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ORIGINAL ARTICLE

Effect of different knee flexion angles with a constant hip and knee torque on the muscle forces and neuromuscular activities of hamstrings and gluteus maximus muscles

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Abstract

Purpose This study examined the effect of different knee flexion angles with a constant hip and knee torque on the muscle force and neuromuscular activity of the hamstrings and gluteus maximus.

Methods Twenty healthy males lay in prone position and held their lower limb with hip flexion at 45° and knee flexion at either 10° or 80°. At these angles, the hip and knee torques are identical. Under three load conditions: passive (referred to as *Unloaded*), active (*Loaded*), and active with 3-kg weight added to the shank (*Loaded + 3 kg*), the muscle stiffness (i.e., an indicator of muscle force) and neuromuscular activity of the hamstrings and gluteus maximus were measured using shear wave elastography and surface electromyography.

Results The muscle stiffness and neuromuscular activity of the hamstrings and gluteus maximus increased significantly with the load. Muscle stiffness in the hamstrings was significantly lower at knee flexion of 80° than at 10° for *Unloaded*, but not for either *Loaded* or *Loaded + 3 kg*. The neuromuscular activity of the hamstrings was significantly greater at knee flexion of 80° than at 10° for both *Loaded* and *Loaded + 3 kg*. The muscle stiffness or neuromuscular activity of the gluteus maximus showed no significant differences between knee angles.

Conclusions When the passive force in the hamstrings decreases with knee flexion, sufficient muscle force to maintain the hip and knee torques against an external load is generated by preferentially increasing the neuromuscular activity of the hamstrings, rather than increasing the synergetic muscle force.

Keywords Muscle force · Neuromuscular activity · Shear wave elastography · Electromyography · Hamstrings · Gluteus maximus

Abbreviations

EMG Electromyography
MVC Maximal voluntary contraction
SWE Shear wave elastography

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Introduction

The muscle force is the total of the passive force caused by the stretching of the skeletal muscles and the active force caused by the muscle contraction in response to the neuromuscular activity (Morgan and Allen 1999). The muscle strength manifests as the joint torque obtained by multiplying the total force by the perpendicular distance from the center of the joint to the line where the muscle force is exerted (Rassier et al. 1999). Therefore, greater neuromuscular activity (i.e., active force) may be required at short muscle lengths than long muscle lengths to exert equivalent muscle strengths because less passive force is expected at short muscle length than long muscle lengths. If the muscle is working on the descending limb of the force–length relationship, the shorter muscle lengths may have a greater total force generating potential. However, Medda et al. (1997) reported that a greater neuromuscular activity (EMG amplitude) is necessary for generating a greater active force. Therefore, to produce the same total force in the short muscle length as in the long muscle length, the neuromuscular activity should increase to compensate for a reduction in the passive force even if the muscle is working between any regions of the force–length relationship. Tateuchi et al. (2015) reported the changes in the neuromuscular activities of the hip abductor muscles when the hip joint changed from a neutral position to adduction or abduction during one-legged standing. As a result, the external hip adduction moments (i.e., the hip abduction moments to be exerted) at hip adduction and abduction were almost the same as those at the neutral hip position. Nevertheless, the neuromuscular activities of the hip abductor muscles decreased at hip adduction because of the abductor having long muscle length and increased at hip abduction owing to short muscle length. This suggested that the change of passive force in the soft tissue influenced the active adjustment of neuromuscular activity.

A study by Kwon and Lee (2013) examined the maximum hip extension strength for a range of knee flexion angles at the neutral hip position and reported that the maximum hip extension strength and the neuromuscular activity of the hamstrings decreased, whereas the neuromuscular activity of the gluteus maximus increased with the knee flexion angle. Considering the study of Chleboun et al. (2001) that at the neutral hip position the hamstrings are located in the ascending limb of the force–length relationship, the exerted torque may decrease with an increase in the knee flexion angle upon exhibiting the maximum hip extension strength. This may lead to a decrease in the neuromuscular activity of hamstrings. On the other hand, the results of the examination of the gluteus maximus suggested that its neuromuscular activity increased as a compensation for the decrease in the neuromuscular activity of the hamstrings because the gluteus maximus was neither stretched nor shortened due to the knee flexion. Therefore, when exerting the same muscle strength but at different joint angles, the neuromuscular activity may increase not only in the muscle for which the passive force decreases, but also in the corresponding synergetic muscle. Since there was no way to measure the muscle force during the exercise, it was not possible to determine how the force in a muscle and its corresponding synergetic muscle was controlled by the neuromuscular activity at different joint angles (i.e., different muscle lengths).

