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Original Research

Characteristics of temporal changes in mechanical energy flow among lower-limb segments during walking with knee extension restriction

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ABSTRACT

Objective: The knee joint plays a critical role during gait. When knee extension is restricted, adjacent joints are known to compensate through changes in spatiotemporal and kinematic parameters. However, the specific compensatory strategies and patterns of mechanical energy transfer remain insufficiently understood. This study aimed to clarify the characteristics of temporal changes in mechanical energy among lower-limb segments during gait with restricted knee extension.

Methods: Sixteen healthy young male participants were enrolled in this study. Each participant performed level walking at a controlled speed under two conditions: walking without a brace (free walking) and walking with a knee device imposing a 20° extension restriction (restricted walking). Three-dimensional motion analysis was used to calculate joint moments and segment moment power for each gait phase (loading response, mid-stance, terminal stance, and pre-swing). Paired statistical analyses were conducted to compare outcomes between the two conditions.

Results: During the loading response phase, proximal positive power of the shank significantly decreased, whereas negative power significantly increased under the restricted walking condition. Increased proximal negative power of the thigh was observed during mid-stance. During terminal stance, proximal thigh and distal shank negative power decreased, whereas distal thigh negative power increased. No significant differences were observed during the pre-swing phase.

Conclusions: Compared with free walking, restricted knee extension was associated not only with alterations in joint moments but also with compensatory segment moment power strategies between lower-limb segments.

Keywords: limited knee extension range of motion when walking, lower-limb mechanical energy, energy transfer, compensatory strategies

INTRODUCTION

During human walking, the knee joint exhibits a double knee action, repeatedly flexing and extending within a single gait cycle. This double knee action plays a crucial role in shock absorption at initial contact and in reducing vertical displacement of the body's center of gravity.[1-4] When knee joint extension is restricted, this mechanism does not function effectively, resulting in impaired walking performance.[5,6] To date, numerous studies have examined the relationship between restricted knee joint extension and spatiotemporal factors, such as walking speed and stride length, as well as kinematic variables, including knee joint moment and power.[7-12] However, because walking requires coordinated control of the body's center of gravity through the combined actions of the knee, ankle, and hip joints, it is essential to understand the effects of knee extension restriction on adjacent joints. Whitehead et al.[13] reported that restricting knee joint extension reduces hip joint flexion range during the early stance phase and induces early ankle dorsiflexion during the late stance phase of gait. Similarly, Cerny et al.[14] demonstrated that walking with restricted knee extension increases muscle activity and prolongs activation duration in the gluteus maximus, vastus lateralis, and soleus muscles. Furthermore, Lage et al.[15] reported increased joint power at the hip and ankle during the stance phase when knee extension is limited. At the segmental level, Robertson et al.[16] analyzed lower-limb movement and identified characteristics of segment moment power between proximal and distal segments arising from differences in angular velocity and muscle contraction patterns. Furthermore, McGibbon et al.[17] compared older adults

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with lower-limb dysfunction with healthy older adults and emphasized that understanding segment moment power patterns is critical for elucidating compensatory movement mechanisms. Although previous studies have consistently reported compensatory adaptations in adjacent joints following restricted knee extension, the underlying mechanisms—particularly from the perspective of segment moment power between segments—remain insufficiently explored. Therefore, the purpose of this study was to clarify the effects of knee joint extension restriction on segment moment power between lower-limb segments during walking. It was hypothesized that limiting knee joint extension would induce changes in power patterns among the lower-limb segments, resulting in compensatory strategies at adjacent joints.

METHODS

Participants. Sixteen healthy adult males participated in this study (mean age: 27.0 ± 5.8 years, mean height: 1.69 ± 0.05 m, mean weight: 62.6 ± 4.8 kg, mean body mass index [BMI]: 21.9 ± 1.7 kg/m2). Inclusion criteria were the absence of psychiatric, neurological, or orthopedic disorders within the preceding year and the ability to walk independently without assistive devices. Exclusion criteria included restricted knee joint extension, the presence of pain during walking, and inability to comply with the study procedures (Fig. 1). This study was approved by the Kyushu University of Nursing and Welfare Ethics Committee (approval number 31-004). All participants received a written explanation of the study purpose and procedures, and written informed consent was obtained before participation.

