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The interrelationship between lower limb movement, muscle activity, and joint moment during half squat and gait

Umi Matsumura [a], Toshiya Tsurusaki [a,*], Rena Ogusu [a], Shimpei Yamamoto [b], Yeonghee Lee [a], Shinya Sunagawa [a,c], W Darlene Reid [d,e,f], Hironobu Koseki [a]

[a] *Department of Health Sciences, Nagasaki University Graduate School of Biomedical Sciences, Nagasaki, Japan*
[b] *Department of Rehabilitation, Fukuoka University Chikushi Hospital, Fukuoka, Japan*
[c] *Department of Rehabilitation, Wajinkai Hospital, Nagasaki, Japan*
[d] *Department of Physical Therapy, University of Toronto, Toronto, ON, Canada*
[e] *Interdepartmental Division of Critical Care Medicine, University of Toronto, Toronto, ON, Canada*
[f] *KITE, Toronto Rehabilitation Institute, University Health Network, Toronto, ON, Canada*

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ABSTRACT

Joint moment is the resultant force of limb movements. However, estimation methods for joint moments using surface electromyography frequently use joint angles instead of limb angles. The limb angle in joint moment estimation using electromyography could clarify the effects of muscle activity on the limbs: acceleration, deceleration, or stabilization. No study has quantified the comprehensive relationship between limb movement, muscle activity, and joint moment. This study aimed to determine the influencing factors for ankle-joint moment and knee-joint moment in the sagittal plane among muscle activities and parameters related to limb movements during half squat and gait. This study included 29 healthy adults (16 female participants, 21.1 ± 2.09 years). Using inertial measurement units, thigh, shank, and foot inclination angles and angular accelerations were calculated as the parameters of limb movements. Muscle activations of the biceps femoris long head, rectus femoris, gastrocnemius, and tibialis anterior were measured. Ankle joint moment and knee-joint moment were measured using a three-dimensional motion capture system and two force plates. Regression models showed high accuracy in measuring ankle-joint moment during a half squat and gait ($R_f^2 = 0.92$, 0.97, respectively) and knee-joint moment during a half squat ($R_f^2 = 0.98$), but not knee-joint moment during gait ($R_f^2 = 0.63$). However, only a maximum of five parameters were selected from muscle activities and limb angular information. Tibialis anterior and gastrocnemius activity were the largest contributors to ankle-joint moment during a half squat and gait, respectively, while muscle activities were not directly reflected in the knee-joint moment during either movement. Consideration of the interrelationships among limb movement, muscle activity, and joint moment is required when adjusting joint movements according to the target and aim of the therapeutic interventions.

* Corresponding author. Department of Health Sciences, Nagasaki University Graduate School of Biomedical Sciences, 1-7-1 Sakamoto, Nagasaki City, 852-8520, Japan.
E-mail address: toshiya@nagasaki-u.ac.jp (T. Tsurusaki).

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1. Introduction

Motion analysis is performed to characterize motor dysfunction and evaluate the effectiveness of the intervention [1]. Joint moment is the resultant force estimated from limb movement and can be adjusted by limb rotation [2]. For example, an increased forward shank inclination (knee in front of the toes) increases knee-joint moment (KJM). It also increases the patellofemoral joint compressive force, which can be the cause of patellofemoral pain syndrome, during squat [3,4]. In contrast, although reduced foot dorsiflexion during squat is associated with decreased KJM, it increases hip extensor moment and lumbar lordosis, which can increase the load on the hip joint and lower back [5]. Thus, to prevent adverse effects or enhance training in rehabilitation, it is critical to adjust the joint moment by considering specific limb rotation according to the characteristics and purpose of the target.

To capture limb movements and estimate joint moments, three-dimensional motion capture systems and force plates have been used in the laboratory as the gold standard for quantitative motion analysis [3–6]. The clinical use of these devices is limited due to their cost, lack of portability, and requirement for a sensitive measurement environment [1,7–9]. In recent years, inertial measurement units (IMUs) comprising accelerometers, gyroscopes, and magnetometers have attracted attention as alternative motion capture systems due to their small size, low cost, and versatility [1,7–9]. The accuracy of IMUs in capturing limb movements has been highlighted [1,9]. Furthermore, some studies have reported that IMUs can predict joint moments accurately even without laboratory equipment [7,8,10].