Moritani and deVries (1978) reported that highly significant correlations were indicated between the isometric muscle contraction force and the neuromuscular activity measured by the surface electromyography (EMG). Therefore, the neuromuscular activity as measured by the EMG has been applied as an indicator of how much and when the muscles exerted force during the exercise (Ekstrom et al. 2007; Pranata et al. 2017). On the other hand, Lunnen et al. (1981) reported that the neuromuscular activity of the biceps femoris further decreased at higher hip flexion angles (i.e., longer hamstrings lengths) when a constant submaximal knee flexion torque developed at a constant knee flexion angle with various hip extension angles. If it is not experimental (e.g., exercise), however, the external joint torque often changes when the joint angle is changed. Consequently, it was difficult to estimate the muscle force at different joint angles from the neuromuscular activity because, it was not possible to determine whether any change in the neuromuscular activity was derived from the change in the muscle length or the change in the exercise load. Therefore, to estimate the muscle force during various exercises, an evaluation method other than the EMG is necessary.

In recent years, a non-invasive method of measuring the muscle stiffness as an index of the muscle force using an ultrasonic shear wave elastography (SWE) has been reported (Nordez and Hug 2010; Shinohara et al. 2010; Dubois et al. 2015). This method is a technology for a sensitive imaging of the tissue stiffness based on the velocity of the shear wave (Bercoff et al. 2004). Previous studies have reported that the muscle stiffness measured by this method exhibits a linear relationship with the individual muscle force exerted during the exercise, or with the passive force that results when the muscle is stretched by changing the joint angle at rest (Bouillard et al. 2011; Maïsetti et al. 2012; Le Sant et al. 2015). Therefore, the muscle stiffness measured by the SWE may be a useful indicator of the total and passive forces in an individual muscle. The present study decided to use the muscle stiffness measured using the SWE as an index of the muscle force.

The goal of the present study was to clarify the effect of different knee flexion angles on the muscle force and neuromuscular activity of the hamstrings and gluteus maximus using the SWE and EMG, while exerting a constant hip and knee joint torque during an isometric hip extension in a prone position. It was hypothesized that, as the knee flexion angle would increase, the passive force in the hamstrings would decrease while the neuromuscular activity would increase in both the hamstrings and the synergic gluteus maximus muscle.

Methods

Participants

Twenty healthy adult males (age 23.0 ± 2.4 years; height 171.8 ± 5.5 cm; weight 63.1 ± 7.4 kg) with no history of neurological or orthopedic diseases in the right lower limb volunteered for this study. A power analysis with an α error = 0.05, power = 0.95, and effect size = 0.48, based on a previous study (Tateuchi et al. 2015), was performed using the G*Power 3.1 analysis software (Heinrich Hein University, Duesseldorf, Germany). This produced a minimum total sample size of 14. The participants provided their informed consent after they had been given an explanation of the objectives of the experiment and the risks involved. The present study was conducted in compliance with the Declaration of Helsinki and was approved by the ethics committee of Kyoto University Graduate School and the Faculty of Medicine (R0520).

Experimental protocol

The experiment was conducted by four examiners, all of whom were physiotherapists. One examiner measured the joint angles and manipulated the ultrasonic equipment in all experiments, while one of the other three held the participants' lower limbs in the rest position.

Setting of task posture

To assess the effects of changes in the muscle length (due to variations in the joint angle) on the neuromuscular activity, as measured at different knee angles, it is necessary to eliminate the influence of the exercise load at the different knee angles. Therefore, those knee angles at which the same external hip and knee joint torques were generated were selected as the task postures. The joint torque at a non-weight-bearing position is expressed as the product of the weight distal to the joint and the perpendicular distance from the center of the joint to the line of gravity distal to the joint. As shown in Fig. 1, the perpendicular distances from the hip and knee joints to the line of gravity of the shank and foot were the same for both the knee angles used as the task postures, that is, the knee flexion at 10° and 80° and the hip flexion at 45°. As such, the same external hip and knee torques were generated in both the task postures. To obtain the measurements, the participants were laid in a prone position on a bed with their right lower limb hanging from the bed, while their pelvis was securely immobilized with a belt to prevent the pelvis from inclining backward. For every measurement, the same examiner set the joint angle using a goniometer and also set the task posture.

