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Article

Research on the required coefficient of friction and muscle force during recovery from unexpected slip

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Abstract: The required friction coefficient (RCOF) and muscle force are significant of exploring the human body recovery strategy after an unexpected slip. This paper quantitatively studied the muscle force distribution and response characteristics after an unexpected slip in conjunction with the variation of the required coefficient of friction (RCOF). Twenty healthy subjects were recruited for this research. Ground reaction force and gait motion data were collected by using the Vicon Motion System and AMTI force platforms. The required friction coefficient was calculated based on the ground reaction force. A musculoskeletal model was built in the Any Body Modeling System to determine the muscle forces. The results show that the RCOF changes significantly ($p < 0.001$) and approaches 0 at 12% of the gait cycle when a slip occurs, compared to non-slip conditions. During the recovery process, the values of semitendinosus, tibialis anterior, medial gastrocnemius, and lateral gastrocnemius increase by 27%, 103%, 34% and 61%, respectively. After successful recovery, there is no substantial change in muscle force in the selected muscles except for biceps femoris, medial gastrocnemius, and lateral gastrocnemius. This research suggests that biceps femoris, medial gastrocnemius, lateral gastrocnemius, tibialis anterior, and semitendinosus are with a greater impact on recovery after an unexpected slip. The paper will assist in rehabilitation training, developing effective anti-slip strategies, and conducting bipedal robot stability studies.

Keywords: the required friction coefficient; slip; musculoskeletal modeling; muscle force; recovery

1. Introduction

Gait instability is a common problem in people's daily life, especially among the elderly [1], and it not only has a significant impact on them as individuals but also plays a role in the surrounding population. Moreover, gait instability is also with devastating consequences for individuals, families, and even society [2]. Of all fatal occupational injuries, 17% are caused by slips, trips, and falls. Slips, trips, and falls have become one of the leading causes of nonfatal injuries in the workplace [3,4]. Slips, trips, and falls occur as a result of failures of normal movement and failure of attempts to regain balance in the presence of induced imbalance. Therefore, injuries caused by slips or falls have become an increasingly serious problem.

Biomechanical analysis of slip helps in assessing body segmental motion during slip [5,6]. Cham and Redfern [7] showed that slippery ground is the most common type of external perturbation that causes human gait instability. In older adults, more than 25% of injuries resulting from falls are caused by slipping, and more than 66%

of hip fractures resulting from falls occur on slippery surfaces [8]. Joint angles for normal walking on dry ground [9] and under slippery conditions [7,10–13] have been studied. It was shown that in slippery conditions, the normal gait cycle is disturbed by slippage. Also, increasing the ankle flexion angle leads to a greater incidence of slippage [13,14]. Reducing the range of motion of the hip and increasing knee flexion movements can keep the center of gravity within the support area to prevent falls. Recent studies on recovery after disturbed gait have pointed out the need for greater joint moments against greater angular momentum to regain the position and velocity of the body's center of gravity when backward sliding [15], forward sliding [16], and falling occur [17].

Ground reaction forces (GRF), also known as foot-ground contact forces, are forces equal in magnitude and opposite in direction to the pressure of the soles of the human feet on the ground during walking, running, jumping, and other sports. It is the superposition of vertical force, medial-lateral shear, and internal-external shear. The magnitude of the ground reaction forces in three directions can directly explain the ability of human lower limb movement, which is crucial of predicting slip and fall. The ground reaction force is affected by the available friction coefficient on the floor, the floor slip coefficient, and the type of shoes used [18]. The first peak of the shear force occurs at about 19% of the gait cycle [10], which is the time period when most slip occurs. The highest shear force occurs during the heel-ground contact and push-out phases of the gait cycle, which are considered as the points with the highest slip incidence [10,19]. The general characteristic of the ground reaction force during slip is the reduction of the vertical force because the weight is not completely transferred to the supporting leg [14]. Harish Chander et al. [18] investigated the mean and peak values of ground reaction forces in the sagittal plane. Jiyun Ahn et al. [20] investigated the ground reaction forces and muscle activity during anteriorly loaded under normal walking conditions. Ripic et al. [21] utilized Azure Kinect-driven musculoskeletal modeling to study the ground reaction forces and joint moments.

The required coefficient of friction (RCOF) is often cited in the literature [22] as an indicator of slip propensity, which is a consequence of the collision aspect of leg motion. A slip occurs when the required foot/floor interface friction to prevent a slip is greater than the available friction. The reduction in the peak RCOFs is caused by postural changes and adaptations during the gait cycle, reduced step lengths, reduced ground reaction forces, and significant changes in joint moments [23]. Burnfield and Powers [24] proposed a slip prediction model by the RCOF and the available slip resistance. Nagano [25] conducted a study on minimum foot clearance and coefficient of friction in older adults and proposed that the strategy to prevent falls is to control the minimum foot clearance, coefficient of friction, and the dynamic equilibrium between the center of the body and the base of support. In recent years, some scholars have studied the friction properties and slip hazard coefficients of different floors. Rafeie et al. [26] investigated the effect of different floors on friction and gait variables in older adults. Walus et al. [27] analyzed the slip hazards of floors of public-use buildings. Li et al. [28] investigated the coefficients of friction required for the process of ascending and descending stairs.

