Acta of Bioengineering and Biomechanics

Biomechanical Characterization of Slope Walking Using Musculoskeletal Model Simulation

Abstract:

Purpose
Upslope and downslope walking are basic activities necessary for normal daily living in the community, and they impose greater joint load on the lower extremities than during level walking. Thus, the purpose of this study was to quantify the resultant and shear forces in the hip and knee joints during slope walking.

Methods
Twelve healthy volunteers were evaluated walking under level and 10° up- and downslope conditions; three-dimensional gait analysis was conducted using a 7-camera optoelectronic motion analysis system combined with a force plate to measure ground reactive force. Joint forces in the hip and knee joints were estimated using musculoskeletal model simulation.

Results
Results showed that the resultant hip force was increased significantly to 117.2% and 126.9%, and the resultant knee force was increased to 133.5% and 144.5% in up- and downslope walking, respectively, compared with that of level walking. Furthermore, increased shear force in the hip and knee joints was noted during both slope walking conditions.

Conclusions
This information may be beneficial for therapists advising elderly people or patients with osteoarthrosis on an appropriate gait pattern, gait assistive devices, or orthoses according to their living environment.

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Keywords:

gait analysis, slope walking, osteoarthrosis, joint force, musculoskeletal model simulation

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Celina Pezowicz, Ph.D
Editor-in-Chief
ACTA OF BIOENGINEERING AND BIOMECHANICS

Dear Editor:

Please find our manuscript titled “Biomechanical Characterization of Slope Walking Using Musculoskeletal Model Simulation” by Masayuki Kawada enclosed, which we would like to submit for publication as a research paper in the *ACTA OF BIOENGINEERING AND BIOMECHANICS.*

The purpose of this study was to quantify the magnitude and direction of joint force during slope walking. Excessive joint load is a known risk factor for musculoskeletal disorders, such as osteoarthrosis in both the hip and knee joints. Joint force is one of the parameters that quantifies joint loads and is often calculated using musculoskeletal model simulations. Excessive joint force under various weight-bearing conditions may be related to joint deformity in the lower limbs. These deformities are caused by longitudinal and shear forces that decrease joint stability. Joint force during walking has been reported by some researchers, but those studies did not include joint force direction. In order to accurately prescribe appropriate physical activity for patients with osteoarthrosis, we must understand the direction of the joint force, not merely its magnitude.

No significant differences were observed in gait velocity or stride length among the three walking conditions tested. However, we found several differences in hip and knee joint moment and force between level and slope walking. These results indicate increased joint load in the hip and knee joints at stance phase during slope walking and kinematic differences between upslope walking and downslope walking.

The results of this study are significant because they demonstrate that hip and knee joint forces are greater during slope walking than level walking. Because these forces might contribute to development of musculoskeletal disorders, therapists are able to better provide their patients with techniques that can decrease joint force while walking.

We believe that these findings will be of great interest to your readers in general and to fellow human kinetics researchers in particular. As the premier journal devoted to this field, the *ACTA OF BIOENGINEERING AND BIOMECHANICS* represents an ideal platform for us to share these results with the international research community.

We confirm that this manuscript has not been published elsewhere and is not under consideration by another journal. All authors have approved the manuscript and agree with submission to the *ACTA OF BIOENGINEERING AND BIOMECHANICS*. The authors have no conflicts of interest to declare.

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We look forward to hearing from you at your earliest convenience.

Sincerely,

Acta of Bioengineering and Biomechanics

Introduction

During daily living, lower extremity joints are subjected to repeated loads when performing antigravitational activities, such as walking and standing. Excessive joint load is a known risk factor for developing musculoskeletal disorders, such as osteoarthrosis in the hip and knee joints [1]-[2]. Osteoarthrosis, which is common in elderly people, can cause joint pain and dysfunction, resulting in functional limitations. Thus, elderly patients and those with osteoarthrosis would benefit from avoiding excessive lower extremity joint load during their daily activities.