Muscle stiffness for unloaded condition

In the task posture, the participants were instructed to place their right thigh on a wooden board, which had been leant against the bed. The measurement was performed at rest while their right shank was held by the examiner (*Unloaded condition*, Fig. 2a). The muscle stiffness for the *Unloaded* condition at each knee angle was measured three times by the same examiner for every participant. The EMG was used to confirm that there was no neuromuscular activity.

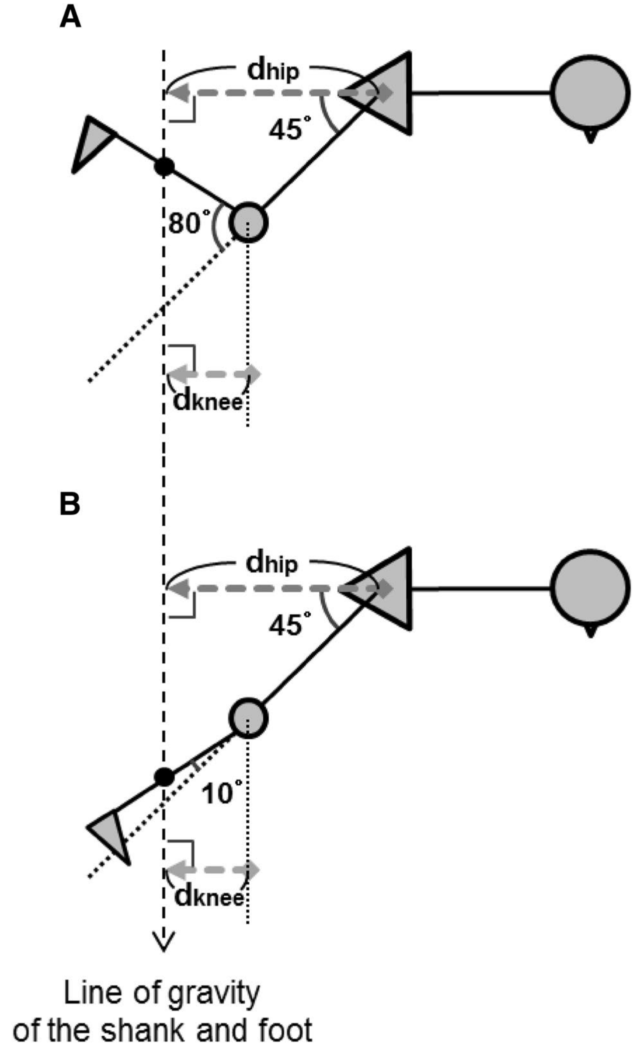

Fig. 1 Schematic representation of the task posture. When the perpendicular distances from the hip and the knee joints to the line of gravity of the shank and the foot (d_{hip} and d_{knee}, respectively) become similar at the respective knee flexion of 80° (**a**) and 10° (**b**), the respective external hip and knee torques at both the angles also become similar

Neuromuscular activity and muscle stiffness for *Loaded* and *Loaded + 3 kg* conditions

In the task posture, the participants maintained the determined joint angles themselves (*Loaded* condition, Fig. 2b), and subsequently maintained the position with a 3-kg weight attached to the shank (*Loaded + 3 kg* condition, Fig. 2c). For each load, the examiner who measured the muscle stiffness for the *Unloaded* condition also obtained three measurements of the neuromuscular activity and the muscle stiffness for the *Loaded* and *Loaded + 3 kg* conditions.