Task. The experimental task consisted of two walking conditions: walking without a knee brace and without any restriction in joint range of motion (hereafter referred to as the “free walking” condition) and walking with a right knee brace imposing a 20° extension restriction (hereafter referred to as the “restricted walking” condition). Data were collected three times for each condition, resulting in a total of six trials per participant. Walking speed was set at 5.0 km/h (approximately 1.39 m/s), which is considered a comfortable walking speed.[18] A metronome was used to regulate walking speed, and sufficient practice was provided beforehand to ensure that participants could walk at the prescribed speed.

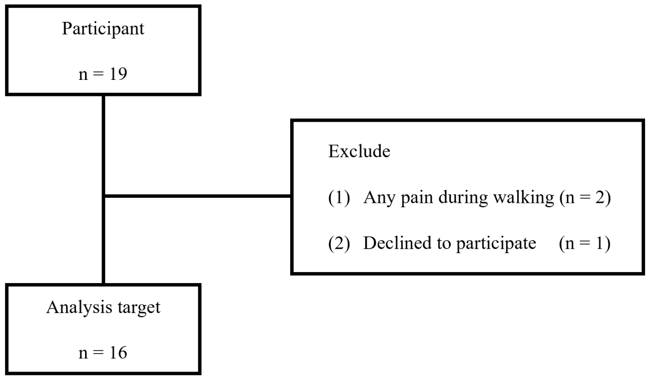

Figure 1. Inclusion criteria flowchart for study participants.

Measurement method. Gait data were collected using a VICON MX-T three-dimensional motion analysis system (Vicon Motion Systems Ltd, Oxford, UK) equipped with 10 infrared cameras, six force plates (AMTI Inc, MA, USA), and a dial-type knee brace with bilateral metal uprights that allowed angle adjustment in 10° increments (I-Ming Sanitary Materials Co Ltd, Changhua, Taiwan). The brace was adjusted to limit maximum knee extension to −20° (corresponding to 20° of flexion). The brace was applied by aligning the flexion–extension axis of the participant’s knee joint as closely as possible with the hinge axis of the orthosis, and the thigh and shank cuffs were adjusted and securely fastened. The knee joint angle was then measured in a quiet standing position using a goniometer to verify that the preset knee extension restriction had been achieved. The sampling frequencies for both the marker coordinate data and the ground reaction force data were set at 100 Hz. Reflective markers (φ14 mm) were attached to 33 anatomical landmarks in accordance with the method described by Kito et al.[19] A seven-rigid-body link model consisting of the foot, lower leg, thigh, and pelvic segments was constructed for analysis (Fig. 2).

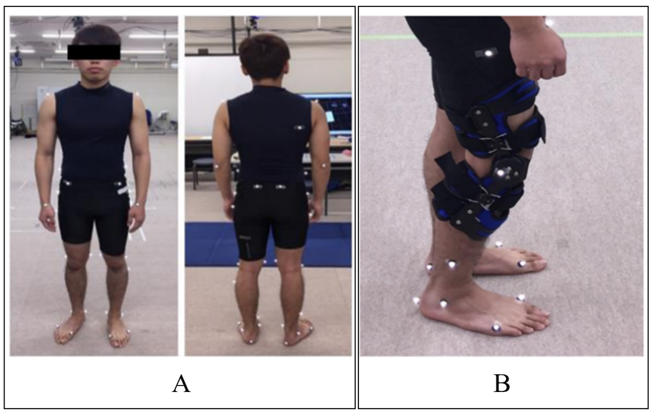

Figure 2. Measurement method.