Muscle activity is the primary mechanism that regulates limb movement [11]. In addition, muscle co-contraction fixates the limb to maintain dynamic stability [6,12]. Excessive or deficient muscle activities can cause limb instability that can lead to chronic joint injuries [13] or affect movement performances and efficacies, resulting in reduced gait speed or increased energy consumption [14, 15]. Surface electromyography (EMG) reflects the magnitude of muscle activity and can be easily used in clinical practice [2,14,16]. Since the relationship between EMG activity and force is non-linear, the EMG-driven forward-dynamics approach has been used for estimating joint moments based on muscle activities. However, this model requires consideration of many anatomical and physiological characteristics, and the estimation is dependent on calibrations, which could reduce its clinical applicability [2,14,16].

With the development of technology, artificial neural networks and machine learning models have been used for processing and analyzing biomedical signals [13,17–22]. Regarding EMG signals, joint moment estimations have been performed using these sophisticated algorithms to overcome the limitations of the EMG drive model [13,17–19]. Xiong et al. estimated lower limb joint moments in the sagittal plane using EMG signals (4–6) and joint angles (4–5) during treadmill walking serving as the inputs of artificial neural network [18]. Li et al. estimated KJM based on EMG signals from the antagonist and agonist muscles of the knee, knee joint angle, and knee joint angular velocity as inputs of the neural network [19]. Camargo et al. used machine learning models to predict joint moments in real-time combined with IMUs, EMG, and electrogoniometers [10]. They showed a neural network or machine learning-based approach using wearable sensors with fewer errors in real-time joint moment predictions [13,17–19]. However, previous studies used joint angles instead of limb angles as kinematic parameters [13,18,19]. Joint angle is calculated by integrating the difference in the angular velocities of adjacent body segments [10]. Gravity affects the generation of joint moments in the sagittal plane [11]. Considering limb angles rather than joint angles clarifies whether limb movement occurs in the same direction as gravity or against it during motion [6–11]. In addition, angular acceleration offers insights into whether muscle activities serve to accelerate, decelerate, or stabilize limb movement in response to the momentum generated by gravity. To the best of our knowledge, no study has quantified the interrelationship between limb movement, muscle activity, and joint moment. Clarifying the interplay of limb movement, muscle activity, and joint moment could not only evaluate the parameters that contribute to joint moment generation but also delineate the contribution of these parameters. A previous simulation study revealed that plantar flexor activities are generated to regulate ankle-joint moment in the sagittal plane and are necessary to maintain dynamic balance during gait [11]. Thus, if the plantar flexor activities are weak, therapeutic interventions that increase these muscle activities may improve balance during gait or address movement disorders.

Our purpose was to determine the influencing factors for ankle-joint moment (AJM) and KJM among muscle activities and parameters related to limb movements when performing motion analysis using a combination of IMUs and EMG. In addition, the interrelationship between limb movement, muscle activity, and joint moment was inferred. Since increasing the EMG signals could increase the complexity of the model and the potential for contamination with motion artifacts, it is desirable to keep the number of sensors as low as possible [20]. Li et al. revealed that muscle pairs did not affect the prediction accuracy in six combinations of agonist and antagonist muscles for joint moment prediction [19]. Thus, we focused on only one agonist muscle and one antagonist muscle that work in tandem to allow ankle and knee joint movements as parameters for joint moment estimation: biceps femoris long head (BFL), rectus femoris (RF), gastrocnemius (GAS), and tibialis anterior (TA). We used the limb inclination angles and the angular accelerations, calculated by IMUs, as the parameters that reflect limb movements. Although previous studies have mostly examined walking at a constant speed on a treadmill [18,19], this study examined squat and gait on the ground, which are frequently used in therapeutic interventions. A half squat, rather than a full squat, was assessed to minimize head and trunk movements [6]. It was hypothesized that 1) TA activity would contribute to the generation of the AJM required to stabilize the foot at half squat, 2) GAS activity would contribute to the generation of the AJM for forward momentum during gait, 3) the activity of the RF would contribute to the generation of KJM by affecting the movement direction of the thigh and shank during a half squat, and 4) RF activity would also contribute to the generation of KJM, suppressing the thigh swing during the stance phase of gait.

2. Materials and methods

2.1. Participants

This study was conducted according to the guidelines of the Declaration of Helsinki and was approved by the Ethics Committee of the Graduate School of Medical Facilities, Nagasaki University, Nagasaki, Japan [18,061,429]. Written informed consent was obtained from all participants after explaining the study.

Twenty-nine healthy adults (13 male and 16 female participants) participated in this cross-sectional study between April 2018 and March 2021. The inclusion criteria were as follows: (1) 18–30 years of age, (2) without current orthopedic or neurological disorders, and (3) body mass index (BMI) within 18–25 kg/m^2. The exclusion criteria were as follows: (1) presence of musculoskeletal or neurological diseases, (2) pain interfering with motion.