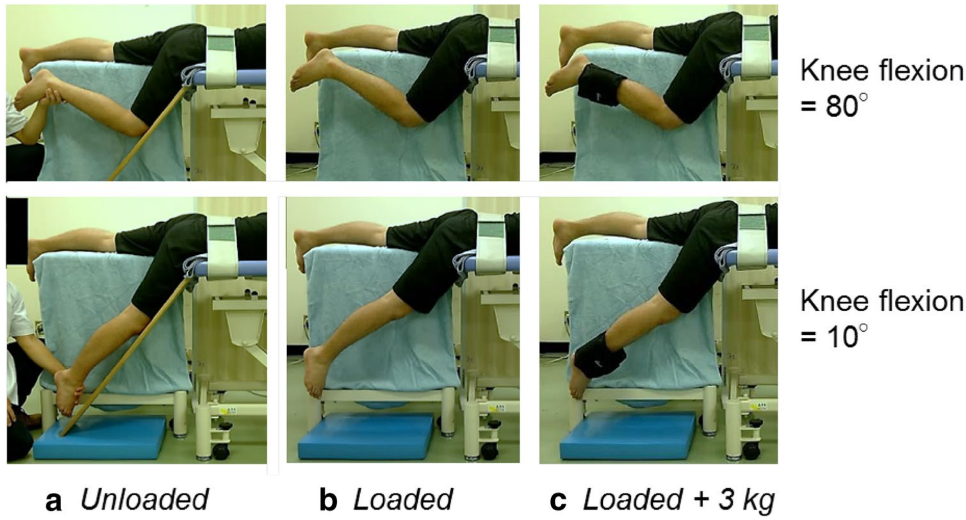

Fig. 2 Measurement position for each loaded condition. **a** *Unloaded* condition: the participant's thigh is placed on a wooden board leant against the bed, while their shank is held at rest by an examiner; **b** *Loaded* condition and **c** *Loaded + 3 kg* condition: The determined joint angle is maintained by the participant without any load and with a 3-kg weight added to the shank

Shear wave elastography

In the present study, the muscle stiffness (i.e., muscle shear elastic modulus) as measured by SWE (Aixplorer, SuperSonic Imagine, France) was used as an index of the muscle force (Bouillard et al. 2011; Maïsetti et al. 2012; Le Sant et al. 2015). The muscle stiffness (μ) is automatically calculated by SWE using the following formula (Bercoff et al. 2004; Gennisson et al. 2010):

$$\mu \ \text{(kPa)} = \rho Vs^2,$$

where ρ is the muscle tissue density (1000 kg/m^3), and Vs is the propagation velocity of the shear wave generated by the ultrasonic transducer. The ultrasonic transducer was an SL 15-4 transducer (4–15-MHz linear array probe, SuperSonic Imagine, France) that was adjusted to be parallel to the muscle fibers. The target muscles were the right biceps femoris, semitendinosus, upper gluteus maximus, and lower gluteus maximus. The measurement sites for each muscle (listed in Table 1a) were based on the methods of the previous studies and the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) recommendations (Lyons et al. 1983; Fujisawa et al. 2014; http://www.seniam.org/), and was determined by palpation and visual inspection. After checking the muscle belly of each muscle with the ultrasonic device, the muscle stiffness in a Ø4-mm circle at the center of the region of interest near the center of each muscle was measured. The stiffness of each muscle measured for the *Unloaded* condition corresponded to "the passive force", while that for *Loaded* and *Loaded + 3 kg* conditions corresponded to "the total force" (Bouillard et al. 2011; Koo et al. 2013). The reliability of the measurements of the muscle stiffness was assessed using an intra-class correlation coefficient (ICC) calculated by the values of all the three measurements for each condition. The ICC (1,1) was considered as "almost perfect" within a range of 0.89–1.00 for all the muscles (Table 2). An average of the three measurements of the muscle stiffness under each load condition was calculated. In each muscle, the change in the muscle stiffness from the *Unloaded* condition (i.e., passive force)

Table 1 Muscle stiffness and neuromuscular activity measurement sites

(a) Muscle stiffness	
Bicep femoris	Slightly to the tibial side at the midpoint between the ischial tuberosity and the lateral condyle of the tibia
Semitendinosus	Slightly to the tibial side at the midpoint between the ischial tuberosity and the medial condyle of the tibia
Upper gluteus maximus	Slightly to the greater trochanter side at the midpoint between the posterior superior iliac spine and the greater trochanter
Lower gluteus maximus	6-cm inferior medial, relative to the upper-fiber measurement site
(b) Neuromuscular activity	
Biceps femoris	Slightly to the ischial side at the midpoint between the ischial tuberosity and the lateral condyle of the tibia
Semitendinosus	Slightly to the ischial side at the midpoint between the ischial tuberosity and the medial condyle of the tibia
Upper gluteus maximus	Slightly to the posterior superior iliac spine side at the midpoint between the posterior superior iliac spine and the greater trochanter
Lower gluteus maximus	6-cm inferior medial, relative to the upper-fiber measurement site