Muscle activity from electromyography consists primarily of isometric or eccentric muscle actions of the lower limb muscles that allow efficient storage and transfer of energy between limb segments and brief, high-energy concentric muscle actions to help move the body forward [29]. During unexpected slip, reactive strategies appear which are defined as the primary corrective response brought about by muscular forces and corrective moments to re-establish dynamic balance after slipping. Some scholars have studied the muscle activity of lower limb muscles such as quadriceps femoris, gastrocnemius, and hamstring muscles under slippery conditions [30–32], and the amplitude is higher and the activation time is longer [33]. To better understand the response mechanisms of muscle, Hof and Duysens [34] analyzed the reaction of muscles around the ankle joint (peroneus longus, tibialis anterior, and soleus) when they were disturbed by mid-lateral balance during walking, and then obtained the balance mechanism. Bakiewicz [35] used the Vicon system and the Kistler platform to obtain kinematics and dynamics parameters, and modeling musculoskeletal in the Any Body Modeling System to calculate the joint reaction force and muscle force, so as to obtain the joint reaction force and muscle force during classical and jazz rotation.

Up to now, most of the above studies carried out gait slip analysis from a certain aspect (e.g., ground reaction force, joint force, etc.) [25,36], which could not well explore a series of reactions of human lower limbs caused by unexpected slip as a whole. For the studies on the muscle, researchers often paid more attention to qualitatively evaluate muscle activity [37,38], not providing the magnitude of muscle force. The purpose of this study was to obtain the magnitude of muscle force during the recovery from unexpected slips and research the relationship between the RCOF and muscle forces from different muscles. In this paper, we calculated the RCOF on the GRF in three directions which were collected by the AMTI force platform and built a musculoskeletal model by using the AnyBody Modeling System to determine the muscle forces. Based on the results of lower limb joint angle, joint torque, joint force, etc. in literature [25,36], we analyzed and explored the mechanism of muscle force induced joint response, and further generated different GRF in three directions through the series reaction of lower limbs, which will change the value of RCOF which is a key factor in predicting slip.

2. Methods

2.1. Participants and muscles

2.1.1. Participants

Twenty healthy young males (Age: 24 ± 0.87 years, Height: 175.6 ± 12.3 cm, Weight: 75.3 ± 8.94 kg) were recruited for this trial. Subjects were required to have no history of lower limb trauma, no type of lower limb surgery, no back or pelvic problems, and no neuromuscular disorders or balance problems in the last six months. There was no strenuous exercise within 24h prior to the start of the trial to avoid residual muscle fatigue from over-exercise that could affect the data [20]. All subjects participated voluntarily and details of the trial were communicated to the subjects in advance. This study was performed following the Declaration of Helsinki, the study protocol was approved by the Institutional Review Board of Luoyang Institute of Science and Technology, and the written informed consent was provided and signed by all subjects.

2.1.2. Muscles

This paper investigates the pattern of change of the lower limb hip, knee, and ankle joints in terms of gait stability, so the muscle groups associated with the operation of the three joints are selected for research. Through the knowledge of human anatomy, muscle groups can be obtained on the surface, but also in the deep. From a medical point of view, the muscles on the surface of the skin are easy to measure surface EMG (sEMG) signals, while the muscles in the deeper layers are not easy to measure, and although technically possible, it is not easy to find subjects and is not ethically feasible. Therefore, the muscles selected in this study require that they can be touched by hand, which is convenient for measuring the EMG signal, and it is also convenient to compare and analyze the data obtained from the musculoskeletal model to improve the credibility of the study. According to human anatomy, the selected muscles are shown in **Table 1**.

Table 1. The selected muscles for the motion of each joint.

Hip		Knee		Ankle	
Extensor group	**Flexor group**	**Extensor group**	**Flexor group**	**Plantar flexor group**	**Dorsiflexor group**
biceps femoris (BF) semitendinosus (ST)	rectus femoris (RF)	vastus lateralis (VL) vastus medialis (VM) rectus femoris (RF)	medial gastrocnemius (MG) lateral gastrocnemius (LG) semitendinosus (ST)	medial gastrocnemius (MG) lateral gastrocnemius (LG)	tibialis anterior (TA)

2.2. Experiment

2.2.1. Instrument

The experiment instruments include data acquisition equipment, test trail, and protection devices, as shown in **Figure 1**. The data acquisition equipment consists of three-dimensional motion analysis, which was performed by using a ten-camera motion analysis system (Vicon Motion System, Oxford Metrics Ltd., Oxford, UK) and AMTI force plate (AMTI Corp., Newton, MA, USA, 464 mm × 508 mm). The test trail consists of flat plates that are connected at both ends of the trail and three AMTI force plates that are arranged adjacent to each other in the center. The height of the support plate is required to be the same as the height of the AMTI force plates. All test trials were conducted on the walkway with the middle AMTI force plates for data collection. The protection device consists of a linear slide and a hook, which is connected to the safety belt to protect the subject in the event of a slip and fall.