Upslope and downslope walking are basic daily activities and are characterized by increased propulsive and braking forces in the ground reaction force (GRF) [3]-[4]. The difference of GRF produces changes in internal joint moment. During 8° upslope walking, hip extension and ankle plantarflexion moments at late stance increase to 150% and 118%, respectively, compared with level walking [3]. During 8° downslope walking, the knee extension moment at early stance was twice that of level walking [3]. These alterations in joint moments result in an increased activation of related muscles [5]-[7]. GRF acts on joint surfaces via the bones, and muscle contractions pull the bones toward each other, resulting in changes in joint load [8]-[9]. In particular, tensile force generated by muscle activation contributes a large part of the joint force [8]-[10]. Thus, the alteration of GRF and muscle activation during slope walking could increase joint load. However, few studies have analyzed joint load during slope walking.

Joint force is one of the parameters that quantifies joint load. Joint force can be calculated using musculoskeletal model simulations [8], [10]-[13] or measured *in vivo* with instrumented implants [14]-[15]. The noninvasive musculoskeletal model is generally utilized in movement science and can safely measure joint force during various activities. Excessive joint force under weight-bearing conditions might be related to joint deformity in the lower limbs, including femoral head migration in hip osteoarthrosis or varus deformity in knee osteoarthrosis. These deformities are caused by longitudinal and shear forces that decrease joint stability. Joint force during walking has been reported by previous studies, but these did not include analysis of joint force direction [8], [13], [15]. The resultant and shear forces during gait are associated with joint pain and deformity; therefore, knowledge regarding these issues would be beneficial for therapists engaged in the rehabilitation of elderly people and those with osteoarthrosis.

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The purpose of this study was to quantify the resultant and shear forces in the hip and knee joints during slope walking using musculoskeletal model simulation. We hypothesized that the resultant and shear forces in the hip and knee joints are increased during slope walking, especially downslope, compared with the forces that occur during level walking. To further interpret the difference in joint force due to walking conditions, changes in GRF and internal joint moments were analyzed.

Materials and Methods

Subjects

Twelve healthy young adults (age: 26.1 ± 5.7 years; height: 170.7 ± 5.5 cm; weight: 64.9 ± 8.2 kg; average ± standard deviation) without any orthopedic or neurological disorders participated in this study. Each participant read and signed an informed consent form approved by the Ethics Committee of Kagoshima University Medical School (No. 155).

Data collection

We calculated joint force using the musculoskeletal model simulation software AnyBody 6.0 (AnyBody Technology, Aalborg, DK) from motion capture and GRF data. The validity of muscle force and joint forces estimated by this musculoskeletal model simulation software has been confirmed during previous study [16]. Subjects were evaluated under three gait conditions: level walking, upslope walking, and downslope walking. The walkway consisted of a 3-m plane inclined to 10° with 3-m horizontal areas at both ends. To minimize the effect of gait velocity, subjects walked at 100 steps/min using a metronome. Five trials were measured for each gait condition [4], [9]. Prior to data collection, the subjects practiced each gait several times.

Motion capture was conducted using a 7-camera optoelectronic motion analysis system (VICON MX3, Oxford Metrics, Oxford, UK) combined with a force plate (9286A, Kistler, Jonsered, SE). The force plate was secured in the middle of the inclined plane to obtain the GRF. Sampling frequencies of the infrared camera and the force plate were 100 Hz and 1000 Hz, respectively.

Each subject wore 25 retro-reflective markers on bony landmarks of the head, thorax, pelvis, and right lower extremities, based on a plug-in-gait marker set.

