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Original Article

Contribution of hip and knee muscles to lateral knee stability during gait

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Abstract. [Purpose] Lateral knee instability is frequently observed in patients with knee injury or risk factors associated with knee osteoarthritis. Physical exercises can strengthen muscles that stabilize the knee joint. The purpose of this study was to define the contribution of the knee and hip muscles to lateral knee stability by comparing the muscle forces, as assessed by musculoskeletal simulation using one or two degrees-of-freedom (1-DOF and 2-DOF) knee models. [Participants and Methods] We evaluated the normal gait of 15 healthy subjects. We conducted a three-dimensional gait analysis using a motion analysis system and a force plate. We considered a muscle as a lateral knee stabilizer when the calculated muscle force was greater with the 2-DOF model than with the 1-DOF model. [Results] During early and late stance, the muscle forces of the lateral knee and hip joint increased in the 2-DOF model as opposed to in the 1-DOF model. In contrast, the forces of the medial knee muscles decreased. Furthermore, hip muscle forces increased during the late stance. [Conclusion] Our results show that the lateral knee and hip muscles contribute to lateral knee stability. Thus, exercises to strengthen these muscles could improve lateral knee stability.

Key words: Lateral knee stability, Musculoskeletal model, Gait analysis

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INTRODUCTION

During activities of daily living and during anti-gravity activities such as walking, the knee joint is exposed to repeated loads. During waking, knee joint stability is necessary to maintain the joint alignment against external knee adduction moment caused by the ground reaction force. Lack of stability in the knee joint on the frontal plane causes lateral thrust seen in the gait of medial knee osteoarthritis (OA) patients[1–3)]. Lateral knee instability is also one of the risk factors for progression of knee OA[2, 3)]. Therefore, lateral knee stability is an important factor to protect the knee joint from the musculoskeletal disorders.

The knee joint is stabilized laterally by bone structure, ligaments, joint capsule, and so on[4–6)]. Contributions of the muscles to the dynamic lateral stability of knee joint increases under the conditions of bone or ligament injury. In addition, muscle force could be enhanced by training; thus, the identification of muscles contributing to knee joint stability is crucial to rehabilitation of the knee joint with lateral instability.

Previous studies have reported that external lateral force to the shank increases the muscle activation around the knee joint in static standing leading to stabilization of the joint alignment[7, 8)]. However, knee joint angle, ground reaction force, and relative positional relationship between a muscle and a joint during walking are markedly different from that during static

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standing. Thus, the results of contributions of the muscles in static standing could not apply to that during walking, and the muscles associated with lateral stability of the knee joint during walking have not been defined yet. In addition to the knee muscles, previous studies report a reduction in hip abductor strength in knee OA patients[9, 10]. Notably, several studies report inconsistent results with respect to hip and knee moment during gait[11, 12], and the mechanism of hip muscle contribution in knee OA has not been yet clarified.

Musculoskeletal modeling and simulation have recently been used to estimate the muscle force. The degrees of freedom (DOFs) in the musculoskeletal model can be modified depending on the aim of the analysis. The musculoskeletal knee model with 1DOF, in which lateral knee stability is maintained mechanically, is used to analyze various physical activities[13–15]. In this model, muscle force is not required for lateral stabilization of the knee joint. Alternatively, the musculoskeletal model with 2DOF knee model while allowing frontal plane motion, requires muscle forces crossing the knee joint to stabilize it laterally. A previous study analyzed the effect of DOFs on the tension force of knee muscle during single-legged hopping, and reported that more DOFs in the knee joint lead to greater force and different muscle activation patterns to maintain the kinetic balance[16]. Therefore, we determined that it would be possible to define the muscles responsible for lateral stabilization of the knee joint by comparing muscle forces estimated using the 1 and 2DOF knee models.

The purpose of this study was to define the contribution of the knee and hip muscles to lateral knee stability by comparing muscle forces as estimated using musculoskeletal model simulation with the 1 and 2DOF knee models.

PARTICIPANTS AND METHODS

Fifteen healthy young males (age: 26.2 ± 4.6 years; height: 170.3 ± 5.1 cm; weight: 65.2 ± 10.0 kg) without any orthopedic or neurological disorders participated in this study. Each participant read and signed informed consent. This study was approved by the Ethics Committee of Kagoshima University Medical School (No. 155).

We calculated muscle forces using the AnyBody 6.0 (AnyBody Technology, Aalborg, DK) musculoskeletal model simulation software from the motion capture and the ground reaction force (GRF) data. The validity of muscle forces estimated using this musculoskeletal model simulation software has been confirmed in previous studies[17–19]. Participants walked on a 9-m walkway and stepped on the force plate with their right foot. To minimize the effect of gait velocity, participants walked at 100 steps/min using a metronome[20]. The subjects practiced gait several times prior to data collection.