A: Markers were attached to the left and right acromion processes, lateral epicondyles of the humerus, styloid process of the ulna, upper part of the iliac crest, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, midpoint between the greater trochanter and lateral condyle of the femur, medial condyle of the femur, lateral condyle of the femur, medial malleolus of the foot, lateral malleolus of the foot, midpoint between the head of the fibula and lateral malleolus of the foot, head of the first metatarsal, head of the fifth metatarsal, calcaneus, and right scapula. Markers were attached to the inside and outside the orthotic joint of the right knee joint. B: Dial-adjustable knee joint brace adjustment.

Analysis method. The analyzed variables included knee joint angle during walking, gait speed, sagittal-plane joint moments of the hip, knee, and ankle joints, as well as the segment moment power generated at the proximal and distal ends of each segment (pelvis, thigh, lower leg, and foot). The primary outcome measure of this study was segment moment power, whereas knee joint angle at initial contact during one gait cycle, gait speed, and joint moment served as secondary outcome measures. Three trials were recorded for each condition, and the averaged values for each participant were used in the analysis. In this study, segment moment power was used as an index to assess the transfer of mechanical energy among segments. Segment moment power was calculated as the product of joint moment and the angular velocity at each segment end, representing the temporal changes in energy (energy flow) among segments. Joint power is calculated as the product of joint moment and joint angular velocity. Robertson[16] reported that each joint power may absorb, generate, or transfer energy depending on the angular velocity of the adjacent segments. Total knee joint power was defined as the sum of the products of the knee joint moment and the angular velocities of the distal thigh and proximal shank (Fig. 3). Hip and ankle joint power were calculated using the same approach. Proximal and distal segments were defined based on their distance from the trunk. Because segment moment power can assume positive or negative values, positive values were defined as positive power (Pp) and negative values as negative power (Np). Pp represents the rate of energy inflow into the segment (an index of energy generation), whereas Np represents the rate of energy outflow from the segment (an index of energy absorption). Based on these definitions, the following parameters were analyzed: distal pelvis ($Pelvis_{distal}$ Pp, $Pelvis_{distal}$ Np), proximal thigh ($Thigh_{proximal}$ Pp, $Thigh_{proximal}$ Np), distal thigh ($Thigh_{distal}$ Pp, $Thigh_{distal}$ Np), proximal shank ($Shank_{proximal}$ Pp, $Shank_{proximal}$ Np), distal shank ($Shank_{distal}$ Pp, $Shank_{distal}$ Np), and proximal ankle ($Ankle_{proximal}$ Pp, $Ankle_{proximal}$ Np).

Before data processing, noise was removed using a Butterworth low-pass filter with cutoff frequencies of 10 Hz for ground reaction force data and 6 Hz for marker coordinate data. Data analysis was conducted using Body Builder software (version 3.6.4; Vicon Motion Systems Ltd, Oxford, UK).

The analysis interval consisted of one gait cycle of the right lower-limb. A gait cycle was defined as the period from the instant when the vertical ground reaction force of the right lower-limb first exceeded 10 N to the next occurrence of the vertical force exceeding 10 N. This interval was time normalized to 100%. Walking speed was calculated using stride length and gait cycle time from three-dimensional motion analysis. The stance phase of the right lower-limb was further divided into loading response (LR), mid-stance (MSt), terminal stance (TSt), and pre-swing (PSw), according to the classification of the Rancho Los Amigos National Rehabilitation Center.[20,21]

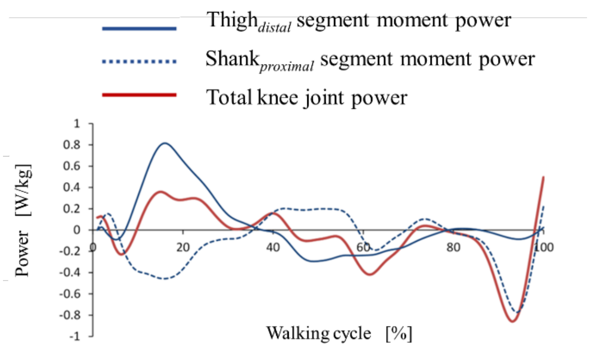

Figure 3. Segment moment power at the distal thigh and proximal shank during the gait cycle.