Data analysis

Marker trajectories and GRF data were filtered using a Butterworth low-pass filter at 5 Hz and 12 Hz cut-off frequencies, respectively. GRF was analyzed according to the force plate reference frame [6]. In this study, we focused on hip and knee joint forces. Internal joint moments and joint forces were calculated using the MocapModel in AMMR 1.6.4, which is the standard model available in AnyBody. The musculoskeletal model includes 170 muscles, 6 segments (head, trunk, pelvis, right thigh, right shank, and right foot), and has 10 degrees of freedom. The model was scaled to subjects according to their segment length and body mass. We used a simple muscle configuration without force-length-velocity relationships in that musculoskeletal mode, according to a previous study which reported that force-length-velocity relationships have little effect on prediction of muscle forces and joint forces while walking [17]. Marker trajectories and GRF data were put into the musculoskeletal model to calculate the hip and knee joint forces. Joint moments and joint forces were estimated by inverse dynamic analysis and optimization. In the optimization process, muscle forces were calculated to minimize the sum of the cubes of muscle stress, described by the ratio of muscle force to maximum muscle force in each muscle [18]. Hip and knee joint forces were calculated from the net joint force and tensile force of the muscles crossing those joints, and resolved into three components based on the reference frame of the child segment. We also obtained resultant force in the hip and knee joints. The glenoid fossa of the knee joint lies perpendicular to the longitudinal direction of the tibia, thus anteroposterior and mediolateral joint forces relative to the shank create a shear force. Meanwhile, the acetabulum is not perpendicular to the longitudinal direction of the thigh due to the neck shaft angle in the frontal plane. It is therefore difficult to estimate the shear force of the hip joint in the frontal plane from components of joint forces relative to the thigh coordinate system. Thus, we calculated the direction of hip joint force vectors for the femoral long axis on the frontal plane at resultant force peak, to define the shear force on the hip joint (Figure 1) [19]. GRF, joint moments, and joint force data were normalized to each subject’s body weight, and time was normalized to percentage of gait cycle. Gait velocity and stride length were calculated from the trajectory of the heel marker.

We analyzed gait velocity, stride length, GRF, joint moments, joint forces, and direction of hip joint force for all walking conditions. We analyzed peak value during the early stance as the first peak and late stance phases as the second peak for kinematic and kinetic data from 5 trials for each walking condition. Results are shown as the mean ± standard deviation and their percentage to the value of level walking. Normality of distribution was tested using the Shapiro-Wilk test. If normality of distribution could be assumed, data were analyzed by one-way repeated measures ANOVA with Schaffer's post hoc test to define the effect of gait condition on joint load. If the normality of distribution could not be assumed, data were analyzed by the Friedman test with Wilcoxon signed rank test adjusted using the Holm method as a post hoc test. All statistical tests were performed using R 2.8.1. For all analyses, the level of significance was set at alpha < 0.05.

Results

No significant differences were observed in gait velocity and stride length between the three walking conditions (Table 1). Vertical GRF was significantly increased at late stance of upslope walking ($P = 0.002$) and at early stance of downslope walking ($P < 0.001$), compared to level walking (Table 2). Braking and medial GRFs during down slope walking were larger than other two conditions ($P < 0.001$). Meanwhile, propulsion GRF during upslope walking was larger than other two conditions ($P < 0.001$). Increased hip extension moments at early stance were observed during upslope walking compared to the other walking conditions (180.8%; $P < 0.001$; Figure 2A; Table 2). Meanwhile, hip abduction moment during downslope walking was increased at early stance and late stance, and was larger than other two conditions (Figure 2B, Table 2; $P < 0.001$). Level and upslope walking showed similar knee extension moment patterns during the stance phase (Figure 2C). Conversely, during downslope walking, knee muscles generated an extension moment throughout the stance phase. The knee extension moment increased during upslope and downslope walking compared to level walking, the latter difference being particularly significant ($P < 0.002$). An increased knee flexion moment at late stance was increase during upslope walking compared to level walking ($P = 0.002$).