Motion capture was performed using a 7-camera optoelectronic motion analysis system (VICON MX3, Oxford Metrics, Oxford, UK) combined with a force plate (9286A, Kistler, Jonsered, SE). The force plate was secured in the middle of the walkway to determine the GRF. Sampling frequencies of the infrared camera and the force plate were 100 Hz and 1,000 Hz, respectively.

Each participant wore 25 retro-reflective markers on the bony landmarks of the head, thorax, pelvis, and right lower extremities. Specifically, markers were placed on the temple and back of the head, the manubrium of the sternum, the xiphoid process, the tenth thoracic vertebra, both sides of the anterior superior iliac spine, the posterior superior iliac spine, the lateral aspect of the thigh, the lateral epicondyle, the lateral aspect of the shank, the lateral malleolus, the first and fifth heads of the metatarsal bone, and the heel. Marker locations were determined based on a plug-in-gait marker set[21, 22].

Marker trajectories and GRF data were filtered using a Butterworth low-pass filter at 5 Hz and 12 Hz cut-off frequencies, respectively. Muscle forces were calculated using the MocapModel in AMMR 1.6.4, which is the standard model available in the AnyBody software. This model includes 170 muscles in 6 segments (head, trunk, pelvis, right thigh, right shank, and right foot). We used a simple muscle model without force-length-velocity relationships as they have previously been shown to have little effect on the prediction of muscle forces and joint contact forces while walking[23]. Hip and ankle joints have 6- and 1-DOF, respectively. This model is scaled based on a standard model of the AnyBody software that uses height, body weight, and segment length. The 1-DOF knee model was a hinge joint, in which the lateral knee stability is maintained mechanically and permitted flexion and extension, whereas the 2-DOF knee model permitted flexion, extension, abduction, and adduction. Thus, the musculoskeletal 2-DOF knee model requires muscle forces crossing the knee joint to stabilize it laterally. Muscle forces were estimated by inverse dynamic analysis and optimization. The muscle force optimization process minimized the sum of the cubes of muscle stress, described by the ratio of muscle force to maximum muscle force in each muscle[24, 25]. We analyzed 13 lower limb muscles: the biceps femoris long head (BFL), semimembranosus (SM), tensor fasciae latae (TFL), rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), medial gastrocnemius (MG), lateral gastrocnemius (LG), gluteus maximus (GMAX), anterior fiber of gluteus medius (GMEDA), posterior fiber of gluteus medius (GMEDP), gluteus minimus (GMIN), and adductor longus (ADDL). Muscle forces were normalized to each participant's body weight, and time was normalized to the percentage of the gait cycle. We analyzed peak muscle forces through the stance phase from four trials. Maximum muscle force was analyzed in SM, VM, VL, GMAX, and GMEDP at the early stance phase (0–30% gait cycle) and in TFL, MG, LG, GMEDA, and ADDL at the late stance phase (31–60% gait cycle). Two force peaks were observed for BFL, RF, and GMIN at the early and late stance, and they were named according to the order of their appearance, for instance, BFL1 and BFL2.

Results are shown as the mean ± standard deviation. Normality of distribution was tested using the Shapiro-Wilk test. If normality of distribution could be assumed, data were analyzed by the paired t-test. If the normality of distribution could not be assumed, data were analyzed by the Wilcoxon signed-rank test. Meanwhile, effect size was estimated using Cohen's d or correlation coefficient (r)[26, 27]. All statistical tests were performed using IBM SPSS Statistics 24. For all analyses, the level of significance was set at 0.05.

RESULTS

The durations of BFL and ADDL muscle force were increased in the 2DOF model compared with the 1DOF model. RF muscle force showed one peak with the 1DOF model and two peak values with the 2DOF model (Fig. 1).

During early stance, BFL, RF, and GMIN muscle forces estimated by the 2DOF model were significantly increased compared with that by the 1DOF model (Table 1). Conversely, muscle forces of SM, VM, VL, GMAX, and GMEDP estimated by the 2DOF model were significantly decreased compared with that by the 1DOF model (Table 1).

During late stance, muscle forces of BFL, TFL, LG, GMEDA, GMIN, and ADDL as estimated by the 2DOF model were significantly increased compared with that by the 1DOF model (Table 1). Conversely, muscle force of MG estimated by the 2DOF model was significantly decreased compared with that by the 1DOF model (Table 1).

