. Mechanical load on the lower limb, energy cost, and dynamic instability were significantly higher during stair climbing than at level walking [1,2]. Elderly people who have difficulty climbing stairs have a smaller life-space area related to their quality of life than those who have no difficulty climbing stairs [3]. Furthermore, most patients with early to moderate osteoarthritis of the lower extremities, which has a high prevalence in elderly people, report difficulty climbing stairs as their first complaint [4].

Understanding the biomechanical characteristics of stair climbing is important to conducting rehabilitation and design for public stairs. Many previous studies have conducted a biomechanical analysis of stair climbing in the indoor or laboratory setting so that the number of steps during stair climbing was from three to sixteen steps [5–7]. On the other hand, in general, a greater number of steps in existing homes and public buildings, (i.e., real-world) is observed, (e.g., 30 or more steps on stairs in a train station building) compared to the laboratory setting in the previous study. Prolonged walking has been shown to alter spatio-temporal parameters, muscle activity, trunk acceleration, the variability of gait rhythm, and muscle oxygenation due to fatigue [8–10]. Therefore, the kinematics and kinetics in the lower limb may also differ between short and long stair climbing. Kretz et al. reported that the mean upward walking speed on the short stairway was rough, twice as large as the one on the long stairway [11]. However, to our knowledge, no biomechanical studies have examined the characteristics of long stair climbing.

In recent years, inertial measurement unit (IMU)-based motion analysis systems have been widely used due to their low cost, comfort, portability, and user-friendliness. Several previous studies have reported that the validity of IMU-based assessments of human movement characteristics depends on the complexity of the task, the intensity of the movement, the placement of the sensors, the specific movement parameters to be analyzed, and the processing method used [12–16]. Teufl and Miezal [12] showed that the root mean square error (RMSE) and range of motion error (ROME) of joint angles in the sagittal plane was less than 1° between IMU-based systems and optical motion capture. Park and Yoon [13] reported that a comparison of gait analysis between the IMU and optical motion capture using statistical parametric mapping analysis showed no significant difference in the hip joint but significant differences in the knee and ankle joints during the swing phase. Biomechanical analyses of stair climbing have traditionally been conducted in a laboratory environment using optical motion capture cameras and force plate systems [17–20]. While these systems provide accurate and reliable biomechanical data on stair climbing, these systems are expensive and labor-intensive. In addition, the measurement of stair climbing using optical motion analysis systems is feasible only in a limited environment, such as a laboratory. Therefore, these systems may not be able to measure the stair climbing movement in the real world entirely. The previous study demonstrated that a wearable motion analysis system could measure ground reaction force and trajectories of the center of pressure in healthy individuals during level walking with the same accuracy as an optical motion analysis system [21–23]. Furthermore, we also demonstrated that a wearable motion analysis system using inertial sensors and load cells could measure ground reaction force and trajectories of the center of pressure in healthy individuals during stair climbing with the same accuracy as the optical motion analysis system [24]. This system can accurately measure long stair climbing in an outdoor environment.

The purpose of this study was to identify differences in kinematic and kinetic in the lower limb between the beginning and end phases of long stair climbing in an outdoor environment using a wearable motion analysis system. We hypothesize that the kinematic and kinetic characteristics in the lower limb would differ between the beginning and end phases of long stair climbing due to fatigue during long stair climbing.

2. Materials and Methods

Participants were recruited as healthy adults who consented to participate in the study. The inclusion criteria were as follows: (1) from 20 to 40 years old and (2) able to continuously ascend and descend 200 stairs. Exclusion criteria were (1) current pain in lower limb joints, (2) a history of previous surgery on the lower limb, and (3) abnormal circulatory and respiratory status. All participants provided informed consent, and the Institutional Review Board approved this study at Tohoku University (approval ID: 2018-1-553). It was conducted according to the principles of the Declaration of Helsinki.