Table 2 Reliability of muscle stiffness

Muscle	Measurement condition		$ICC_{1,1}$	95% CI
Biceps femoris	*Unloaded*	80°	1.00	1.00–1.00
		10°	0.99	0.98–0.99
	Loaded	80°	0.99	0.98–1.00
		10°	0.99	0.97–0.99
	Loaded + 3 kg	80°	0.98	0.96–0.99
		10°	0.98	0.97–0.99
Semitendinosus	*Unloaded*	80°	1.00	1.00–1.00
		10°	0.99	0.98–0.99
	Loaded	80°	0.98	0.96–0.99
		10°	0.95	0.91–0.98
	Loaded + 3 kg	80°	0.98	0.96–0.99
		10°	0.96	0.91–0.98
Upper gluteus maximus	*Unloaded*	80°	0.97	0.93–0.99
		10°	0.94	0.88–0.97
	Loaded	80°	0.98	0.96–0.99
		10°	0.99	0.97–0.99
	Loaded + 3 kg	80°	0.98	0.96–0.99
		10°	0.99	0.97–0.99
Lower gluteus maximus	*Unloaded*	80°	0.94	0.89–0.97
		10°	0.89	0.78–0.95
	Loaded	80°	1.00	0.99–1.00
		10°	0.98	0.96–0.99
	Loaded + 3 kg	80°	0.99	0.99–1.00
		10°	0.99	0.99–1.00

ICC intra-class coefficient correlation, *CI* confidence interval

to the *Loaded* and the *Loaded + 3 kg* conditions (i.e., total force) was calculated as the value of "the active force" in the *Loaded* and the *Loaded + 3 kg* conditions. Furthermore, an average value for the biceps femoris and the semitendinosus was calculated as the value for the hamstrings; similarly, an average value for the upper and lower gluteus maximus was calculated as the value for the gluteus maximus.

Electromyography

Using the same method as that used to measure the muscle stiffness, the neuromuscular activity was measured for the right biceps femoris, semitendinosus, upper gluteus maximus, and lower gluteus maximus. The measurement sites for each muscle (listed in Table 1b) were determined by referring to the previous studies and the SENIAM recommendations (Lyons et al. 1983; Fujisawa et al. 2014; http://www.seniam.org/), and by confirming the contraction of each muscle by palpation and visual inspection. The skin at each measurement site was cleaned with ethanol. Bipolar-surface EMG electrodes (Telemyo DTS (1500 Hz), Noraxon USA Inc., USA) with an interelectrode distance of 20 mm (center-to-center) were attached, after which the measurement was performed for each muscle. The neuromuscular activity of each muscle was analyzed using the value normalized to that at the maximum isometric contraction in the position used for the manual muscle test (%MVC). Furthermore, in the same way as for the muscle stiffness measurement, the average for the biceps femoris and semitendinosus was calculated as the value for the hamstrings, while the average for the upper and lower gluteus maximus was calculated as the value for the gluteus maximus.

Statistical analysis

Statistical analysis was performed using a SPSS software (SPSS Statistics version 22.0, SPSS Japan Inc., Japan). To determine the muscle stiffness for all the 6 tasks, and the change in the muscle stiffness and the neuromuscular activity for 4 tasks (excluding that for the *Unloaded* condition), a two-way analysis of variance (ANOVA) with repeated measurements was performed using the two factors of the knee angle (a knee flexion of 10° and 80°) and the load (*Unloaded, Loaded*, and *Loaded + 3 kg*). If a significant interaction or effect was observed, multiple comparisons using the Bonferroni post hoc test were performed. The statistical significance was determined to be $p < 0.05$.