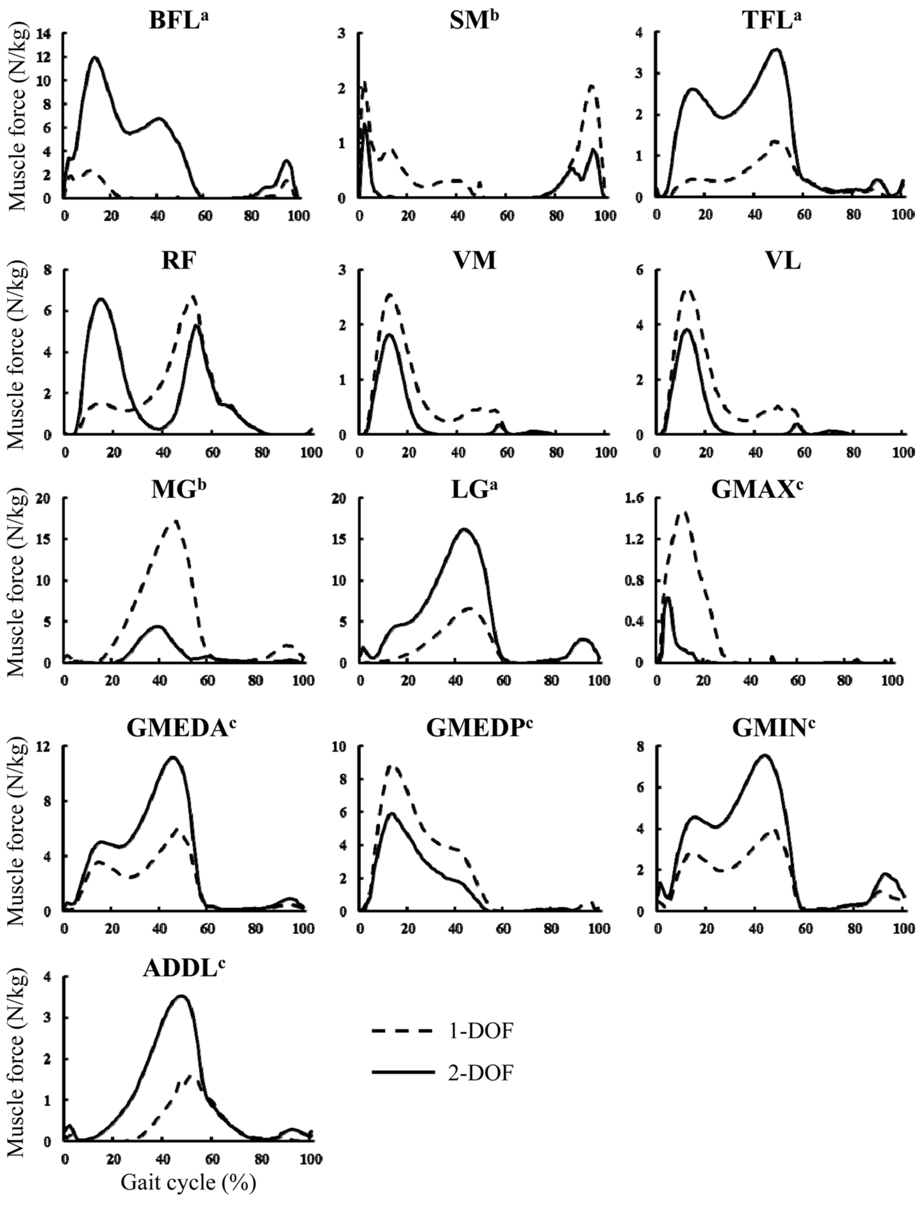

Fig. 1. Ensemble average of muscle forces measured for all participants calculated using musculoskeletal model simulation. [a]Lateral knee muscle, [b]Medial knee muscle, [c]Hip muscle.

Table 1. Muscle forces in the 1 and 2 degrees of freedom (DOFs) knee models

	1DOF	2DOF	p	Effect size (r)
BFL1	2.75 ± 0.51	12.23 ± 6.27	<0.001	0.84
BFL2*	2.08 ± 1.07	8.28 ± 4.72	<0.001	0.88
SM	2.66 ± 0.90	1.60 ± 0.54	<0.001	0.74
TFL	1.45 ± 0.40	3.93 ± 0.92	<0.001	0.88
RF1	1.84 ± 0.87	7.06 ± 3.63	<0.001	0.87
RF2	7.15 ± 1.73	6.01 ± 2.03	0.064	0.48
VM	2.64 ± 1.44	1.96 ± 1.55	0.002	0.70
VL	5.58 ± 3.07	4.16 ± 3.28	0.003	0.70
MG	17.57 ± 4.16	5.23 ± 4.70	<0.001	0.88
LG	6.74 ± 1.21	16.80 ± 5.45	<0.001	0.88
GMAX*	1.88 ± 0.48	1.20 ± 0.57	<0.001	0.82
GMEDA	6.24 ± 3.28	11.53 ± 4.09	<0.001	0.88
GMEDP	9.18 ± 1.20	6.24 ± 2.54	<0.001	0.80
GMIN1	2.93 ± 0.42	5.21 ± 1.58	<0.001	0.85
GMIN2	4.21 ± 1.57	7.80 ± 1.88	<0.001	0.88
ADDL	1.98 ± 1.11	3.75 ± 1.61	<0.001	0.87