Measurements were taken in the precincts of Shiogama Shrine (Figure 1A), located in Matsuyama, Shiogama City, Miyagi Prefecture. Permission to conduct the measurements was obtained from the administration office of Shiogama Shrine before the measurements. The Shiogama Shrine is one of the representative shrines in Miyagi Prefecture, visited by 450,000 people annually. The measurement site was Omotesaka, one of the Omotesando approaches at the shrine, consisting of 202 stone steps (Figure 1B). The stair used for the measurement was 66 consecutive steps out of 202 stone steps. The dimensions of the stair used for this study had a riser height 15.6 $\pm$ 2.5 cm, a tread of 58.4 $\pm$ 8.2 cm, and a slope of 14.8 $\pm$ 2.5 degrees.

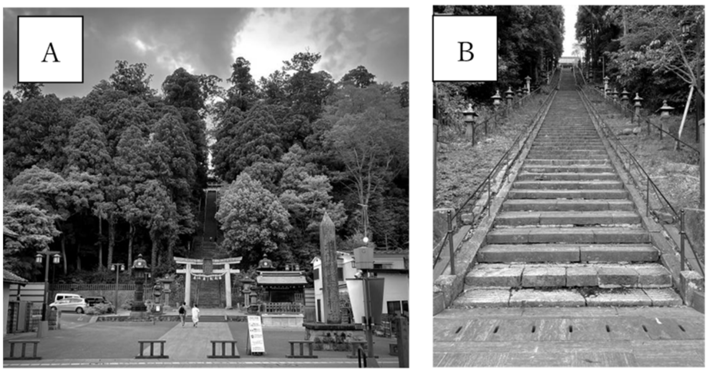

Figure 1. Shiogama Shrine. (**A**). Shiogama Shrine; (**B**). Omotesaka: 202 stone steps.

The wearable motion analysis system consists of six IMUs (TSND121, ATR-Promotions, Japan, Figure 2). IMUs were attached to the middle of the participant's upper leg, lower leg, and feet with elastic bands (Figure 3). Each IMU recorded acceleration and angular velocity in three axes at a sample rate of 250 Hz. In addition, the foot pressure sensors (Balance Aid, Leimac, Japan) were used to estimate the center of foot pressure and ground reaction force (Figure 2B). Four sections, (i.e., forward-left, forward-right, backward-right, and backward-left), of foot pressure data were recorded at a sampling rate of 50 Hz.

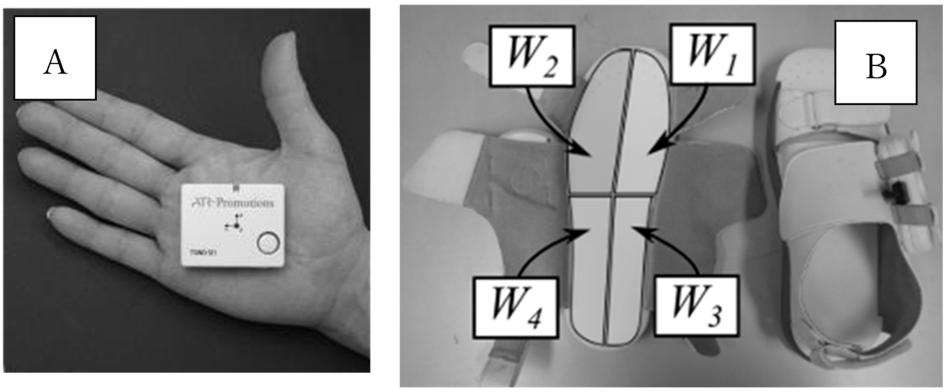

Figure 2. Measurement device. (**A**). The Inertial Measurement Unit; (**B**). The foot pressure sensors; W1: forward-right, W2: forward-left, W3: backward-right, W4: backward-left.

Figure 3. The wearable motion analysis system.