2.2. Data acquisition

Angular information regarding the lower limbs was calculated using IMUs (LP-WSD1101-OA, 5G/300dps, LOGICAL PRODUCT, Fukuoka, Japan) sampled at 1000 Hz to represent the limb rotations. Three IMUs were mounted on the lateral aspect of the participant's right leg using double-sided adhesive tape with their axis of rotation perpendicular to the sagittal plane. Attachment positions were: *thigh*, halfway between the greater trochanter and lateral femoral condyle; *shank*, halfway between the lateral femoral condyle and lateral malleolus; and *foot*, middle of the dorsum of the foot (Fig. 1) [9].

Four major muscle activities representing ankle dorsiflexion/plantarflexion and knee extension/flexion were measured using surface EMG (LP-IW2PAD, LOGICAL PRODUCT, Fukuoka, Japan) with a 4-channel system (LP-WSD1002-OA, LOGICAL PRODUCT, Fukuoka, Japan). Bipolar Ag/AgCl EMG electrodes (VL-00-S/25, METS, 1–7, Tokyo, Japan) were positioned on the right side of the BFL, RF, GAS, and TA, as in previous studies [6,11,15], according to the Surface EMG for Non-Invasive Assessment of Muscles guidelines (Fig. 1) [23]. Before the electrodes were attached, the skin was shaved and sanitized with alcohol. When errors caused by dynamic contractions during clinical measurement were considered, the interelectrode distance was 30 mm, which was significantly wider than the standard 20 mm [24]. Physical therapists palpated the muscle valleys to prevent artifacts caused by muscle crosstalk. EMG data were digitally sampled at 1000 Hz, actual gain of × 500, and then passed to the 16-bit analog-to-digital converter and personal computer.

A six-camera motion capture system with two force plates (MA-3000, ANIMA, Tokyo, Japan) was used at 60 Hz to obtain the AJM

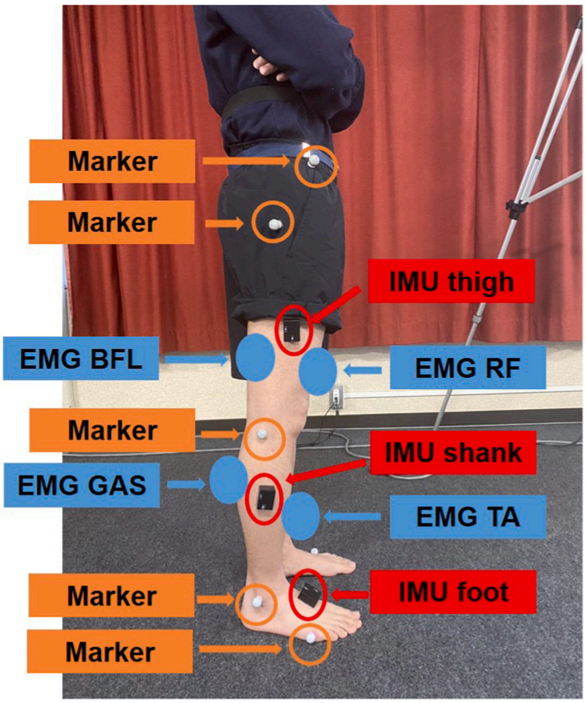

Fig. 1. EMG and IMU sensor placement
EMG electrodes are positioned on four major muscles in the right lower limb. IMUs are positioned on each segment in the right lower limb. Attachment positions of reflective markers are the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, and fifth metatarsal bone. EMG: electromyography, RF: rectus femoris, BFL: biceps femoris long head, GAS: gastrocnemius, TA: tibialis anterior, IMU: inertial measurement units.

and KJM, respectively, in the sagittal plane. AJM and KJM were calculated by inverse dynamic estimation according to previous studies [3,6,7]. Reflective markers were affixed to the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, and fifth metatarsal bone (Fig. 1) [4,6,14,25]. Using a wireless 8-channel logger (LP-WSD1311-OA, LOGICAL PRODUCT, Fukuoka, Japan), all data were time-axis synchronized and downsampled to a video camera scale of 1/30 s [26], assuming quantification of limb rotations using a video camera in the future.