Statistical analysis. Statistical analyses were performed using R software (version 4.1.2; R Foundation for Statistical Computing, Vienna, Austria). Data normality was evaluated using the Shapiro–Wilk test. For variables that followed a normal distribution, paired t-tests were applied, whereas the Wilcoxon signed-rank test was used for variables that violated the assumption of normality. Knee joint angle at initial contact and mean values of joint moments and segment moment power during each gait phase— LR, MSt, TSt, and PSw—were compared between free walking and knee extension-restricted walking conditions. The level of statistical significance was set at $p < 0.05$. Sample size estimation was conducted using G*Power software (version 3.1.9.7; Heinrich Heine University, Düsseldorf, Germany). Based on a previous study investigating segment moment power during the sit-to-stand movement,[22] the significance level (α) was set at 0.05, statistical power ($1 - \beta$) at 0.80, and the effect size (r) was assumed to be 0.71. Accordingly, the minimum required sample size was estimated to be 10 participants.

RESULTS

Comparison of knee extension angle at initial contact over a gait cycle. The knee extension angle at initial contact was 4.1° ± 1.2° in the free-walking condition and 18.3° ± 2.2° in the restricted condition, with a significantly greater angle observed in the restricted condition ($p < 0.05$).

Comparison of walking speed. The mean walking speeds were 1.31 ± 0.11 m/s for the free walking condition and 1.25 ± 0.10 m/s for the restricted walking condition. No significant difference in walking speed was observed between the two conditions.

Comparison during LR. During LR, Shank$_{proximal}$ Pp was 0.05 W/kg in the free walking condition and 0.04 W/kg in the restricted walking condition, representing a statistically significant decrease in the restricted walking group ($p < 0.05$). In contrast, Shank$_{proximal}$ Np was 1.23 W/kg in the free walking condition and 1.61 W/kg in the restricted walking condition, indicating a statistically significant increase in the restricted walking condition ($p < 0.01$).

Table 1. Joint moments and segment moment powers during LR.

Variables	Free walking	Restricted walking	Mean difference	Median difference	95% CI	*p*	Effect size
Mean joint moment (Nm/kg)							
Hip extension	0.10(0.01, 0.34)	−0.15(−0.35, −0.01)	−	0.05	−0.19, −0.03	0.01**	0.88
Knee extension	0.47(0.34, 1.21)	0.69(0.46, 1.00)	−	0.22	0.04, 0.34	0.01**	0.69
Ankle plantar-flexion	−0.27 ± 0.05	−0.30 ± 0.05	0.03	−	−0.01, 0.04	0.06	0.51
Mean segment moment power (W/kg)							
Pelvis$_{distal}$ Pp	0.00(0.00, 0.01)	0.00(0.00, 0.02)	−	0.00	0.00, 0.00	0.82	0.08
Pelvis$_{distal}$ Np	0.36 ± 0.23	0.33 ± 0.20	0.03	−	−0.10, 0.02	0.24	0.14
Thigh$_{proximal}$ Pp	0.03(0.00, 0.16)	0.05(0.00, 0.20)	−	0.02	−0.02, 0.45	0.16	0.36
Thigh$_{proximal}$ Np	0.49 ± 0.25	0.42 ± 0.18	0.07	−	−0.21, 0.08	0.35	0.30
Thigh$_{distal}$ Pp	0.85 ± 0.33	0.89 ± 0.37	0.04	−	−0.27, 0.18	0.72	0.11
Thigh$_{distal}$ Np	0.00(0.00, 0.03)	0.00(0.00, 0.15)	−	0.00	−0.01, 0.01	1.00	0.00
Shank$_{proximal}$ Pp	0.05(0.00, 0.27)	0.04(0.00, 0.11)	−	0.00	−0.08, 0.00	0.02*	0.60
Shank$_{proximal}$ Np	1.23(0.63, 3.15)	1.61(1.01, 3.45)	−	0.38	−0.63, −0.07	0.01**	0.65
Shank$_{distal}$ Pp	0.69 ± 0.20	0.72 ± 0.22	0.03	−	−0.11, 0.05	0.44	0.14
Shank$_{distal}$ Np	0.00	0.00	−	−	−	−	−
Ankle$_{proximal}$ Pp	0.15(0.01, 0.46)	0.13(0.01, 0.21)	−	0.02	−0.11, 0.00	0.08	0.44
Ankle$_{proximal}$ Np	0.32(0.02, 1.00)	0.43(0.02, 1.02)	−	0.11	−0.03, 0.00	0.05	0.50