Results

The results of all the measurements of the muscle stiffness and neuromuscular activity of the hamstrings are listed in Table 3. In the two-way ANOVA results, the muscle stiffness indicated a significant interaction between the knee angle and the load, and increased significantly with an increase in the load at the knee flexions of 10° and 80°, respectively. The muscle stiffness for the *Unloaded* condition was significantly lower at the knee flexion of 80° than that at 10°, but there were no significant differences between the muscle stiffness of the knee flexion angles for either *Loaded* or *Loaded + 3 kg* condition. The change in the muscle stiffness from the *Unloaded* condition to the *Loaded* and *Loaded + 3 kg* conditions indicated significant main effects of both the knee angle and the load. The change in the muscle stiffness was significantly higher for the *Loaded + 3 kg* condition than that for the *Loaded* condition regardless of the knee angle, and was significantly higher at the knee flexion of 80° than that at 10° regardless of the load. The neuromuscular activity indicated a significant interaction between the knee angle and the load. The neuromuscular activity was significantly higher for the *Loaded + 3 kg* condition than that for the *Loaded* condition at both the knee flexion angles, and was significantly higher at the knee flexion of 80° than that at 10° for both the *Loaded* and *Loaded + 3 kg* conditions.

Table 3 Muscle stiffness and neuromuscular activity of hamstrings

Load condition	*Unloaded*		*Loaded*		*Loaded + 3 kg*	
Angle of knee flexion	10°	80°	10°	80°	10°	80°
Muscle stiffness (kPa)	20.3 (7.5)	7.7* (2.3)	35.7† (12.6)	36.4† (12.9)	50.5†,‡ (14.5)	53.5†,‡ (17.9)
Change in muscle stiffness (kPa)	–	–	15.4 (10.2)	28.7* (13.6)	30.2‡ (17.7)	45.8‡,* (19.0)
Neuromuscular activity (%MVC)	–	–	5.7 (5.5)	13.5* (8.8)	12.2‡ (7.9)	22.6‡,* (11.8)

Values are expressed as the means (SD)

MVC maximum voluntary contraction

†$p < 0.05$ vs. *Unloaded*; ‡$p < 0.05$ vs. *Loaded*; *$p < 0.05$ vs. 10°

The results of all the measurements of the muscle stiffness and neuromuscular activity for the gluteus maximus are listed in Table 4. The results of the two-way ANOVA revealed that the muscle stiffness exhibited no significant interaction and indicated significant main effect of the load condition only. The muscle stiffness for the *Loaded* and *Loaded + 3 kg* conditions was significantly higher than the muscle stiffness for the *Unloaded* condition, but there were no significant differences between the respective muscle stiffness for both the *Loaded* and *Loaded + 3 kg* conditions, regardless of the knee angle. The changes in the muscle stiffness for the *Loaded* and *Loaded + 3 kg* conditions and the neuromuscular activity also indicated a significant main effect of the load condition only, and it was significantly higher for the *Loaded + 3 kg* condition than that for the *Loaded* condition, regardless of the knee angle.

Discussion

The present study examined the effect of different knee flexion angles with a constant hip and knee joint torque on the muscle forces and neuromuscular activities of hamstrings and gluteus maximus muscles during the isometric hip extension in the prone position using the SWE and conventional EMG measurement. To the best of our knowledge, this is the first study revealing the effect of the changes in the hamstring muscle length (due to variations in the knee flexion angle) on the muscle forces and neuromuscular activities of hamstrings and gluteus maximus muscles. The results showed that both the muscle stiffness (i.e., muscle force) and neuromuscular activity of the gluteus maximus tended to increase with the load, but there was no significant difference between knee flexions of 10° and 80°. On the other hand, both the muscle stiffness (i.e., muscle force) and neuromuscular activity of the hamstrings increased significantly with the load. The muscle stiffness for the *Unloaded* condition (i.e., passive force) in the hamstrings decreased significantly, while the neuromuscular activity increased significantly at knee flexion of 80° than 10°. The muscle stiffness for the *Loaded* and *Loaded + 3 kg* conditions (i.e., total force) in the hamstrings showed no significant difference between the knee angles. It was assumed that, when the length of the hamstring muscles decreased with knee flexion, an increase in the neuromuscular activity of the hamstrings may occur preferentially rather than a compensatory increase in the muscle force of the gluteus maximus, such that the position was maintained regardless of the joint angle.