2.3. Experimental Procedure

The participants were dressed in shorts, and measurements were performed barefoot. The leg width during the squat was set at shoulder width, and the squat was performed with the arms folded in front of the chest. The participants were instructed to keep their head and trunk vertical over the hip joints (not bending the head and trunk) to minimize head and trunk movements during the half squat [6]. Before starting the task, the participants were familiarized with the half squat, and physical therapists provided feedback on the motion. Half squats and gait were measured at two speeds: natural speed and fast (as quickly as possible). The participants performed half squats thrice and walked three gait cycles. A 1-min rest period was provided between each trial to minimize the effect of fatigue. After the trial, the participants performed a maximal voluntary isometric contraction (MVIC) task.

The MVIC trial of each muscle was carried out at the specific positions according to the manual muscle test [25]. The physical therapist added resistance against the participants while holding the specific position. Verbal encouragement was provided 2 s after starting the MVIC trial, ensuring that maximal effort was exerted, and the position was held for the last 3 s of a total of 5 s. The average values from the 0.5-s window during maximal effort were used as the reference MVIC value [27].

2.4. Data processing

In both half squat and gait, muscle activities, limb angular information, and joint moments were measured from the start to the end of the motions, and each parameter was converted 100 % from the start to the end of the motion.

The angular velocities in the sagittal plane, measured using the IMUs ω_j (deg/s), were integrated with time displacement dt (s) to calculate the limb inclination angles [8]. The lineal resetting mechanism was used through weighting linearly during performing integration to eradicate measurement errors resulting from noise and drift.

θ_j (deg). j = 1–3 1: *Thigh*, 2: *Shank*, 3: *Foot*

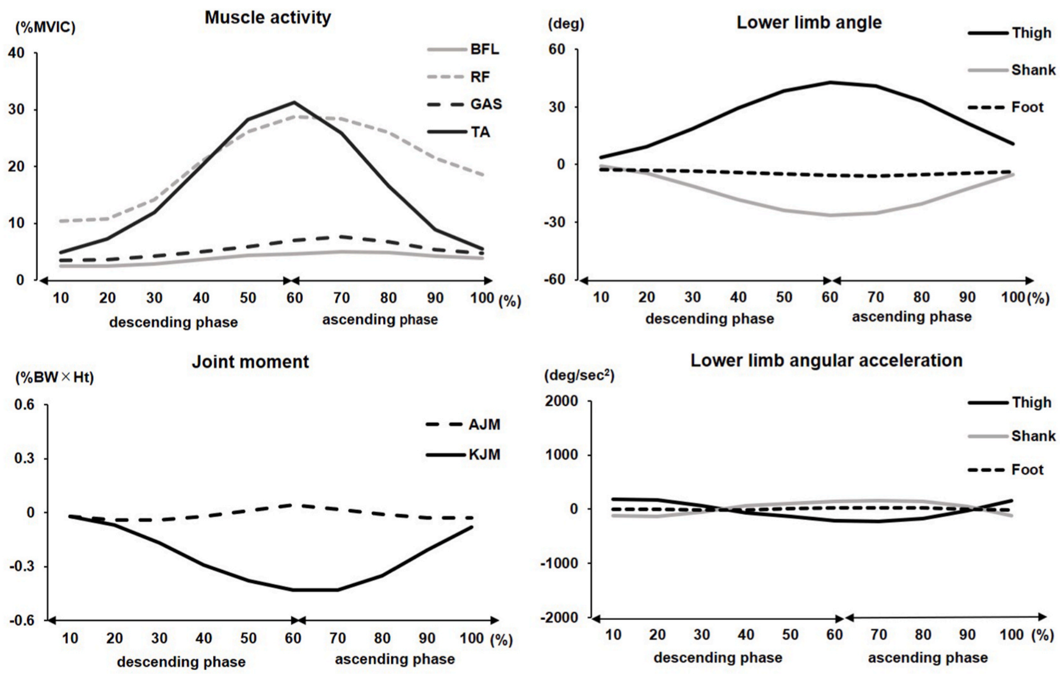

Fig. 2. Transitions of each parameter during half squat
The lower limb angles are positive for counterclockwise rotation and negative for clockwise rotation regarding the direction of gravity. Joint moments are positive for knee extension and ankle plantar flexion and negative for knee flexion and ankle dorsiflexion. Data are the average values of one trial converted to 100 % of the normal speed of all participants. The descending phase occupies 60 % of the movement duration, followed by the ascending phase.
MVIC: maximal voluntary isometric contraction, BFL: biceps femoris long head, RF: rectus femoris, GAS: gastrocnemius, TA: tibialis anterior, AJM: ankle-joint moment, KJM: knee-joint moment, BW: body weight, Ht: height.

$$\theta_j = \int \omega_j(t) \times dt$$

A Kalman filter was applied to minimize the integration error common to the IMUs [9]. The inclination angles were then low-passed at 6 Hz using a zero-phase 4th-order Butterworth filter [13]. Since the acquisition of the inclination angles using videography in the future was assumed, the angular accelerations a_j (deg/s^2) were converted from the calculated angles θ_j by double differentiation [8], as follows:

$$a_j = d^2\theta_j / dt^2$$

EMG signals were band-pass filtered at 20–450 Hz, full-wave rectified, and then low-pass filtered at 6 Hz with a 4th-order zero-lag Butterworth filter [6]. EMG signals were subsequently normalized based on the average EMG measurement of the MVIC.