The measurement task consisted of stair climbing 66 consecutive steps. Three trials were conducted in total, with sufficient rest periods between trials. The analysis section was divided into the beginning and end climbing phases to examine the kinematic and kinetic characteristics of the lower limbs during long stair climbing. The beginning phase consisted of the first three to seven stairs, and the end phase consisted of the last five to nine stairs. The stair climbing speed was defined as the level at which the subjects could ascend the stairs comfortably without stopping on the steps. The stair climbing cycle can be divided into stance and swing phases. The stance phase is divided into the weight acceptance phase, the pull-up phase, and the forward movement phase. The weight acceptance response phase is the period from the initial contact of the reference leg to the release of the contralateral leg. The pull-up phase is the period from the release of the contralateral leg to contact of the same leg with the following front step. Finally, the forward movement phase is the period from the contact of the contralateral leg with the floor to the contact of the reference leg with the following front step [25].

Acceleration and angular velocity data obtained from each IMU during stair climbing were resampled from 250 to 50 Hz and synchronized with foot pressure data obtained from the foot pressure sensors using a custom LabVIEW (National Instruments, Austin, TX, USA) application that provided the graphical interfaces to align IMU signals and foot pressure signals. Second, according to previous studies, the software automatically detected the event of foot contact and toe-off [26,27]. Third, the hip, knee, and ankle joint angles in the sagittal plane were calculated based on the angular velocity data of each segment during each stair climbing cycle. Forth, based on these kinematic data and the foot pressure sensor data, the ground reaction force and center of pressure were calculated using an estimation model [24]. Fifth, the musculoskeletal modeling software (OpenSim 3.0) calculated the kinematic and kinetic data. OpenSim is a software that creates dynamic motion simulations and calculates moments at each joint by inverse dynamics [28]. After entering the study participants' age, height, and weight and defining their body scaling, the obtained kinematic data, ground reaction force data, and center of pressure were input to calculate the joint moments for each stair climbing cycle. Only the left lower limb was analyzed in this study.

To clarify the kinematic characteristics of the lower limb during long stair climbing, the following parameters were analyzed. The spatial-temporal parameter was stair climbing speed. The kinematic parameters were a hip joint angle, knee joint angle, and ankle joint angle on the sagittal plane. During the stair climbing cycle, the maximum and minimum values were extracted as representative values. The kinetic parameters were hip joint moment (+: extension moment, −: flexion moment), knee joint moment (+: extension moment, −: flexion moment), and ankle joint moment (+: plantarflexion moment, −: dorsalflexion moment). During the stair climbing cycle, the maximum and minimum values were extracted as representative values. The body weight of the participants normalized the kinetic parameters.

A pulse oximeter (The Masimo Rad-57 Pulse CO-Oximeter, Masimo Inc., Irvine, CA, USA) was used to measure pulse rate before and after stair climbing in order to observe physical fatigue during stair climbing.

Statistical analysis was performed to compare the beginning and end phases of long stair climbing. First, the Shapiro–Wilk test of normality was performed. Then, the minimum hip joint angle and maximum hip joint maximum flexion moment were not normal distributions; therefore, the Wilcoxon signed-rank test was performed. The other parameters were a normal distribution; therefore, the paired *t*-test was performed. The significance level was set at less than 0.05. The statistical analysis was performed using SPSS (version 22, SPSS Inc., Chicago, IL, USA) statistical analysis software.

3. Results

The study included ten participants. Two of them had missing data and were excluded from the data analysis. Eight subjects (four males and four females) were included in the data analysis (age: 23.6 ± 0.5 years, height: 165.6 ± 8.1 cm, and weight: 62.2 ± 6.4 kg).

The stair climbing speed was 81.6 ± 6.9 and 82.3 ± 6.3 steps/min in the beginning and end phases, respectively. There was no significant difference in the speed ($p = 0.557$) (Table 1). The pulse rate was 83.9 ± 11.9 and 143.9 ± 18.3 bpm before and after stair climbing, respectively. There was a significant difference in the pulse rate ($p < 0.001$) (Table 2).

Table 1. Stair climbing speed.

	Stair Climbing Speed (Steps/min)		
	Mean	**SD**	***p*-Value**
The beginning phase of stair climbing	81.6	6.9	0.557
The end phase of stair climbing	82.3	6.3	

SD: standard deviation.