The result of the present study, which differed from the hypothesis, showed that the neuromuscular activity of the hamstrings increased but that of the gluteus maximus did not increase when, the muscle stiffness for the *Unloaded* condition (i.e., passive force) in the hamstrings decreased as the knee flexion angle increased. The stiffness in the hamstring muscles showed no significant differences between the knee angles for both *Loaded* and *Loaded + 3 kg* conditions. This may indicate that, in the hamstrings, sufficient total force to support an external load is exerted by increasing the neuromuscular activity to compensate for the decrease in the passive force caused by an increase in the knee flexion

Table 4 Muscle stiffness and neuromuscular activity of gluteus maximus

Load condition	*Unloaded*		*Loaded*		*Loaded + 3 kg*	
Angle of knee flexion	10°	80°	10°	80°	10°	80°
Muscle stiffness (kPa)	5.4 (1.2)	5.6 (1.3)	11.0† (7.0)	12.4† (6.4)	14.4† (9.2)	16.0† (10.5)
Change in muscle stiffness (kPa)	–	–	5.6 (6.9)	6.8 (6.3)	9.1‡ (9.3)	10.4‡ (10.4)
Neuromuscular activity (%MVC)	–	–	3.4 (2.1)	3.2 (1.5)	4.7‡ (3.7)	5.4‡ (3.2)

Values are expressed as the means (SD)

MVC maximum voluntary contraction

†$p < 0.05$ vs. *Unloaded*; ‡$p < 0.05$ vs. *Loaded*

angle. Therefore, the neuromuscular activity of the gluteus maximus may not be increased because it is not necessary to increase the total force in the gluteus maximus to maintain hip extension. These results may indicate that an increase in the neuromuscular activity of the hamstrings precedes an increase in the muscle force in the synergetic gluteus maximus when the passive force in the hamstrings decreases. On the other hand, a simulation study using a musculoskeletal model (Lewis et al. 2009) showed that if the total force in the hamstrings during prone hip extension is reduced by half, the total force in the gluteus maximus increases to compensate. It is assumed that this compensation by the gluteus maximus is very likely to occur in persons with osteoarthritis of the knee or in those who have undergone total knee arthroplasty, for whom the maximum muscle strength of the hamstrings has been reported to have been reduced by 30 to 50% relative to a healthy person (Alnahdi et al. 2012; Schache et al. 2014). Further studies using persons with hip or knee joint diseases are needed to clarify how the muscle force is controlled, as well as the relationship between muscles.

No significant difference in the muscle stiffness was observed between the *Loaded* and *Loaded + 3 kg* conditions in terms of the total force of the gluteus maximus, although an increasing trend was apparent ($p = 0.074$). This may be because, the change in the load was small, although the load of 3 kg used in this study constituted 10–15% of the maximum hip extension strength (Worrell et al. 2001; Kwon and Lee 2013) or 15–20% of the maximum knee flexion strength (Worrell et al. 2001; Mohamed et al. 2002). However, the prior anthropometric study reported that the average weight of the shank and foot of young adult males was 5.74% of the body weight (Park et al. 1999). Therefore, in this study, the weight of the shank and foot and the torque applied to the knee joint under the *Loaded + 3 kg* condition was about 183% and about 220% of the Loaded condition, respectively. This indicated clearly that the load for the *Loaded + 3 kg* condition differed from the *Loaded* condition. While there was no significant difference in the total force of the gluteus maximus between the *Loaded* and *Loaded + 3 kg* conditions, the active force (i.e., the change in the muscle stiffness) obtained by subtracting the passive force from the total force indicated a significant difference between the *Loaded* and *Loaded + 3 kg* conditions as well as in the neuromuscular activity. Upon applying a higher load, a sufficient total force to maintain the knee flexion and hip extension might not have been produced by an increase in the neuromuscular activity of the hamstrings alone, necessitating a more clearly increase in the total force of the gluteus maximus as well. However, the present study suggested the active force to be an indicator of the load differences more likely than the total force. Although, the compensation by the gluteus maximus did not occur due to the preferential increase in the neuromuscular activity of the hamstrings for the *Loaded + 3 kg* condition, as used in this study. Further research using higher loads will be necessary to gain a better understanding of the relationship between the muscles.

In the present study, we evaluated not only the neuromuscular activity as measured by conventional EMG but also the muscle force (which could not be evaluated in vivo before), using SWE to measure the muscle stiffness, and we investigated the muscle activity as the joint angle changed during exercise. The results of the present study suggested that the change in the neuromuscular activity that occurs with a change in the joint angle did not simply reflect the change in the total force exerted by the muscle, but was produced to compensate for the change in the passive force and to produce a total force that was sufficient to support an external load. Therefore, the simultaneous measurement of the muscle force and neuromuscular activity at different joint angles during exercise would allow us to clarify not only the differences in the active neurological reaction in the muscle but also the difference in the muscle activity with changes in the joint angle or in the exercise load. As both the evaluation methods used in the present study were non-invasive, they could be applied to the investigate the variation in the passive or active force resulting from a change in the posture or due to disease (Tateuchi et al. 2016; Botanlioglu et al. 2013; Du et al. 2016). These methods are expected to find a wide range of applications in the future.