*<0.05, **<0.01, CI: Confidence interval, Effect size: r, Mean ± SD, Median(minimum value,maximum value)
Np values are presented as absolute values.

Comparison during MSt. Thigh$_{proximal}$ Np was 0.71 W/kg in the free walking condition and 0.86 W/kg in the restricted walking condition, reflecting a statistically significant increase in the restricted walking condition ($p < 0.05$).

Table 2. Joint moments and segment moment powers during MSt.

Variables	Free walking	Restricted walking	Mean difference	Median difference	95% CI	*p*	Effect size
Mean joint moment (Nm/kg)							
Hip extension	−0.26±0.14	−0.43±0.19	−	0.17	0.06, 0.27	0.01**	0.71
Knee extension	0.51±0.14	0.65±0.17	−	0.14	−0.21, −0.06	0.01**	0.67
Ankle plantar-flexion	0.17±0.11	0.20±0.11	−	0.03	−0.08, 0.02	0.19	0.26
Mean segment moment power (W/kg)							
Pelvis$_{distal}$ Pp	0.33±0.09	0.31±0.09	−	0.02	−0.03, 0.07	0.39	0.21
Pelvis$_{distal}$ Np	0.03(0.00, 0.10)	0.01(0.00, 0.12)	0.02	−	−0.02, 0.03	0.75	0.10
Thigh$_{proximal}$ Pp	0.00(0.00, 0.21)	0.00(0.00, 0.00)	0.00	−	0.00, 0.00	1.00	0.25
Thigh$_{proximal}$ Np	0.71±0.24	0.86±0.21	−	0.15	0.00, 0.30	0.04*	0.56
Thigh$_{distal}$ Pp	1.16±0.32	1.22±0.39	−	0.06	−0.27, 0.16	0.59	0.16
Thigh$_{distal}$ Np	0.00(0.00, 0.08)	0.00(0.00, 0.19)	0.00	−	−0.03, 0.00	0.06	0.63
Shank$_{proximal}$ Pp	0.04±0.03	0.03±0.06	−	0.01	0.00, 0.03	0.27	0.21
Shank$_{proximal}$ Np	0.53±0.17	0.66±0.27	−	0.13	0.00, 0.27	0.05	0.50
Shank$_{distal}$ Pp	0.05(0.00, 0.14)	0.04(0.00, 0.17)	0.01	−	−0.04, 0.03	0.91	0.03
Shank$_{distal}$ Np	0.31±0.13	0.30±0.12	−	0.01	−0.07, 0.06	0.89	0.08
Ankle$_{proximal}$ Pp	0.14(0.10, 0.40)	0.16(0.07, 0.39)	0.02	−	−0.06, 0.07	0.69	0.13
Ankle$_{proximal}$ Np	0.00(0.00, 0.15)	0.00(0.00, 0.02)	0.00	−	0.00, 0.00	1.00	0.09

*<0.05, **<0.01, CI: Confidence interval, Effect size: r, Mean ± SD, Median(minimum value,maximum value)
Np values are presented as absolute values.