Joint moments were normalized as a percentage of the participant's body weight, multiplied by body height expressed in meters (% BW $\times$ Ht) to eliminate the influence of body size.

2.5. Statistical analyses

Statistical analyses were performed using JMP Pro 15 software (SAS INSTITUTE JAPAN, Tokyo, Japan). The sample size was calculated using a significance level of 0.05, a standard deviation of 0.5, an effect size of 0.3, and a detection power of 0.8. Paired t-tests were used for investigating the differences at each point for each parameter due to motion speed. Stepwise multiple regression analysis was performed to assess the contribution of lower limb angular information and muscle activities to joint moments. AJM and KJM were targeted variables, and the angular information of each of the three limbs and four muscle activities were explanatory variables. The coefficient of determination, adjusted for degrees of freedom (R_f^2), was used for indicating the degree of adequacy of the regression models. Standardized regression coefficients (standard β) were calculated to compare the degree of influence of muscle activities and limb angular information on joint moments. Statistical significance was set at $p < 0.05$.

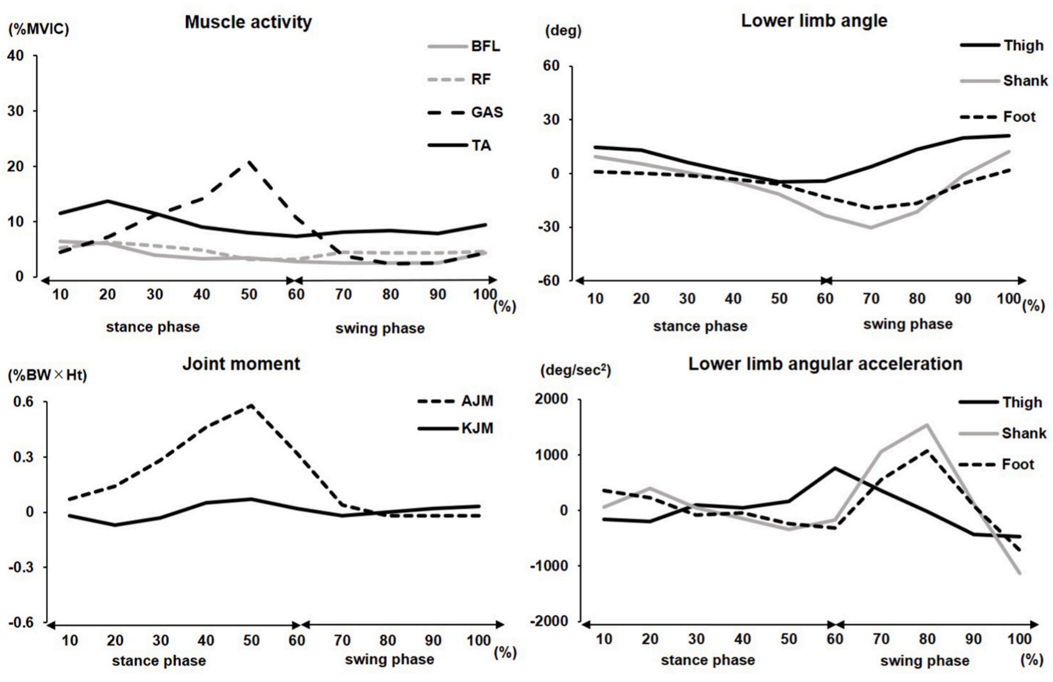

Fig. 3. Transitions of each parameter during a gait cycle
The lower limb angles are positive for counterclockwise rotation and negative for clockwise rotation regarding the direction of gravity. Joint moments are positive for knee extension and ankle plantar flexion and negative for knee flexion and ankle dorsiflexion. Data are the average values of one gait cycle converted to 100 % of the normal speed of all participants. The stance phase occupies 60 % of the movement duration, followed by the swing phase.
MVIC: maximal voluntary isometric contraction BFL: biceps femoris long head, RF: rectus femoris, GAS: gastrocnemius, TA: tibialis anterior, AJM: ankle-joint moment, KJM: knee-joint moment, BW: body weight, Ht: height.