Hip joint forces, except for the anterior-posterior force, showed two peaks during the stance phase and these were directed to the medial superior during the three walking conditions in the frontal plane (Figure 3). In comparison to level walking, resultant hip joint forces at early stance increased to 117.2% during upslope walking, and to 126.9% during downslope walking, indicating significant differences among the three walking conditions (Figure 3A; $P < 0.031$). Vertical hip joint force showed a similar tendency of resultant force

(Figure 3B). Anterior-posterior hip joint force showed a similar pattern during level and upslope walking (Figure 3C). Conversely, during downslope walking, posterior hip joint force was observed throughout the stance phase. Posterior hip joint force during early stance was significantly greater during upslope (200.0%) and downslope walking (199.7%) than for level walking (Figure 3B). Increased anterior hip joint force (120.0%) was observed at late stance during upslope walking compared with level walking. Medial hip joint force at early stance was increased significantly to 122.1% ($P = 0.013$) and 114.2% ($P = 0.029$) during up- and downslope walking, respectively, compared with level walking. The hip joint force vector was acute to the femoral long axis at early ($P = 0.005$) and late stances ($P = 0.004$) during downslope walking compared to level and upslope walking (Table 4).

In a similar manner to hip joint force, and with the exception of anterior-posterior force, knee joint force showed two peaks during the stance phase and was directed to the medial superior during all walking conditions in the frontal plane (Figure 4). In comparison to level walking, resultant knee joint forces increased to 133.5% at early stance during upslope walking, and to 144.5% at early stance and to 124.9% at late stance during downslope walking, indicating significant differences among the three walking conditions (Figure 3A; $P < 0.025$). Vertical knee joint forces had a similar tendency of resultant force (Figure 3B). The knee joint vector was directed backward in early stance and forward in late stance during the three walking conditions (Figure 4C). Posterior knee joint force during early stance was significantly greater in upslope (441.7%) and downslope (233.3%) walking, especially the former ($P < 0.002$). Medial knee joint force at early stance was increased significantly to 130.0% ($P < 0.001$) during upslope gait and to 150.0% ($P < 0.001$) during downslope walking compared with level walking.

Discussion

We investigated the load of hip and knee joints during sloped walking through joint moment and joint force as calculated using a musculoskeletal model simulation. We found several significant differences in the hip and knee joint moments: i.e. that resultant forces of the hip and knee joints were increased during slope walking compared with those of level walking, especially during downslope walking. In addition, the present study showed the largest shear force during downslope walking, this included: upward directed vector of resultant force in the hip joint and increased medial-lateral force in the knee joint. These results are consistent with our hypothesis, with the exception of anterior force in the hip joint and posterior force in the knee joint.

Peak resultant hip and knee joint forces during level walking were measured at 3.42 BW and 4.41 BW, respectively, and were similar to previous studies based on musculoskeletal simulation [8], [11]-[12]. Meanwhile, increases of resultant joint force, GRF and joint moment results during slope walking agree with a previous studies [3]-[4], [20]. The increases in joint force have a close relationship with the alteration of GRF and joint moment, and previous study report that approximately 80% of joint force is derived from muscle force in the hip and knee joint force [9].

During upslope walking, the resultant force and posterior shear force at early stance and anterior shear force at late stance in the hip joint were greater than those during level walking. In early stance, the hip extension moment was increased in upslope walking. Previous studies report that magnitude and duration of activity of the gluteus maximus and hamstring are increased at the stance phase during upslope walking compared with level walking [5]-[7]. Increased hip extensor muscles pull the thigh upward and backward, resulting in increased upward and backward shear forces [8]. Meanwhile, increased propulsion force contributed to the increased anterior shear force in the hip joint at late stance.

Similar to the hip joint, the resultant and shear forces in the posterior and medial direction at early stance and posterior share force in the knee joint were increased in upslope walking when compared to level walking. Knee extension moment at early stance was increased in upslope walking compared with level walking, and required the activation of the quadriceps femoris. These results are consistent with previous studies which analyze slope walking using electromyography [5]-[7]. Resultant knee joint force during upslope walking increased due to greater activation of the quadriceps femoris and hamstring muscles [10]. Posterior knee joint force may be increased by greater activation of the hamstrings following the hip extension moment. Because the hamstrings run posterior to the hip and knee joints, they generate a hip extension moment and a posterior force on the tibia. Meanwhile, increased propulsion force contributed to increased anterior shear force in the knee joint at late stance.