There were some limitations to this study. First, as mentioned above, the neuromuscular activity of the hamstring muscles was lower than 22% MVC; different results may have been obtained under higher loads in which the muscle activity is higher. Second, it is unclear whether the moment arm of the hamstring muscles is the same for each of the knee angles used in the present study. To clarify the influence of the muscle lengths with a change in the joint angle on the muscle force and neuromuscular activity, two different knee flexion angles (10° and 80°) were applied to produce a constant hip or knee joint torque. However, it is possible that the moment arm of the hamstrings differed depending on the knee angle. Buford et al. (1997) calculated the moment arm of the hamstrings of cadavers at various knee angles, applying the tendon excursion method, and reported that the moment arms of the biceps femoris and semitendinosus were about 20 mm and 35 mm, respectively, at a knee flexion of 10° and about 30 mm and 50 mm, respectively, at 80°. However, this research was not performed in vivo and did not consider the effect of the hip joint angle. A previous study investigating the moment arm using a video X-ray image reported that the moment arm of the biceps femoris was 25.4 mm at a knee flexion of 11–20° and 24.3 mm at 71–80° (Kellis and Baltzopoulos 1999). Therefore, it was thought that the moment arm for both angles considered in the present study would be almost the same. However, the influence of the moment arm of the hamstrings could have affected the results. Since the moment arm is an important factor affecting the manifestation of the muscle strength, further studies considering the muscle moment arm may be needed to verify the muscle force control. Third, the individual differences in the force–length relationship of the hamstrings might have influenced the results. Moltubakk et al. (2016) reported that the angle corresponding to the peak torque during isokinetic knee flexion changed to the more flexed knee angle in people who generated a larger passive torque when passively extending the knee joint. This suggested that the difference in the stiffness of the hamstrings' muscle tendon unit affects the activity of the hamstrings at a specific knee flexion angle, and may have a great influence on the relationship between the hamstrings and gluteus maximus. By focusing on the stiffness of the hamstrings, the relationship between the hamstrings and gluteus maximus can be clarified in more detail. Fourth, this study did not measure the neuromuscular activity and passive force of the antagonist muscles. Therefore, it cannot be denied that, the passive force of the quadriceps, which is the antagonist of the hamstrings, might have influenced the knee joint torque; and that, the neurological mechanism, such as reciprocal inhibition accompanying co-contraction of the quadriceps, might have influenced the active force of the hamstrings to certain extent. However, since the torque due to the weight of the shank and foot acts in the knee extension direction in the task posture used in this study, there might have been little influence by the neuromuscular activity of the quadriceps.

Conclusion

We measured the muscle forces and neuromuscular activities of hamstrings and gluteus maximus muscles during prone hip extension at different knee flexion angles and investigated how these were controlled to maintain the same hip and knee torques. In the hamstrings, the passive force was lower at a knee flexion of 80° than at 10° while the neuromuscular activity was higher at 80° than at 10°. Therefore, the total force showed no significant difference between the knee angles. As a result, the muscle force and neuromuscular activity of the gluteus maximus were constant, again regardless of the knee flexion angle. These results suggest that, when the passive force in the hamstrings decreases due to an increase in the knee flexion angle, an increase in the neuromuscular activity of the hamstrings may occur preferentially rather than a compensatory increase in the total force of the synergetic gluteus maximus muscle, to exert sufficient total force to maintain a constant the hip and knee torque.

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Author contributions YM, HT, SN, and NI conceived and designed the research. YM, IS, TK, and YK conducted the experiments. YM, HT, and NI analyzed the data. YM, HT, SN, and NI wrote the manuscript. All the authors have read and approved the manuscript.

Compliance with ethical standards

Conflict of interest The authors declare no conflict of interest.

Ethical approval All the procedures performed in the studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards.

Informed consent Informed consent was obtained from all individual participants involved in the study.

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