Mean ± SD.
BFL: biceps femoris long head; SM: semimembranosus; TFL: tensor fasciae latae; RF: rectus femoris; VM: vastus medialis; VL: vastus lateralis; MG: medial gastrocnemius; LG: lateral gastrocnemius; GMAX: gluteus maximus; GMEDA: anterior fiber of gluteus medius; GMEDP: posterior fiber of gluteus medius; GMIN: gluteus minimus; ADDL: adductor longus. BFL, RF, and GMIN had two peaks at the early and late stance, thus they were named according to the order of their appearance.
*Because the normality of distribution could not be assumed in BFL2 and GMAX, they were analyzed using the Wilcoxon signed-rank test.

DISCUSSION

The purpose of this study was to define the contribution of the knee and hip muscles to lateral knee stability by comparing muscle forces estimated using musculoskeletal models with 1 or 2DOFs. Muscles whose force increased the musculoskeletal knee strength with the 2DOF model, were identified as the knee joint stabilizers in the frontal plane. As we expected, the tension force of the lateral knee muscles—BFL, TFL, and LG—was increased in the 2DOF knee model compared with the 1DOF model. In addition to lateral knee muscle, muscle forces of the hip abductor and others, including GMEDA, GMIN, RF, and ADDL, were increased in the 2DOF knee model.

The lateral knee muscles increased the tension force throughout the whole stance phase. Similarly, the external knee adduction moment was observed throughout the stance phase during normal gait. The 2-DOF knee model represents deteriorating joint condition and loss of stability resulting from dysfunction of the bones and ligaments. Therefore, internal moments generated by muscle forces are necessary to counteract the external knee adduction moment and to stabilize the knee joint in the 2-DOF knee model. Because BFL, TFL, and LG pass through the lateral of the knee joint, these muscle forces generate the internal moments counteracting the external knee adduction moment. On the other hand, muscles such as SM or MG that run inside the knee joint, strengthen external knee adduction moment by inducing internal knee adduction moment. Thus, an increase in lateral knee muscle force and a decrease in medial knee muscle force would benefit knee stability by effectively counteracting the external knee adduction moment. These muscle adaptations would enable a patient with lateral instability of the knee to utilize the GRF for controlling the center of mass during walking.

Alterations of hip muscle forces were seen in the 2DOF knee model, compared with the 1DOF knee model. These results indicate the necessity of inter-muscle coordination in the knee and the hip joint for lateral stabilization of the knee joint. Increased muscle activation is associated with a reduced activation of synergistic muscles with similar functions, and an increased activation of the antagonistic muscles with opposite functions[28]. The BFL passes the medial and posterior sides of the femoral head; thus, the BFL acts as a hip extensor and adductor. Muscle force of BFL that stabilizes the knee joint also affects the hip muscles. An increase in BFL co-relates with an increase in the antagonist muscles RF, TFL, GMEDA, and GMIN and a reduction of the synergistic muscle GAMX. Similarly, increased muscle force of the RF reduces the muscle force of the synergistic muscles VM and VL during early stance. Increased muscle force of the TFL also reduces the muscle force of the synergistic muscle GMEDA during late stance. In addition, increased co-contraction of GMIN and ADDL would contribute to maintain the balance of the hip moment on the frontal plane in the 2DOF knee model. Our results indicated the importance of hip muscle function for lateral stabilization of the knee joint.

The present study demonstrated that lateral knee muscles and hip muscles play a key role in maintaining lateral knee stability during gait. The strengthening of these muscles would help reduce the load in the medial knee compartment and help prevent the progression of knee OA. In addition, neuromuscular training, which encourages correct knee alignment under weight-bearing conditions, is beneficial in facilitating motor learning associated with inter-muscle coordination in the knee and hip joint[29, 30].

This study had several limitations. We utilized the 2DOF knee joint model without ligaments to analyze the muscles related to lateral knee joint stability. Furthermore, our knee model was not fully consistent with the in vivo knee joint. We analyzed the muscle forces in healthy participants; therefore, further studies including a more detailed knee joint model reflecting joint deformities and an adaptation of muscle activation due to pain is needed to apply the present findings to knee OA patients.

In conclusion, we determined that the lateral knee muscles such as BFL, TFL, LG, and hip muscles, in addition to abductors, contribute to lateral knee stability. These results indicated the benefits associated with strengthening of the lateral knee muscles and hip muscles, and neuromuscular training to help improve lateral knee stability.

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Conflict of interest

The authors have no conflicts of interest to declare.

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