In downslope walking, resultant and backward shear in the hip joint at early stance during downslope walking were greater than during level walking. In addition, the vector of resultant force of the hip was directed upward relative to the longitudinal axis of the femur, compared to the other gait conditions. At early stance, hip abduction moment was increased compared to the other two gait conditions due to an increase in the vertical and medial GRF. Thus, an increase in gluteus medius activity following the hip abduction moment and increased breaking GRF contributed to a hip joint force greater than that of level walking. Meanwhile, increased vertical GRF and upward and lateral force generated by the gluteus medius altered the direction of the vector of the resultant force during downslope walking.

The resultant and medial shear forces in knee joint force during downslope walking were greater than those measured during level and upslope walking, especially at the late stance. During downslope walking, the knee extension moment was generated through the stance phase, and those were the largest of the three gait conditions. Previous studies also report that activation of the rectus femoris increases during downslope walking compared with level and upslope walking at similar levels of inclination [5]-[6]. The increase in vertical knee joint force was dependent on the activity of the quadriceps femoris following the knee extension moment. Increased posterior and medial forces in the knee joint were caused by increased posterior and medial GRF.

Joint forces act directly on joints and are understood to be related to mechanical stress and progression of bone and joint disease. A previous study reported that mechanical stress on the joint is the main contributor to osteoarthrosis progression [1]. Abnormal and excessive hip joint forces cause anterior hip joint pain and instability and can lead to pathology of the acetabular labrum [2]. Another study reported femoral head migration that showed either an anterior and superior pattern or a posterior and medial pattern [21]. Increased vertical, anterior, and posterior hip joint forces during slope walking are considered to be related to this phenomenon. In addition, the resultant hip joint force acute to the femoral longitudinal axis during downslope walking would relative to the high risk of femoral head migration. In a previous study, extreme medial shear forces have been observed in patients with medial compartment knee osteoarthrosis during the stance phase, resulting in progression of osteoarthritis [22]. Previous studies reported decreased walking velocity, decreased step length, and use of a cane can help reduce joint forces [11]-[12]. Thus, therapists should advise elderly people with joint pain or osteoarthrosis on an appropriate gait pattern, a gait assistive device, or orthoses, according to their living environment [23]-[24].

This study has several limitations. Although the results of the present study agree with previous studies reporting joint reaction force during walking using a musculoskeletal model [8], [11]-[12], our results showed greater joint forces than those estimated by instrumented prostheses [14]-[15]. As a result, use of a musculoskeletal model might overestimate joint reaction forces. These results should be interpreted while considering differences in calculation methods. In addition, we did not analyze the distribution of knee joint forces to the medial and lateral knee joint components. A previous study report that the walking load on the medial knee joint is greater than on the lateral knee joint [25]. Further study is needed to fully describe the relationship between joint forces and osteoarthrosis.

The present study measured the biomechanical characteristics of upslope and downslope walking using magnitude, and direction of joint force. In slope walking, the resultant and shear hip and knee joint forces were greater than when measured during level walking, especially during downslope walking. These forces may lead to musculoskeletal disorders. Therefore, therapists should advise patients on methods that decrease joint force during slope walking to limit development of these disorders.

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Tables

Table 1. Gait parameters

	Level	Upslope	Downslope	F	P
Velocity (m/s)	1.08 ± .08	1.06 ± .10	1.10 ± .16	.98	0.392
Stride length (m)	1.30 ± .10	1.31 ± .11	1.31 ± .18	.017	0.983

Values are presented as the mean ± standard deviation.