Comparison during TSt. Thigh$_{proximal}$ Np was 0.65 W/kg in the free walking condition and 0.44 W/kg in the restricted walking condition, showing a statistically significant decrease in the restricted walking condition ($p < 0.01$). In contrast, Thigh$_{distal}$ Np was 0.16 W/kg in the free walking condition and 0.59 W/kg in the restricted walking condition, indicating a statistically significant increase in the restricted walking condition ($p < 0.01$). Shank$_{distal}$ Np was 2.81 W/kg in the free walking condition and 2.45 W/kg in the restricted walking condition, showing a statistically significant decrease in the restricted walking condition ($p < 0.01$).

Table 3. Joint moments and segment moment powers during TSt.

Variables	Free walking	Restricted walking	Mean difference	Median difference	95% CI	*p*	Effect size
Mean joint moment (Nm/kg)							
Hip extension	−0.73±0.16	−0.80±0.18	−	0.07	−0.04, 0.19	0.16	0.38
Knee extension	0.15±0.10	0.27±0.15	−	0.12	−0.19, −0.05	0.01**	0.69
Ankle plantar-flexion	1.00±0.08	0.99±0.09	−	0.01	−0.04, 0.05	0.75	0.12
Mean segment moment power (W/kg)							
Pelvis$_{distal}$ Pp	0.12(0.00, 0.36)	0.08(0.01, 0.28)	0.04	−	−0.10, 0.00	0.15	0.36
Pelvis$_{distal}$ Np	0.05(0.00, 0.23)	0.05(0.00, 0.14)	0.00	−	−0.05, 0.05	0.75	0.09
Thigh$_{proximal}$ Pp	0.03(0.00, 0.21)	0.06(0.00, 0.52)	0.03	−	−0.02, 0.08	0.09	0.49
Thigh$_{proximal}$ Np	0.65±0.27	0.44±0.24	−	0.21	0.00, 0.30	0.01**	0.63
Thigh$_{distal}$ Pp	0.01(0.00, 0.59)	0.01(0.00, 0.63)	0.00	−	−0.07, 0.09	1.00	0.02
Thigh$_{distal}$ Np	0.16(0.00, 0.70)	0.59(0.00, 1.10)	0.43	−	−0.59, 0.03	0.01**	0.67
Shank$_{proximal}$ Pp	0.38±0.25	0.30±0.22	−	0.08	−0.05, 0.21	0.21	0.32
Shank$_{proximal}$ Np	0.00(0.00, 0.37)	0.01(0.00, 0.38)	0.01	−	−0.04, 0.01	0.55	0.21
Shank$_{distal}$ Pp	0.00±0.00	0.00±0.00	−	−	−	−	−
Shank$_{distal}$ Np	2.81±0.50	2.45±0.57	−	0.36	−0.64, −0.07	0.02*	0.56
Ankle$_{proximal}$ Pp	2.66±0.84	2.65±0.75	−	0.01	−0.42, 0.45	0.94	0.01
Ankle$_{proximal}$ Np	0.00±0.01	0.00±0.00	−	0.00	−0.01, 0.00	0.45	0.00

*<0.05, **<0.01, CI: Confidence interval, Effect size: r, Mean ± SD, Median(minimum value,maximum value)
Np values are presented as absolute values.

Comparison during PSw. No statistically significant differences in segment moment power were observed between the two walking conditions.

Table 4. Joint moments and segment moment powers during PSw.