Table 2. The pulse rate before and after stair climbing.

	Pulse Rate (bpm)		
	Before Stair Climbing	**After Stair Climbing**	***p*-Value**
Mean	83.9	143.9	<0.001
SD	11.9	18.3	

SD: standard deviation, bpm: beat per minute.

The kinematic data of the hip, knee, and ankle joints in the sagittal plane during the stair climbing cycle were represented in Figure 4. The maximum ankle joint flexion angle was significantly more significant in the end phase than in the beginning phase ($p < 0.001$, Table 3). The maximum ankle dorsiflexion angle trend in the sagittal plane during stair climbing was represented in Figure 5. On the other hand, the other kinematic parameters did not show significant differences between phases (Table 3).

Table 3. The sagittal plane kinematic data of hip, knee, and ankle during stair climbing.

ID	Hip Joint Angle (Degree)			
	Maximum		**Minimum**	
	Beginning Phase	**End Phase**	**Beginning Phase**	**End Phase**
1	58.94	55.53	−2.14	−2.43
2	70.70	69.62	−5.80	−4.94
4	64.87	65.16	−6.74	−6.60
5	59.56	60.93	−6.06	−4.02
6	51.38	47.60	−4.88	−8.06
7	61.56	60.69	−2.86	−1.45
8	77.45	76.95	8.10	11.08
9	69.67	70.34	−1.66	−0.01
Mean	64.26	63.35	−2.75	−2.06
SD	7.66	8.65	4.47	5.55
95% CI	−2.46, 0.63		−0.88, 2.27	
p-value	0.21		0.31	

Table 3. *Cont.*

ID	Knee Joint Angle (Degree)			
	Maximum		Minimum	
	Beginning Phase	End Phase	Beginning Phase	End Phase
1	71.96	67.09	19.38	19.21
2	78.75	77.96	20.66	22.73
4	86.00	87.47	12.11	16.87
5	82.70	79.27	15.80	15.05
6	89.61	88.00	21.18	20.93
7	90.67	89.57	23.63	26.32
8	98.29	99.90	27.02	35.01
9	100.73	99.64	28.64	26.02
Mean	87.34	86.11	21.05	22.77
SD	9.01	10.40	5.13	5.95
95% CI	−3.06, 0.61		−1.15, 4.57	
p-value	0.16		0.20	

ID	Ankle Joint Angle (Degree)			
	Maximum		Minimum	
	Beginning Phase	End Phase	Beginning Phase	End Phase
1	18.77	21.03	−32.18	−23.96
2	31.54	36.82	−18.62	−13.50
4	22.78	28.26	−21.25	−16.95
5	24.97	25.69	−27.45	−29.99
6	31.99	38.22	−6.04	−7.09
7	28.12	34.73	−14.90	−15.30
8	27.46	36.14	−7.84	−6.85
9	37.49	42.16	−6.64	−2.30
Mean	27.89	32.88	−16.86	−14.49
SD	5.48	6.69	9.19	8.65
95% CI	2.90, 7.08		−0.71, 5.45	
p-value	<0.001		0.11	

SD: standard deviation, CI: Confidence interval.

The vertical and horizontal bars represent the maximum ankle dorsiflexion angle and the steps of the stair, respectively. The number of steps in the stair is 33, as only the left side was analyzed. The black lines represent the mean maximum ankle dorsiflexion angle during stair climbing. The gray shades of vertical bars represent the standard deviation during the mean maximum ankle dorsiflexion angle during stair climbing.

The hip, knee, and ankle kinetic data during the stair climbing cycle were represented in Figure 6. The kinetic parameters showed no significant differences between phases (Table 4).