Table 2. GRF and internal joint moment

		Level	Upslope	Downslope	F	P
GRF (N/kg)						
Vertical	1st	10.55 ± .81	10.51 ± .71††	12.30 ± .92**	61.17	< 0.001
	2nd	10.42 ± .44	10.80 ± .57*††	9.37 ± .66**	29.87	< 0.001
Braking force		-1.86 ± .34	-.31 ± .18**††	-4.05 ± .57**	364.94	< 0.001
Propulsion force		1.98 ± .21	3.52 ± .34**††	1.01 ± .34**	355.43	< 0.001
Medial	1st	.55 ± .15	.51 ± .18††	.77 ± .19**	62.81	< 0.001
	2nd	.54 ± .16	.56 ± .15	.61 ± .14*	4.41	0.024
Internal joint moment (Nm/kg)						
Hip extension		.52 ±.08	.94 ± .12**††	.60 ± .14	46.74	< 0.001
Hip flexion		-.55 ± .15	-.41 ± .11**	-.56 ± .20	11.19	< 0.001
Hip abduction	1st	.62 ± .11	.48 ± .12**††	.82 ± .17**	20.167	< 0.001
	2nd	.60 ± .12	.50 ± .14**††	.68 ± .11*	23.09	< 0.001
Knee extension		.43 ± .27	.68 ± .22**††	.93 ± .28**	45.989	< 0.001
Knee flexion		-.19 ± .19	-.31 ± .19**††	.61 ± .19**	185.43	< 0.001

Values are presented as the mean ± standard deviation.

Only result of hip abduction moment in 1st peak was analyzed by the Friedman test.

*P < 0.05, significant difference vs. level walking

**P < 0.01, significant difference vs. level walking

†P < 0.05, significant difference vs. downslope walking

††P < 0.01, significant difference vs. downslope walking

Table 3. Joint forces

	Level	Upslope	Downslope	F	P
Hip (*BW)					
Resultant 1st	3.09 ± .32	3.62 ± .47**†	3.92 ± .46**	36.74	< 0.001
2nd	3.42 ±.64	2.88 ± .75**†	3.42 ± .71	8.62	0.002
Vertical 1st	2.86 ± .29	3.29 ± .40**††	3.65 ± .39**	46.63	< 0.001
2nd	3.31 ± .60	2.76 ± .71**††	3.34 ± .69	11.08	< 0.001
Posterior	-.30 ± .15	-.60 ± .15**	-.59 ± .17**	41.12	< 0.001
Anterior	.15 ± .08	.18 ± .07**††	-.03 ± .05**	70.11	< 0.001
Medial 1st	1.13 ± .17	1.38 ± .25*	1.29 ± .26*	6.87	< 0.001
2nd	.83 ± .26	.74 ± .32	.69 ± .21	2.05	0.169
Knee (*BW)					
Resultant 1st	2.63 ± .47	3.51 ± .64**†	3.80 ± .68**	53.81	< 0.001
2nd	4.41 ± .54	4.26 ± .88††	5.51 ± .80**	12.39	0.004
Vertical 1st	2.58 ± .46	3.42 ± .61**†	3.71 ± .67**	53.02	< 0.001
2nd	4.33 ± .52	4.18 ± .85††	5.40 ± .79**	12.50	0.004
Posterior	-.12 ± .04	-.53 ± .17**††	-.28 ± .14**	53.62	< 0.001
Anterior	.31 ± .17	.35 ± .26††	.14 ± .16**	11.25	< 0.001
Medial 1st	.50 ± .11	.65 ± .11**††	.75 ± .12**	63.51	< 0.001
2nd	.79 ± .11	.72 ± .15**††	1.05 ± .14**	27.69	< 0.001

Values are presented as the mean ± standard deviation.
*P < 0.05, significant difference vs. level walking
**P < 0.01, significant difference vs. level walking
†P < 0.05, significant difference vs. downslope walking.
††P < 0.01, significant difference vs. downslope walking

Table 4. Hip joint force angle (degree)

	Level	Upslope	Downslope	F	P
1st (degree)	21.4 ± 2.2	22.5 ± 1.9††	19.3 ± 2.3**	12.38	< 0.001
2nd (degree)	14.6 ± 2.2	15.5 ± 2.9*††	11.2 ± 1.7**	13.54	< 0.001

Values are presented as mean ± standard deviation.