The coefficient of friction of different floor materials (marble, glass, ceramic tiles, wooden floors) in combination with wet slippery media (water, soapy water, vegetable oils) has been explored [6], and it was shown that marble and vegetable oils have the worst anti-slip properties, which provide the smallest coefficient of friction. Therefore, the floor material selected in the test is marble which is placed on an AMTI force plate. The wet slip medium was vegetable oil, which was uniformly applied to the marble during the test, and the test trail was treated as a normal marble straight trail or marble straight trail coated with wet slip medium [39,40], with all the rest of the parameters being the same.

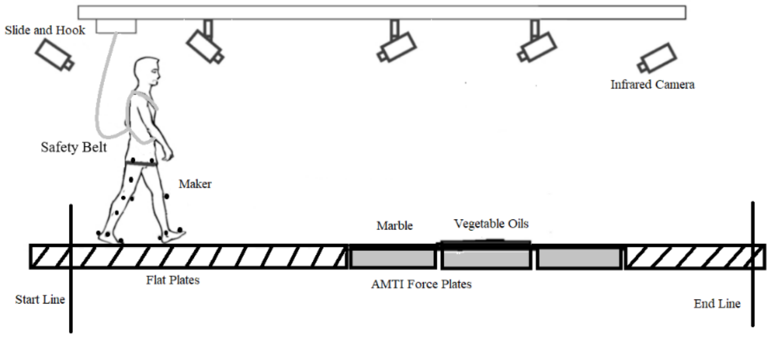


Figure 1. Instrument and procedures.

2.2.2. Procedures

To make the test closer to daily life and protect the subjects, subjects positioned and strapped inside the harness connecting to the slide and hook were briefed on the walking gait conditions and allowed to practice walking at their normal pace across the lab walkway. Several practice gait trials were performed until the subjects walked normally and at the same speed. After the accommodation period, subjects took a 30-minute break and were informed that they would be performing different walking trails and they would be asked to leave the room following several times of walking trails. Subjects were informed that during their absence, the principle investigator would either review data from previous trials or, on one occasion, apply oil that would make the floor slippery. They were not told during which trial the contaminant would be applied nor the location of the contaminant application. Next, each subject was asked to wear exactly the same shoes and tests at normal walking speed under two conditions without knowing the ground medium, and the subjects continued to walk after slipping until the collector issued a stop command. Following a repeated number of gait trials under normal dry conditions, one particular trial was chosen randomly to be the unexpected slip trial. Only the first novel slip was included to avoid predictive feed forward modifications to locomotion and reveal non-adapted spontaneous motor responses. After the slip occurred, one group trial was completed. At the same time, all subjects were given a 5-minute rest after every group trial to exclude the effect of muscle fatigue on the test results [41]. Each subject performed 10 trials, all of which included 2–3 slips.

2.2.3. Musculoskeletal model

AnyBody Modeling System provides abundant human musculoskeletal models. In this paper, the human body model in AnyGait is selected as the basis, combining with the subject's height, weight, thigh, calf, and foot length, and other morphological parameters, using the Scaling Length MassFat function to scale the model based on AnyScript programming. The gravity field was 9.81 N/kg. The musculoskeletal model was obtained as shown in **Figure 2**. The problem of muscle redundancy caused by the number of muscle forces being greater than the degrees of freedom of the musculoskeletal model occurs when performing inverse dynamics analysis calculations, which is solved in this study by using the optimization method provided in AnyBody Modeling System [42-44].

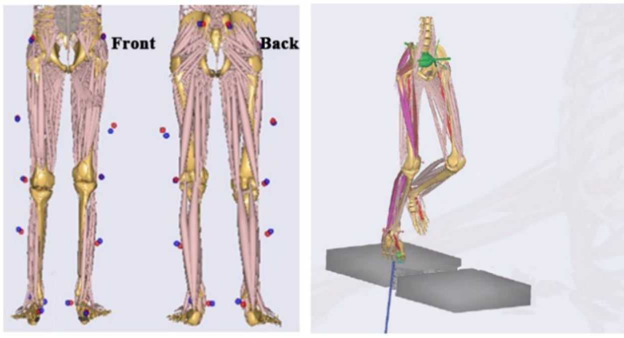


Figure 2. The musculoskeletal model.