Variables	Free walking	Restricted walking	Mean difference	Median difference	95% CI	*p*	Effect size
Mean joint moment (Nm/kg)							
Hip extension	−0.51±0.13	−0.52±0.10	−	0.01	−0.08, 0.09	0.83	0.28
Knee extension	0.26±0.06	0.31±0.10	−	0.05	−0.09, −0.02	0.01**	0.52
Ankle plantar-flexion	0.56±0.09	0.54±0.10	−	0.02	−0.04, 0.08	0.44	0.21
Mean segment moment power (W/kg)							
Pelvis$_{distal}$ Pp	0.34±0.14	0.26±0.14	−	0.02	−0.01, 0.17	0.08	0.50
Pelvis$_{distal}$ Np	0.00(0.00, 0.01)	0.00(0.00, 0.04)	0.00	−	0.00, 0.00	0.34	0.54
Thigh$_{proximal}$ Pp	0.93±0.30	0.80±0.24	−	0.13	−0.05, 0.31	0.16	0.43
Thigh$_{proximal}$ Np	0.00(0.00, 0.06)	0.00(0.00, 0.06)	0.00	−	0.00, 0.00	0.78	0.18
Thigh$_{distal}$ Pp	0.00(0.00, 0.02)	0.00(0.00, 0.07)	0.00	−	0.00, 0.00	1.00	0.00
Thigh$_{distal}$ Np	0.71±0.23	0.71±0.29	−	0.00	−0.16, 0.15	0.97	0.00
Shank$_{proximal}$ Pp	0.29±0.23	0.33±0.23	−	0.01	−0.18, 0.11	0.60	0.16
Shank$_{proximal}$ Np	0.06(0.00, 0.21)	0.10(0.00, 0.20)	0.04	−	−0.07, 0.02	0.46	0.19
Shank$_{distal}$ Pp	0.00(0.00, 0.03)	0.01(0.00, 0.03)	0.01	−	−0.01, 0.01	0.27	0.35
Shank$_{distal}$ Np	2.71±0.58	2.28±0.84	−	0.43	−0.95, 0.09	0.09	0.51
Ankle$_{proximal}$ Pp	4.00±1.19	3.36±1.27	−	0.04	−0.22, 1.51	0.14	0.47
Ankle$_{proximal}$ Np	0.02(0.00, 0.07)	0.01(0.00, 0.05)	0.01	−	0.01, 0.02	0.83	0.07

*<0.05, **<0.01, CI: Confidence interval, Effect size: r, Mean ± SD, Median(minimum value,maximum value)
Np values are presented as absolute values.

DISCUSSION

The purpose of this study was to clarify the characteristics of segment moment power among lower-limb segments during walking with knee extension restriction. Segment moment power values (Pp and Np) were calculated as phase-averaged values for each gait phase (LR, MSt, TSt, and PSw) and compared between conditions.

Compared with the free walking condition, the restricted walking condition demonstrated a significant decrease in Shank$_{proximal}$ Pp and a significant increase in Shank$_{proximal}$ Np during LR. During MSt, Thigh$_{proximal}$ Np increased significantly. Furthermore, during TSt, Thigh$_{proximal}$ Np and Shank$_{distal}$ Np decreased significantly, whereas Thigh$_{distal}$ Np increased significantly. These phase-specific changes were accompanied by increases in joint moments, suggesting that different lower-limb segments functioned as energy suppliers at distinct gait phases as part of a compensatory strategy in response to knee extension restriction.

Characteristics of knee extension restriction. The knee extension angle at initial contact was significantly greater in the restricted walking condition than in the free walking condition. These results suggest that the knee brace functioned approximately as intended, effectively maintaining the knee in a more flexed position during walking.

Characteristics of LR. During LR, the restricted walking condition exhibited a reduced phase-averaged $Shank_{proximal}$ Pp and a significantly increased $Shank_{proximal}$ Np compared with free walking. This pattern indicates a decreased power representing the rate of energy inflow into the proximal shank and an increased power representing the rate of energy outflow from this segment throughout LR, suggesting that the proximal shank showed a relatively greater contribution as an energy source.

Because no significant differences were observed in phase-averaged power of the adjacent distal thigh segment, the energy released from the proximal shank likely contributed primarily to increased joint moments via activation of the knee extensor muscles. Winter[23] described LR as a phase requiring mechanical energy absorption at the lower-limb joints to attenuate impact immediately after initial contact. Nagano et al.[24] reported that increasing knee extension restriction results in elevated knee extension moments that facilitate shock absorption. Furthermore, Sotelo et al.[25] suggested that walking with restricted knee extension may impair ankle function, whereas McGibbon[26] reported that insufficient shock absorption at distal joints can induce compensatory reliance on more proximal segments.