3. Results

3.1. Descriptive characteristics of participants and measured motions

The participant characteristics were: age (years), 21.1 ± 2.09; sex (males/females), 13/16; height (m), 1.64 ± 0.10; weight (kg), 58.6 ± 13.3; BMI (kg/m^2), 21.4 ± 2.8. Changes in muscle activities, limb angles, limb angular accelerations, and joint moments during half squat and gate are shown in Figs. 2 and 3, respectively.

Table 1 shows the difference due to the motion speed of each parameter at the peak time as an example. BFL and RF activities increased at 50–90 % of the motion ($p \leqq 0.034$), while TA activity increased at 0–30 % of the motion ($p \leqq 0.025$) during fast-speed half squat compared to normal speed. KJM of fast-speed half squat increased at 40–70 % of the motion ($p \leqq 0.035$). The angular acceleration of each segment increased at a fast-speed squat before and after the transition from the descending phase to the ascending phase ($p \leqq 0.013$).

The comparison between natural speed and fast gait showed no significant differences among all four muscle activities and AJM or KJM ($p > 0.05$). Motion speed had negligible effects on the lower limb angles for gait.

3.2. Contributed factors of joint moments during half squat

GAS and TA activities and shank and foot angular information contributed to AJM. The greatest contributor to AJM was TA activity (standard $\beta = 1.1$). Only the shank angle and thigh angular acceleration contributed to KJM. The greatest contributor to KJM was the shank angle (standard $\beta = 0.83$). The coefficient of determination was quite high for both AJM and KJM (AJM: R_f^2; 0.92, KJM: R_f^2; 0.98) (Table 2).

3.3. Contributing factors to joint moments during gait

GAS activity, thigh angle, and angular acceleration contributed to AJM with high accuracy ($R_f^2 = 0.97$), with the highest contribution from GAS activity (standard $\beta = 0.65$). The muscle activities of the GAS and TA, thigh, and shank angular information were determined to be contributors to the KJM. The greatest contributor to KJM was the shank angle (standard $\beta = -0.85$). The degree of adequacy of the regression model was lower than AJM ($R_f^2 = 0.63$) (Table 3).

4. Discussion

This was the first study considering limb angle and angular acceleration as the kinematic parameters instead of joint angle when estimating joint moment using EMG signals during half squat and gait. Considering limb angle and angular acceleration could provide insight into how muscle activity affects limb movements that generate joint moments (accelerate, decelerate, or stabilize the limb movements). Furthermore, four major muscles were included as EMG parameters to simplify the model complexity and measurement sensors: one agonist and one antagonist involved in ankle movement and knee movement. Regarding AJM, even though only 3–4 parameters were selected for both half squat and gait, the degree of adequacy of the model was high ($R_f^2 = 0.92$, 0.97, respectively). KJM during half squat was estimated with high accuracy only from the shank angle and thigh angular acceleration ($R_f^2 = 0.98$). However, KJM during gait was less accurate compared with other parameters, regardless of five parameters that wereincluded ($R_f^2 = 0.63$). Our results showed that TA and GAS activities were most reflected in estimating AJM during half squat and gait, respectively. However, BFL and RF activities were not directly reflected in measuring KJM during both half squat and gait.

Regarding AJM during half squat, TA activity was the most relevant factor, followed by shank angle, GAS activity, and foot angular

Table 1
Differences due to the speed of each parameter at the peak point.

		Half squat			Gait		
		Self-selected	Fast	*p*	Self-selected	Fast	*p*
Muscle activity (%MVIC)	BFL	4.6 (2.9)	5.9 (3.6)	**0.004**	3.4 (5.0)	3.8 (5.4)	0.79
	RF	29 (16)	39 (25)	**< 0.001**	3.2 (3.4)	6.8 (18)	0.29
	GAS	7.0 (7.6)	9.0 (9.2)	0.11	21 (13)	25 (12)	0.18
	TA	31 (23)	31 (24)	0.83	7.9 (9.1)	8.8 (6.1)	0.65
Lower limb angle (deg)	Thigh	43 (11)	43 (12)	0.87	−4.6 (6.9)	−4.2 (9.8)	0.89
	Shank	−26 (9.3)	−26 (8.3)	0.90	−12 (6.3)	−11 (8.5)	0.95
	Foot	−5.8 (2.1)	−5.8 (2.2)	0.99	−5.8 (6.5)	−5.8 (5.9)	0.76
Lower limb angular acceleration (deg/sec^2)	Thigh	−215 (118)	−478 (283)	**< 0.001**	168 (213)	429 (400)	**0.003**
	Shank	143 (118)	285 (263)	**< 0.001**	−336 (190)	−627 (357)	**< 0.001**
	Foot	29 (37)	53 (82)	**0.013**	−234 (327)	−323 (345)	0.33
Joint moment (%BW × Ht)	AJM	0.04 (0.13)	0.07 (0.16)	0.16	0.58 (0.14)	0.62 (0.14)	0.32
	KJM	−0.43 (0.12)	−0.5 (0.16)	**< 0.001**	0.07 (0.15)	0.07 (0.13)	0.94