2.3. Data processing

A three-dimensional motion capture system captured gait movements that should be divided according to the gait cycle. A complete gait cycle starts with the heel touchdown on either side and ends with the heel touchdown on the same side again. This time is standardized. 0% represents the heel touchdown, and 100% represents the heel touchdown again on the same side. A standard gait cycle includes two phases: the stance phase and the swing phase. For the normal gait cycle, the standing phase accounts for about 60%~62% of the entire gait cycle, and the swing phase accounts for about 38%~40% of the entire gait cycle. The stance phase can also be subdivided into three stages according to the heel touch bottom and toe off the ground. The first double-support phase refers to the right foot heel to the left foot tip off the ground moment, accounting for 12% of the whole gait cycle. The single support phase refers to the left toe off the ground to the left heel landing moment, accounting for 38% of the whole gait cycle. The second double support phase refers to the moment of the left heel landing to the moment of the right toe off the ground, accounting for 12% of the entire gait cycle. The swing phase ends from the moment of the right foot tip off the ground to the moment of the right foot tip landing [45].

The Vicon Motion System (VMS) was used to capture the trajectories of 16 spherical marker points located in the foot, calf, thigh, and waist, and the experimental motion data were obtained at a frequency of 100 Hz. The AMTI force platform (464 mm × 508 mm) was used to collect the ground reaction force at a frequency of 1000Hz, and the relevant data were collected synchronously with the Vicon system through the digital-to-analog converter [42].

Because the ground reaction force is related to body weight, it is standardized by dividing the reaction force by body weight [35]. After standardization, the RCOF is calculated by GRF. The RCOF value is the ratio of the resultant force of the anterior-posterior shear force (F_y) and the medial-lateral shear force (F_x) to the vertical force (F_z) [28]. The calculation formula of RCOF is:

$$RCOF = \frac{\sqrt{F_x^2 + F_y^2}}{F_z} \tag{1}$$

The data in the international standard format collected by Vicon Motion System and AMTI are imported into AnyBody Modeling System 7.0 in the form of C3D files using the C3D-to-Script program to define the movement of the musculoskeletal model drive its movement and calculate the lower limb muscle forces, joint reaction moment and other parameters. The kinematic and kinetic data were filtered using a Butterworth low pass filter with 10 and 15 Hz cut-off frequencies, respectively. For estimating muscle forces, static optimization was solved by minimizing the polynomial muscle recruitment criterion, defined as:

$$G = \sum_{i=1}^{nM} \left(\frac{f_i^M}{N_i} \right)^3 \tag{2}$$

where nM is the number of muscles, f_i^M is the respective muscle force, N_i is equal to the isometric muscle strength in the simple muscle model [42–44].

2.4. Statistical analyses

The values of all measurements are presented as mean ± standard deviation (M ± SD). The software (MyoResearch XP Master Edition 1.07) provided by Telemyo 2400 DTS, which was used in the analysis of sEMG. The data of GRF, RCOF and muscle forces were smoothed and plotted curves in Origin 2022. Post-hoc comparisons using paired *t*-tests were conducted on GRF, RCOF and muscle forces during the gait cycle within SPSS 27.0 (IBM Corp, Armonk, NY, USA). Pearson's correlation analysis was performed between the muscle activity from Telemyo 2400 DTS and calculated in the AnyBody Modeling System and used to examine the relationships between the RCOF and muscle force. The significance level of 0.05 was used throughout.

3. Result

Based on the real experimental process, all subjects completed the test according to the requirements, and all slips were backward slipping. At the same time, sliding failures, falls, or obstruction by markers that caused segment loss were excluded, leaving 55 sliding tests and 126 anti-skid tests for analysis.

3.1. Validation model

From **Figure 3**, it can be seen that for normal walking and recovery after accidental slipping, the correlation coefficients between the muscle activity of the semitendinosus (ST), biceps femoris (BF), and rectus femoris (RF) measured in the experiment and the muscle activity calculated in the AnyBody modeling system are all over 0.6. Those results indicated a strong alignment between the parameters generated by the musculoskeletal model and the actual obtained muscle activities, confirming the validity of the musculoskeletal model developed in this study [42,43].

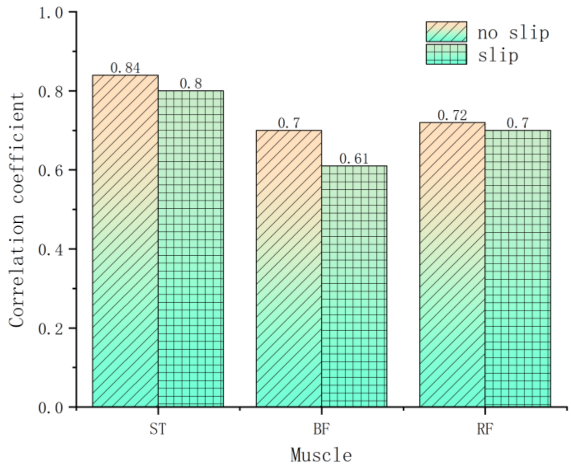


Figure 3. Correlation coefficient of muscle activity.