Collectively, these findings suggest that, under restricted walking conditions, the proximal shank served as an energy source for the knee extensors to compensate for reduced ankle shock absorption, thereby enhancing impact attenuation at the knee joint.

Characteristics of MSt. During MSt, the restricted walking condition demonstrated a significant increase in phase-averaged $Thigh_{proximal}$ Np compared with free walking. This finding reflects an increased power representing the rate of energy outflow from the proximal thigh throughout MSt, indicating that the proximal thigh showed an enhanced contribution as an energy source.

MSt is a single-limb support phase that requires stabilization of the body's center of mass while maintaining forward progression. Fukui et al.[27] demonstrated that the knee extension moment plays a critical role in body support during MSt. However, when knee extension is restricted, increased trunk forward lean and reduced vertical forces acting on the knee joint have been reported.[28,29] Under such conditions, reliance on the knee joint for postural support is diminished, necessitating compensatory contributions from the hip joint and surrounding musculature.

The present results suggest that, in the restricted walking condition, the proximal thigh showed a relatively increased contribution as an energy source for the muscles around the hip and knee joints, thereby possibly compensating for stabilization and forward progression of the body center of mass.

Characteristics of TSt. During TSt, the restricted walking condition exhibited decreased phase-averaged $Thigh_{proximal}$ Np and $Shank_{distal}$ Np, whereas $Thigh_{distal}$ Np increased significantly. These findings indicate reduced contributions of the proximal thigh and distal shank as energy suppliers, accompanied by a relatively increased contribution of the distal thigh as an energy supplier during TSt.

TSt is typically a propulsive phase in which forward progression of the center of mass is achieved through coordinated ankle plantarflexion and hip extension moments.[30–36] However, under conditions of knee extension restriction, propulsive function at the hip and ankle has been reported to decline.[13,23] Robertson et al.[16] reported that when energy generation or absorption at a specific joint is reduced, changes in power patterns occur in adjacent joints. In the present study, increased knee extension moments were observed during restricted walking, suggesting that compensatory forward propulsion via knee extension occurred to offset reduced contributions from the hip and ankle. The increased $Thigh_{distal}$ Np likely reflects an enhanced role of the distal thigh as an energy supplier within this compensatory mechanism.

Characteristics of PSw. During PSw, no significant differences were observed in phase-averaged segment moment power between the free walking and restricted walking conditions. This finding suggests that intersegmental changes in power patterns occur selectively during gait phases requiring substantial energy absorption and transmission, particularly during the support and propulsion phases of stance, rather than uniformly throughout the entire stance phase.

This study has several limitations. First, the primary indicator used, segment moment power (rate of work), does not directly quantify the absolute amount of energy or the magnitude of energy transfer between segments. Therefore, the findings reflect changes in power patterns rather than direct evidence of energy transfer or redistribution, which should be considered when interpreting the results. Second, statistical analyses were performed across multiple gait phases and mechanical variables; therefore, the potential influence of type I error due to multiple comparisons cannot be entirely ruled out.

CONCLUSION

This study examined the effects of knee extension restriction on segment moment power among lower-limb segments during walking. Although joint moments were altered throughout the stance phase, segment moment power demonstrated phase-specific patterns: increased negative segment moment power at the $\text{Shank}_{proximal}$ during LR, at the $\text{Thigh}_{proximal}$ during MSt, and at the Thigh_{distal} during TSt, each contributing to increased joint moments within the corresponding phase. These findings indicate that, during walking with knee extension restriction, compensatory mechanisms are characterized not only by modifications in joint moments but also by changes in power patterns across lower-limb segments.

AUTHOR CONTRIBUTIONS

MS and HK contributed to the conception and planning of the study. MS wrote the manuscript. HK supervised and advised on data analysis and contributed to the interpretation of the results. All authors reviewed the final manuscript and agreed to its publication.

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CONFLICT OF INTEREST

The authors have no conflicts of interest to disclose.

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