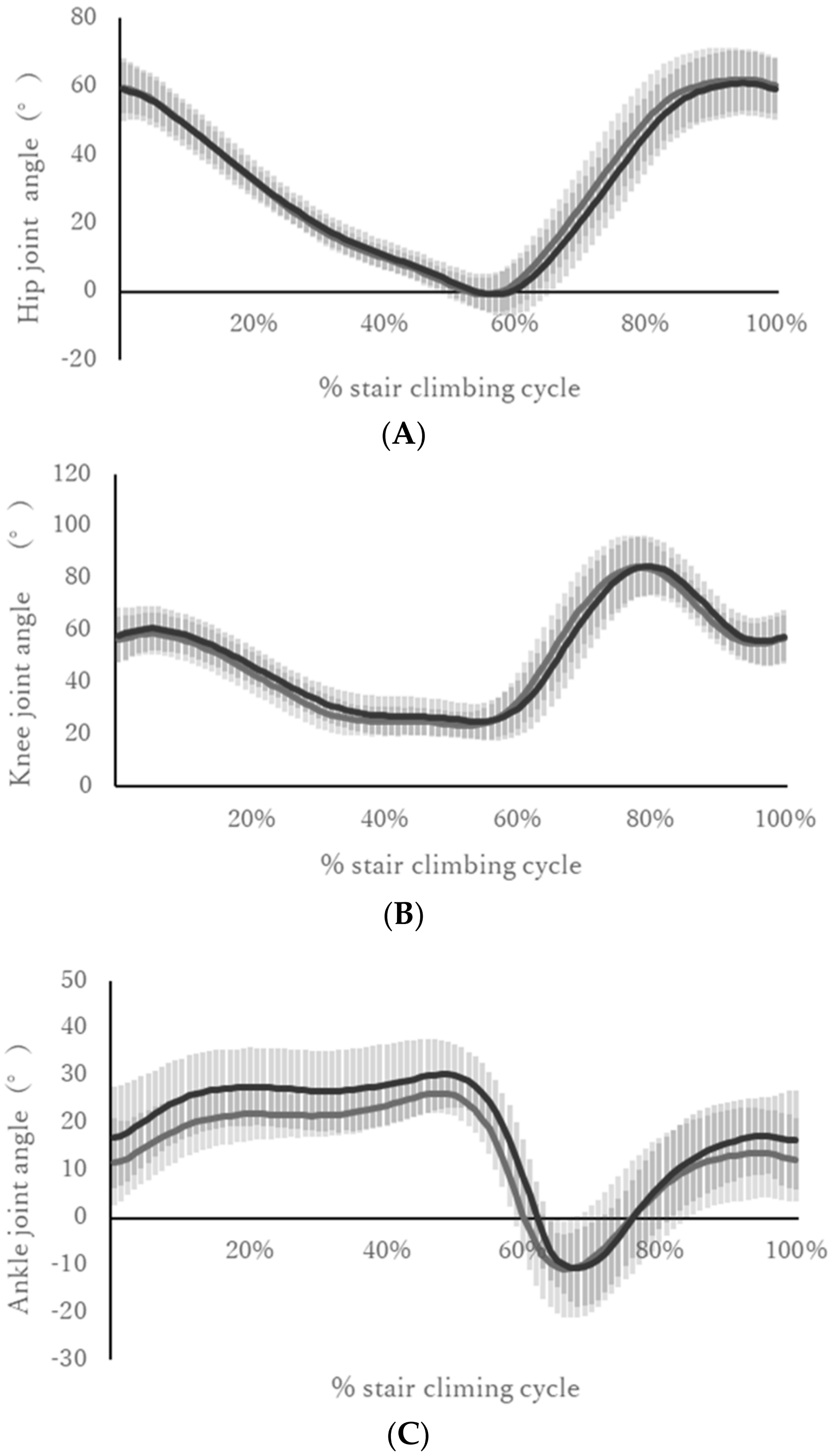

Figure 4. Mean sagittal plane joint angle of the hip (**A**, +: Flexion, −: Extension), knee (**B**, +: Flexion, −: Extension), and ankle (**C**, +: Dorsiflexion, −: Plantarflexion) joint during stair climbing. The gray and black lines represent the mean during the mean beginning phase of stair climbing and the end phase of stair climbing, respectively. The gray and black shaded vertical bars represent the standard deviation during the mean beginning phase of stair climbing and the end phase of stair climbing, respectively.