3.2. Ground reaction force and RCOF

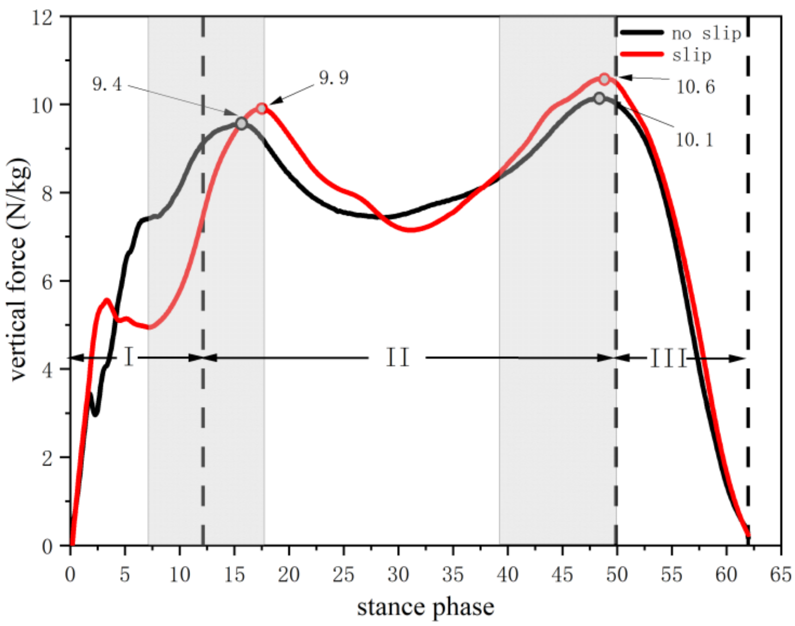
As shown in **Figure 4**, both the no-slip and slip conditions exhibit pronounced bimodal characteristics in the vertical force (VF). The peak values during the slip condition (9.9 N/kg and 10.6 N/kg) were notably higher than those observed in the no-slip condition (9.4 N/kg and 10.1 N/kg). During the first double support phase, there was a significant difference in the VGRF ($p < 0.001$, shown in **Table 2**). Specifically, when the heel was in contact with the ground, the VF was significantly lower in the slip condition due to the subject slipping on the oil surface. In both the single-support phase and the second double-support phase, the VF followed a similar trend and also showed a significant difference ($p < 0.001$, shown in **Table 2**). Within the range of 7%~17%, highlighted by the gray bar areas, the VF gradually increased, reaching its maximum at 17%. In the range of 38%~50%, compared with the non-slip condition, the VF increased in the recovery phase following a slip.

In **Figure 4**, under the no-slip condition, the medial force (MF) began to manifest after 7%, while the lateral force (LF) became apparent when the toe was off the ground. In contrast, when slip occurred, the change in MLF within the first 7% of the gait cycle was chaotic, showing a more pronounced difference compared to the no-slip condition ($p < 0.001$, shown in **Table 2**). The LF was evident within the 7%~15% range, as indicated by the gray bar areas, after which it transitioned to the MF. The MF was exhibited in 13%~53%, displaying a similar trend to that of the no-slip but with a significant difference ($p < 0.001$, shown in **Table 2**). Notably, its value exceeded the peak value observed in the no-slip condition. Furthermore, the LF appeared earlier when the slip occurred, as highlighted in gray bar areas in **Figure 2**, with its peak value surpassing that of the no-slip scenario. This suggested that the MF played a critical role in post-slip recovery, accompanied by an increased LF following the slip.

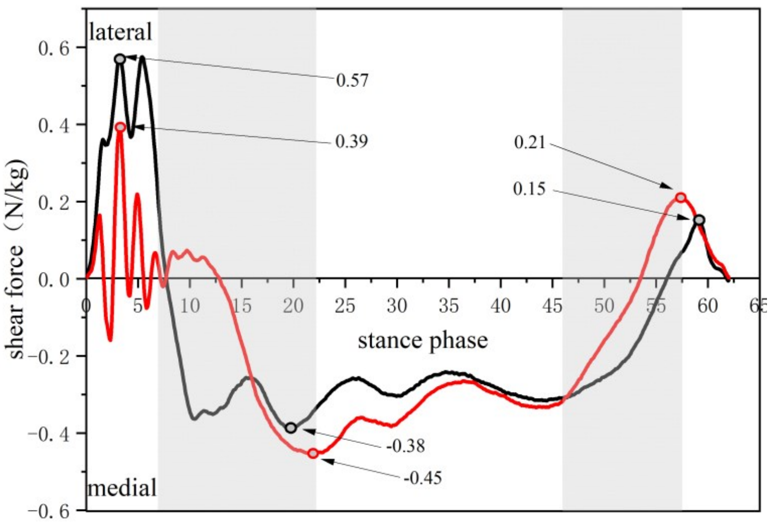
As can be seen in **Figure 4**, the anterior force (AF) decreased during the occurrence of unintentional slip ($p < 0.05$, shown in **Table 2**), while the posterior force gradually increased, remaining greater than that in the no-slip condition throughout the single-support phase ($p < 0.001$, shown in **Table 2**). The APF increased significantly during the slip recovery action within the 7%~38% range of the stance phase, as indicated by the gray bar areas, along with an increase in peak value.