*P < 0.05, significant difference vs. level walking

**P < 0.01, significant difference vs. level walking

†P < 0.05, significant difference vs. downslope walking

††P < 0.01, significant difference vs. downslope walking

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Fig 1

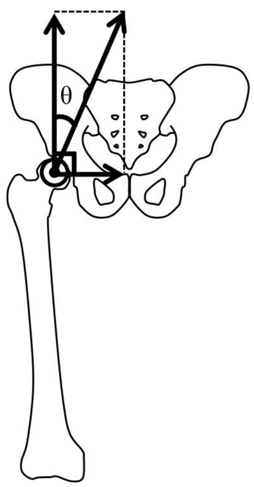

Figure 1 Hip joint force angle (θ) was defined as the angle between the vector of the resultant force on the frontal plane and the vertical axis of the thigh.

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Fig 2

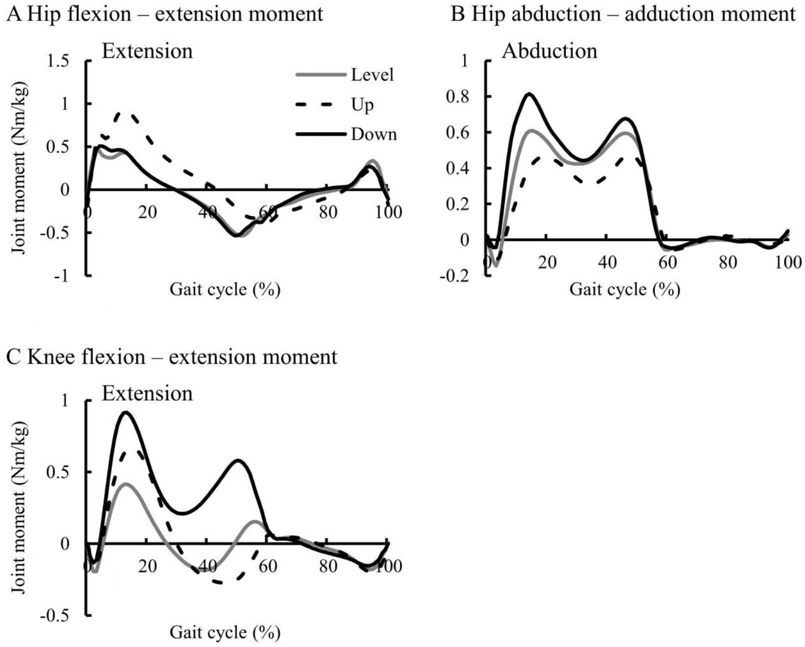

Figure 2 Ensemble average of internal hip and knee joint moment across all subjects. The largest hip extension moment was observed during upslope walking. The largest hip abduction and knee extension moment were observed during downslope walking.