Data are presented as mean (SD) at the peak point of each parameter. Bold characters in the column of *p* indicate significant speed differences. MVIC: maximal voluntary isometric contraction, BFL: biceps femoris long head, RF: rectus femoris, GAS: gastrocnemius, TA: tibialis anterior, AJM: ankle-joint moment, KJM: knee-joint moment, BW: body weight, Ht: height.

Table 2
Joint moments prediction at half squat.

AJM			KJM		
	standard β	*p*		standard β	*p*
GAS	0.48	<0.001	Shank angle	0.83	<0.001
TA	1.1	<0.001	Thigh angular acceleration	0.18	<0.001
Shank angle	0.96	<0.001			
Foot angular acceleration	0.44	<0.001			
R_f^2: 0.92			R_f^2: 0.98		

R_f^2: coefficient of determination adjusted for degrees of freedom, GAS: gastrocnemius, TA: tibialis anterior, AJM: ankle-joint moment, KJM: knee-joint moment.

Table 3
Joint moments prediction at gait.

AJM			KJM		
	standard β	*p*		standard β	*p*
GAS	0.65	<0.001	GAS	0.36	0.01
Thigh angle	−0.54	<0.001	TA	−0.36	0.02
Thigh angular acceleration	−0.21	0.002	Shank angle	−0.85	0.003
			Thigh angular acceleration	−0.82	<0.001
			Shank angular acceleration	−0.72	0.001
R_f^2: 0.97			R_f^2: 0.63		

R_f^2: coefficient of determination adjusted for degrees of freedom, GAS: gastrocnemius, TA: tibialis anterior, AJM: ankle-joint moment, KJM: knee-joint moment.

acceleration, which supports our hypothesis. Given that the shank angle had a positive coefficient factor, the larger the shank angle was, the greater the AJM generated in dorsiflexion should be (Table 2, Fig. 2). However, the actual AJM remained almost constant toward plantarflexion, and little foot rotation was observed during half squat (Fig. 2). As shown in previous studies [6,12], our results showed that TA and GAS activities worked to fixate the foot during the squat. Since GAS functions isometrically as biarticular muscles of the knee joint [12], little muscle activity was seen in our results (Fig. 2). Hence, TA activity acting on foot fixation rather than foot rotation was the most influential on AJM during half squat.

Contrary to our hypothesis, muscle activities were not included, and only shank angle and thigh angular acceleration were extracted as relevant factors of KJM during half squat. Although the shank angle did not change significantly, even when the speed changed, the peak KJM and thigh angular acceleration increased at the higher speed (Table 1). Thus, once the shank reaches a certain angle, larger thigh angular acceleration allows greater KJM to occur. A previous study suggests that BFL and RF are required to resist gravity-induced acceleration toward flexion [6]. Furthermore, RF activity also acts to decelerate thigh rotation and then changes movement direction to extension [6]. Therefore, although BFL and RF activities were not reflected directly in the KJM, it is suggested that these play important roles in coordinating rotational movements of the lower limbs during squats.

Regarding the AJM during gait, our results support the hypothesis that GAS activity was more influential for AJM than thigh angular information. GAS activity had a positive coefficient, and thigh angle information had a negative coefficient (Table 3). Thus, increased GAS activity and/or larger and faster thigh movement contributed to greater AJM toward plantar flexion (Fig. 3). A plantar flexion moment is a required force at the end of the stance phase for the forward progression of the body [28]. GAS accelerates foot rotation backward to offset the forward momentum generated by gravity and provides a forward progression of the body in the late stance of gait [11,29]. Furthermore, hip flexion compensates for GAS activity in producing forward propulsion [30]. Thus, the thigh angular information was included in the relevant factors of AJM instead of the hip flexion strategy.