Table 2. Results of ground reaction force and RCOF.

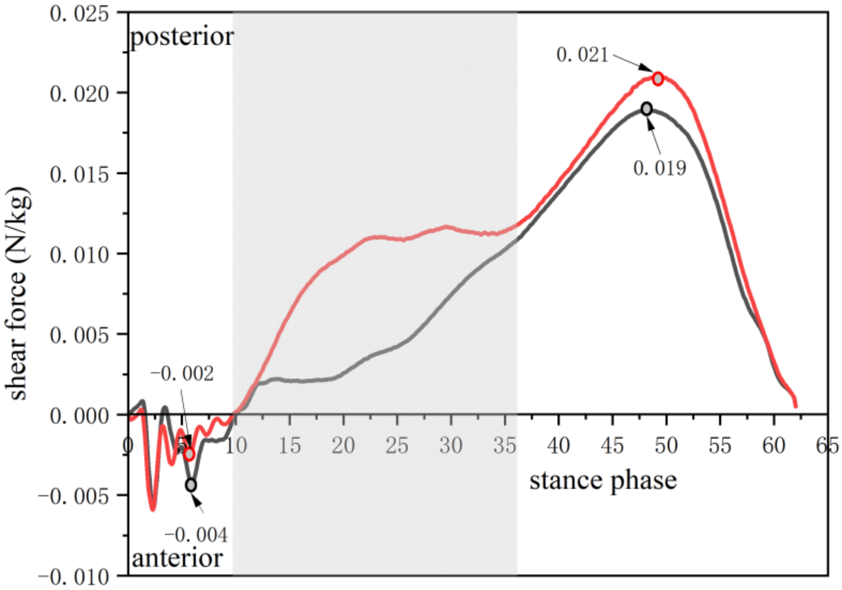
Gait Cycle (0-100%)	**Parameters**		**No slip (Mean ± SD)**	**Slip (Mean ± SD)**	***P* (0.05)**
First double support (0%-12%)	GRF (N/kg)	VF	5.963 ± 2.433	4.895 ± 1.498	<0.001
		APF	−0.0013 ± 0.0018	-0.0010 ± 0.0016	<0.001
		MLF	0.131 ± 0.33	0.060 ± 0.103	P = 0. 003 < 0.05
	RCOF		0.08 ± 0.063	0.03 ± 0.035	<0.001
single support (12%-50%)	GRF (N/kg)	VF	8.547 ± 0.896	8.719 ± 1.076	<0.001
		APF	0.0090 ± 0.006	0.0123 ± 0.004	<0.001
		MLF	−0.295 ± 0.036	-0.311 ± 0.101	<0.001
	RCOF		0.035 ± 0.008	0.034 ± 0.011	P = 0.689
Second double support (50%-62%)	GRF (N/kg)	VF	5.573 ± 3.348	5.961 ± 3.377	<0.001
		APF	0.0101 ± 0.006	0.0123 ± 0.007	<0.001
		MLF	−0.044 ± 0.127	0.060 ± 0.112	<0.001
	RCOF		0.038 ± 0.034	0.035 ± 0.026	P = 0.013 < 0.05



(a)



(b)



(c)

Figure 4. The curve of ground reaction force. **(a)** The curve of VF; **(b)** The curve of MLF; **(c)** The curve of APF.

Note: "I", "II" and "III", respectively, indicate the first double-supported phase, the single-support phase, and the second double-supported phase, with gray bar areas indicating the recovery of GRF.

Comparing the no-slip to the oil-slip, the RCOF during slip events was significantly smaller ($p = 0.003 < 0.05$, shown in **Table 2**) in the oil-slip case, as illustrated in **Figure 5**. The curves indicated that during the first double support phase, the peak value of RCOF in the post-slip recovery case decreased continuously, approaching nearly 0 at 12%. As shown in the grey bar area, RCOF gradually increased, as the body transitioned into the recovery state, peaking at approximately 20%, with stability similar to the no-slip condition ($p = 0.689 > 0.05$, shown in **Table 2**). In the second double-support phase, indicated by the gray bar, there was a significant difference ($p < 0.001$, shown in **Table 2**). This change occurred early in the slip scenario because, as the toe approached leaving the oil surface, the subject anticipated the slippery environment, resulting in a lower RCOF [23]. This proactive adjustment aimed to prevent backward slipping when the foot lifted off the ground.