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Fig 3

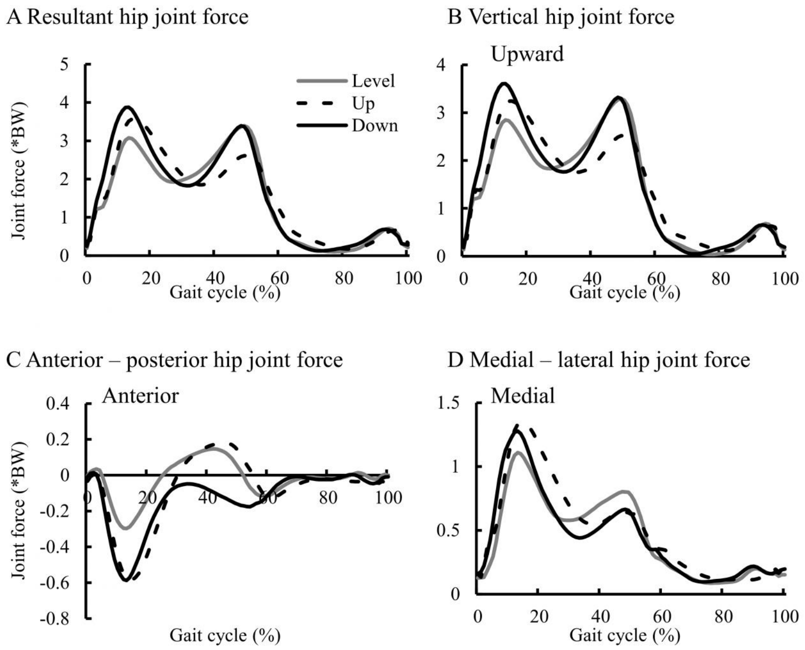

Figure 3 Ensemble average of hip joint force across all subjects. Hip joint forces of all direction during slope walking were larger than level walking at early stance.

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Fig 4

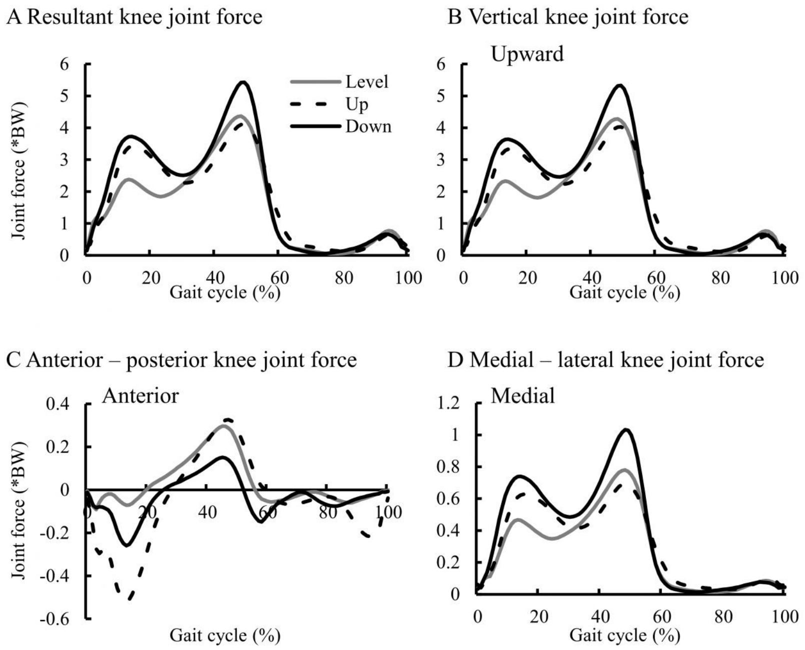

Figure 4 Ensemble average of knee joint force across all subjects. During downslope walking, resultant, vertical and medial knee joint forces were the largest among all conditions throughout stance phase. The largest posterior knee joint force at early stance was observed during upslope walking.

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Figures

Figure 1 - Download source file (225.69 kB)

Figure 1 Hip joint force angle (θ) was defined as the angle between the vector of the resultant force on the frontal plane and the vertical axis of the thigh.

Figure 2 - Download source file (490.24 kB)

Figure 2 Ensemble average of internal hip and knee joint moment across all subjects. The largest hip extension moment was observed during upslope walking. The largest hip abduction and knee extension moment were observed during downslope walking.

Figure 3 - Download source file (573.57 kB)

Figure 3 Ensemble average of hip joint force across all subjects. Hip joint forces of all direction during slope walking were larger than level walking at early stance.

Figure 4 - Download source file (573.8 kB)

Figure 4 Ensemble average of knee joint force across all subjects. During downslope walking, resultant, vertical and medial knee joint forces were the largest among all conditions throughout stance phase. The largest posterior knee joint force at early stance was observed during upslope walking.