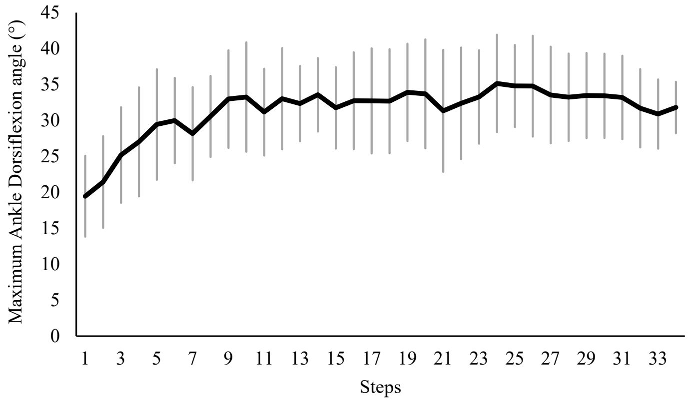

Figure 5. Maximum ankle dorsiflexion angle trend.

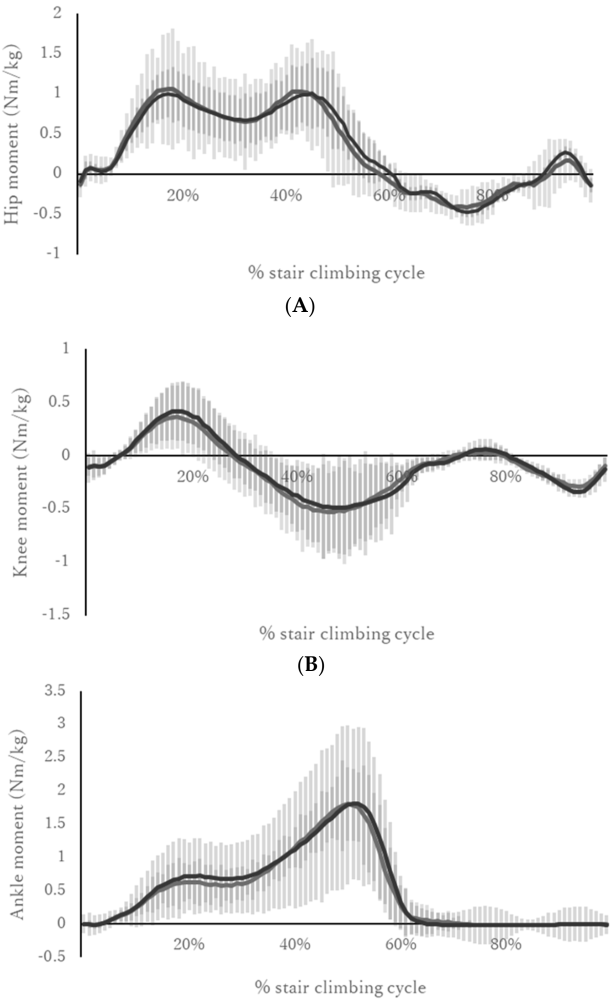

Figure 6. Mean sagittal plane joint moment of the hip (**A**, +: Flexion, −: Extension), knee (**B**, +:Flexion, −: Extension), and ankle (**C**, +: Plantarflexion, −: Dorsal-flexion) joint during stair climbing. The gray and black lines represent the mean during the mean beginning phase of stair climbing and the end phase of stair climbing, respectively. The gray and black shades vertical bar represent the standard deviation during the mean beginning phase of stair climbing and the end phase of stair climbing, respectively.

Table 4. The sagittal plane kinetics of hip, knee, and ankle during stair climbing.