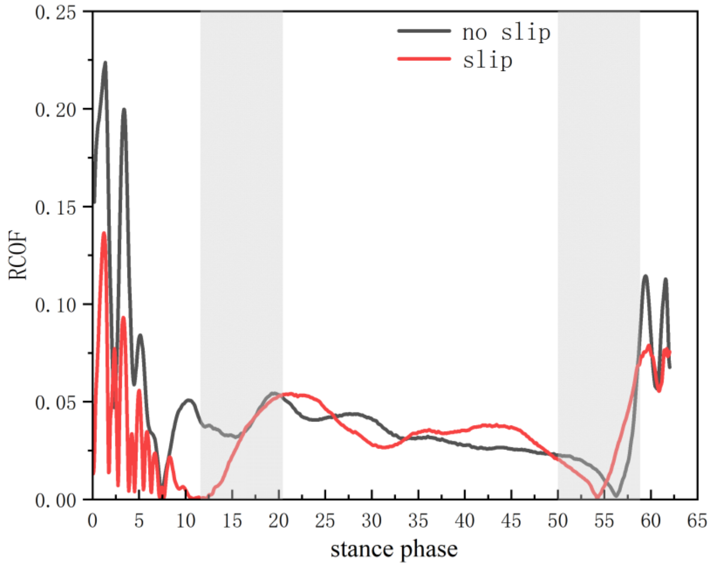


Figure 5. The curve of RCOF.

Note: The gray bar areas indicate the recovery-specific phase of the stance phase.

3.3. Muscle force

As shown in **Figure 6**, the maximum values of BF and ST in the hip extensor group (10.9 N/kg, 9.4 N/kg; 11.9 N/kg, 11.1 N/kg) were significantly reduced following an unexpected slip, with both muscle groups exhibiting a similar trend during the single-support phase. The mean of muscle force produced by BF (4.29 ± 3.44 N / kg) was significantly greater than that in the non-slip (4.71 ± 3.39 N/kg) (p < 0.001). During recovery from the slip, the range of muscle force exhibited substantial variability, with an accelerated rate of change. The timing of RF onset in the hip flexor group remained consistent, however, the maximum muscle force post-slip (11.1 N/kg) was less than that in the no-slip condition (11.9 N/kg), with a significant difference noted in the gray area bar (p < 0.001).

The VM exhibited a delayed onset of muscle force generation after slipping compared to the no-slip condition, although the overall trend remained unchanged, showing no significant difference in the gray bar area (p = 0.962 > 0.05). Similarly, the VL, part of the knee extensor group, showed a delayed response in muscle force generation post-slip, aligning with the same trend observed during the stance phase. Notably, the maximum muscle force generated during the post-slip recovery phase (6.5 N/kg) was greater than that in the no-slip condition (4.2 N/kg). In the knee flexor group, the LG demonstrated a longer duration of muscle force generation during the slip (15%~60% of the gait cycle) compared to the no-slip condition (20%~55% of the gait cycle), with a significant difference observed in the grey bar area of the graph (p < 0.001). As depicted in the grey bar area, the muscle force generated by the TA decreased significantly (p < 0.001) during recovery from slip (4.72 ± 1.18 N/kg) compared to the no-slip condition (7.04 ± 2.69 N/kg). During the swing phase, TA changed significantly (p < 0.001), while the remaining muscle forces did not substantially change. Except for RF and LG, the other muscles generated force approximately 3% earlier in the late swing phase (80%~100% of the gait cycle).