Contrary to our hypothesis, the shank angle had a stronger effect on KJM than did muscle activities. However, the degree of adequacy of the regression models for KJM during gait was lower than for others, and BFL and RF activities did not reflect the influential factors of KJM. Although TA activity was selected as one of the related factors, TA primarily works to hold the foot in dorsiflexion during the swing phase and fix it for preparing heel contact [29]. Despite a previous study showing that BFL and RF were activated to slow down the forward acceleration and stabilize the limbs at late stance [11], BFL and RF showed little activity in the participants in this study (Fig. 3). Beyaert et al. showed that the knee joint passively extends as the ground reaction force moves anteriorly during midstance [31]. In addition, KJM during the swing phase resulted from energy propagation owing to inertia when moving the body forward [28]. Thus, it is suggested that KJM during gait is not due to the muscle activities but rather ground reaction force and forward inertia.

Regarding the clinical contributions of this research, the proposed method could not only enable the estimation of joint moments based on a small number of sensors and parameters in clinical settings without large-scale laboratory equipment, but also may provide comprehensive insight into muscle activity, limb movement, and joint moment. For instance, since GAS activity and thigh angular information contributed to estimating AJM during gait (Table 3), insufficient GAS activity and/or less thigh swing during gait may lead to weak forward propulsion [11]. Decreases in forward propulsion during gait cause limited gait speed and stride length, which can cause falls in older people [32] or those with stroke [30]. Although therapeutic interventions that improve GAS activity and/or thigh swing may be desirable for fall prevention during gait, further research is required to determine whether this model could be applied to older adults and those with stroke. On the other hand, muscle activities involved in the knee joint movement were not reflected in estimating KJM during gait in the participants (Table 3). However, excessive muscle activities during gait have been observed in individuals with knee osteoarthritis, and may be involved in reducing gait speed and increasing energy consumption or progression of joint diseases [14,15]. Further research is necessary for examining how the increased muscle activities seen in people with knee osteoarthritis could affect limb movements and joint moments.

The present study has several limitations. First, the participants were limited to young, healthy adults. Joint moments differ between healthy individuals and those with orthopedic diseases [15,31] and differ even among the healthy participants depending on age [32], sex, and body shape [33]. Thus, further studies are necessary for clarifying the differences between the patterns of healthy adults and those with various types and severities of diseases while also considering the effects of age or body type. Second, we examined only the AJM and KJM in the sagittal plane. A more comprehensive evaluation is desirable, including the hip joint and trunk movements. However, this was a foundational study; hence, a simplified model was used. Lastly, verification motions were limited to two simple motions, half squat and gait. Verification with more advanced and comprehensive motions like running and jumping should be considered in the future for application to sports.

5. Conclusions

Regression models of the joint moments showed high accuracies, except for KJM during gait, even though only a maximum of five parameters were selected from muscle activities and limb angular information. Our findings showed that AJM throughout the half squat was influenced by TA activity acting to stabilize the foot. On the other hand, GAS activity and thigh swing were involved in the generation of AJM-producing forward propulsion during gait. Regarding KJM, muscle activities did not directly reflect both half squat and gait. This needs exploring the involvement of other conditions besides young healthy adults such as various ages or orthopedic and neurological diseases. Muscle activities work to regulate the rotational movement of limbs, and excessive or insufficient activities affect movement performances or efficacies. Considering the interrelationships among limb movement, muscle activity and joint moment comprehensively could enable clinicians to analyze human movement even based on a small number of parameters and may provide further insights into therapeutic interventions in that joint moments can be adjusted based on muscle activity and limb movement.

Data availability statement

Data will be made available on request.

Ethics Declarations

- This study was reviewed and approved by the Ethics Committee of the Graduate School of Medical Facilities, Nagasaki University, Nagasaki, Japan [approval no. 18061429].
- All participants provided informed consent for participation in the study.

CRediT authorship contribution statement

Umi Matsumura: Writing – review & editing, Writing – original draft, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Toshiya Tsurusaki:** Writing – review & editing, Software, Resources, Project administration, Methodology, Investigation, Formal analysis, Conceptualization. **Rena Ogusu:** Writing – original draft, Investigation, Data curation. **Shimpei Yamamoto:** Writing – original draft, Investigation, Data curation. **Yeonghee Lee:** Writing – original draft, Investigation, Data curation. **Shinya Sunagawa:** Writing – original draft, Investigation, Data curation. **W Darlene Reid:** Writing – review & editing, Validation. **Hironobu Koseki:** Writing – review & editing, Supervision, Conceptualization.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests:

Umi Matsumura reports writing assistance was provided by editage.

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