ID	Hip Joint Moment (Nm/kg)			
	Maximum Flexion		Maximum Extension	
	Beginning Phase	End Phase	Beginning Phase	End Phase
1	−1.40	−1.29	1.47	1.29
2	−0.63	−0.70	1.05	1.04
4	−0.67	−0.61	1.52	1.49
5	−0.52	−0.52	1.39	1.48
6	−0.64	−0.56	1.13	1.08
7	−0.55	−0.51	1.02	1.07
8	−0.42	−0.40	1.33	1.23
9	−0.62	−0.56	1.24	1.43
Mean	−0.68	−0.64	1.27	1.26
SD	0.30	0.27	0.19	0.19
95% CI	−0.10, 0.093		−0.01, 0.085	
p-value	0.15		0.93	
ID	**Knee Joint Moment (Nm/kg)**			
	Maximum Extension		**Maximum Flexion**	
	Beginning Phase	**End Phase**	**Beginning Phase**	**End Phase**
1	0.57	0.48	−0.84	−0.80
2	0.51	0.52	−0.58	−0.58
4	0.14	0.15	−0.90	−0.78
5	0.18	0.26	−0.77	−0.79
6	0.74	0.82	−0.66	−0.61
7	0.75	0.77	−0.57	−0.55
8	0.12	0.21	−0.48	−0.39
9	0.71	0.51	−0.36	−0.52
Mean	0.46	0.47	−0.65	−0.63
SD	0.28	0.25	0.18	0.15
95% CI	−0.08, 0.082		−0.05, 0.09	
p-value	0.98		0.55	
ID	**Ankle Joint Moment (Nm/kg)**			
	Maximum Dorsiflexion		**Maximum Plantarflexion**	
	Beginning Phase	**End Phase**	**Beginning Phase**	**End Phase**
1	−0.06	−0.05	1.88	1.91
2	−0.13	−0.16	2.02	1.92
4	−0.08	−0.11	1.90	1.93
5	−0.05	−0.06	2.00	1.87
6	−0.07	−0.07	1.97	1.86
7	−0.24	−0.23	1.90	1.82
8	−0.18	−0.23	1.65	1.69
9	−0.10	−0.17	2.11	2.01
Mean	−0.11	−0.13	1.93	1.87
SD	0.07	0.07	0.14	0.09
95% CI	−0.12, 0.01		−0.04, 0.001	
p-value	0.06		0.07	

SD: standard deviation, CI: Confidence interval.

4. Discussion

The present study examined the differences in kinematic and kinetic characteristics of the lower limb between the beginning and end phases of long stair climbing in an outdoor environment for healthy young individuals using a wearable motion analysis system. We found that the dorsiflexion angle of the ankle joint increased in the end phase of the long stair climbing compared to the beginning phase. This finding supports our hypothesis.

In contrast, the kinetic characteristics of the lower limb showed no significant differences between phases. This study is the first to characterize the biomechanical properties of a long outdoor stair in a real-life environment using a wearable motion analysis system.

A comparison of kinematic changes during long stair climbing between the beginning and end phases showed no significant differences in the hip and knee joints. On the other hand, the ankle joint showed a significant increase in the maximum dorsiflexion angle in the end phase of stair climbing compared to the beginning phase. The maximum ankle dorsiflexion angle in this study was found immediately before the change in the direction of ankle joint dorsi-plantarflexion (see Figure 4). Furthermore, the maximum ankle dorsiflexion angle was simultaneous with the maximum ankle plantarflexion moment (see Figure 5). The maximum dorsiflexion angle of the ankle joint in the present study was found between the pull-up phase and the forward movement phase of the stair climbing cycle. The maximum ankle dorsiflexion angle in the end phase of climbing may increase due to the compensatory forward shift of the center of mass caused by the increased anterior tilt of the lower leg. In other words, the forward shift of the center of mass in the end phase of climbing may be caused by a passive increase in the dorsiflexion angle of the ankle joint rather than an increase in the plantarflexion moment of the ankle joint.

There were no significant kinetic differences in the hip, knee, and ankle joints moment between the beginning and end phases of long stair climbing. Furthermore, there were no significant differences in stair climbing speed between the beginning and end phases. These results showed that 66 stair ascents were achievable in healthy young adults without any kinematic changes. The stairs used in the previous study ranged from 13 to 24 cm for the kick-up and 27 to 30 cm for the tread [6,17,19,29–31]. In addition, the standard for outdoor stairs in the Building Standard Law requires a kick-up of 23 cm or less and a tread of 26 cm or more. Compared to these stair dimensions, those in this study were similar in kick-up but with wider treads. Previous studies have reported an increase in moments at each joint in the lower limb as the kick-up height increases [26], but the tread size has not been examined. Compared to the stair dimensions used in the previous study in healthy adults [26], the wider stair tread in the present study resulted in greater joint moments at the hip and ankle joints. A wider tread may result in a gentler stair slope and an increase in stride length; thus, the larger hip extension moment and ankle plantarflexion moment may be required for generating the forward propulsive force.

There are three limitations of this study. Firstly, the study included only healthy young adults (four males and four females); therefore, results may differ for the elderly, other genders, those with osteoarthritis, and those with other diseases. A previous study comparing the analysis of stairs in elderly and healthy young subjects reported significant differences in kinematic parameters [32]. Furthermore, previous biomechanical studies of stair ascent and descent concerning gender differences found significantly higher hip and knee joint angles in women than in men during ascent and significantly higher hip and ankle joint angles in women than in men during descent [33]. Secondly, we analyzed only the lower limb sagittal plane; analysis in the frontal and transverse planes was not performed. It should also be acknowledged that we did not measure trunk kinematics because the sensors were only applied to the lower limb. The previous study reported significant differences in the kinematic parameters of the trunk between a symptomatic group with femoroacetabular impingement and an asymptomatic group during stair ascent and descent [31]. In the future, we will attempt to verify the accuracy of the system's analysis of frontal and transverse planes. Furthermore, we will investigate the effects of each disease on stair climbing in a real-life environment using a wearable motion analysis system. The third limitation of the study concerns the measurement equipment. This study's lower extremity load measuring device is a rigid material that does not deform the outsole due to the built-in foot pressure gauge. A previous study [34] showed significant effects of shoe insole hardness on balance control during stair ascent and descent. The biomechanical properties may differ depending on the shoe material used during the measurement. In the future, we should consider using sheet-type foot pressure sensors [22], which may have less influence on the movement. Finally, the fourth limitation of this study concerns the measurement environment. In this study, we conducted measurements on stone steps in the winter. A previous study reported that motor control differs when walking on slippery floors [35]. Therefore, biomechanical properties may differ depending on the stair's temperature, shape, and material. Therefore, we will investigate the influence of the measurement environment on biomechanical properties.

5. Conclusions

We demonstrated the kinematic characteristics of the lower limb during long stair climbing in an outdoor environment using a wearable motion analysis system. In the end phase of stair climbing, the maximum ankle dorsiflexion angle during the stair climbing cycle was significantly larger than in the beginning phase. This finding indicated that fatigue during long stair climbing might increase ankle dorsiflexion to compensate for forwarding propulsion in the end phase.

Author Contributions: H.Y.: conception and design of the study, acquisition of data, analysis, and interpretation of data, drafting the article. Y.S. and K.H.: conception and design of the study, acquisition of data, analysis and interpretation of data, and revision of the article. K.F., C.H. and K.N.: analysis and interpretation of data and modification of the article. C.Z.: analysis and interpretation of data. S.-I.I.: conception and design of the study and modification of the article. All authors have read and agreed to the published version of the manuscript.

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Institutional Review Board Statement: All participants provided informed consent and the Institutional Review Board approved this study of Tohoku University (approval ID: 2018-1-553). It was conducted according to the principles of the Declaration of Helsinki.

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study. Written informed consent has been obtained from the patient(s) to publish this paper, if applicable.

Data Availability Statement: Data are available upon request due to privacy and ethical restrictions.

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Conflicts of Interest: K. Fukushi, C. Huang, and K. Nakahara are employees of the NEC corporation.

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