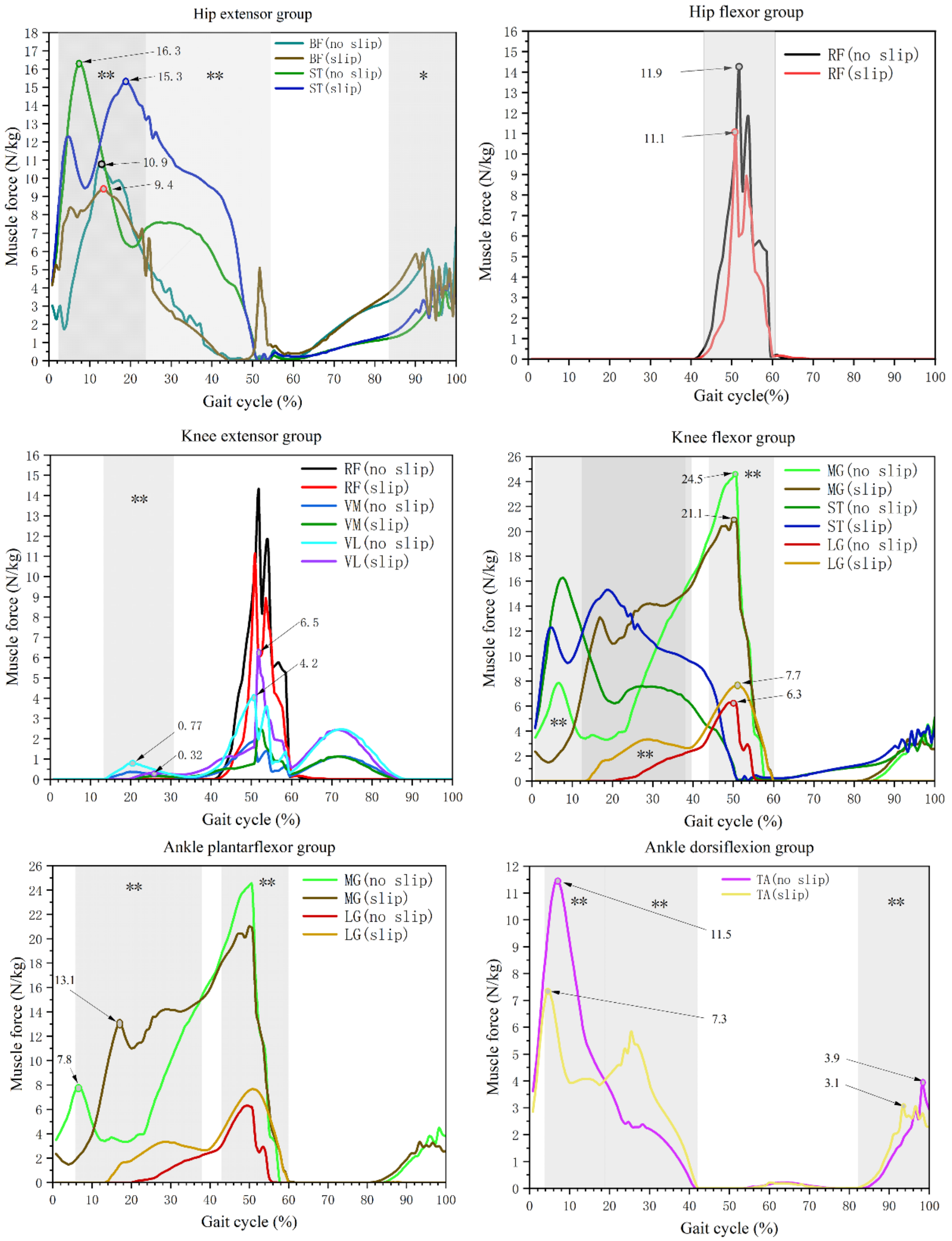


Figure 6. The change of muscle force.

Note: The gray bar areas indicate specific phase differences in the gait cycle. "**"and "*" indicate p<0.001 and p<0.05 respectively.

3.4. Correlation

To further investigate the relationship between properties of muscle response and VF, MLF, and RCOF during recovery after a slip, the correlations presented in **Figure 7** were analyzed in this paper. Among the selected muscles, except for ST and TA, the rest were closely and negatively correlated with RCOF, with correlation values less than 0.4.

The correlation values of ST and TA were close to 0. ST, VM, VL, RF, MG, and LG were all positively correlated with VF, with the strongest for MG (r = 0.85), and LG (r = 0.68). Concerning MLF, ST, MG, and LG demonstrated negative correlations, with ST (r = −0.54), and MG (r = −0.77) showing a stronger association with MLF.

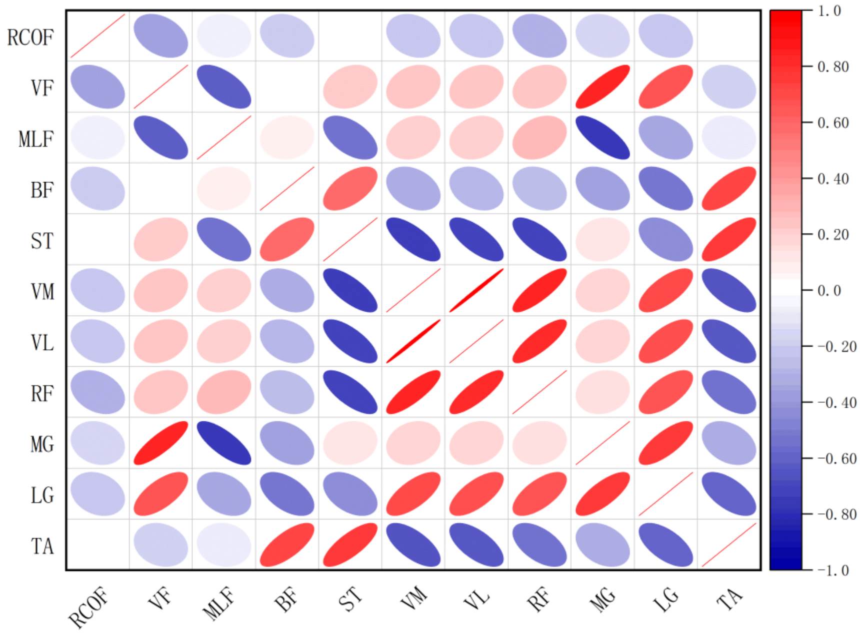


Figure 7. The correlations of VF, MLF, RCOF and muscles.

4. Discussion

In the early stage of slip (5%~10% of the stance phase), the VF experienced a sudden decrease, accompanied by a significant drop in MLF, which peak in the lateral direction at 10%. This contrast with no-slip trail, where the APF also declined in the same direction. The alterations in these directional forces lead to substantial fluctuations in the required coefficient of friction (RCOF), which reaches its lowest point at 10% of the stance phase. This phenomenon occurs because the available coefficient of friction (ACOF) on the oiled surface diminishes upon heel contact, while the RCOF required for stable walking remains unchanged. Consequently, when ACOF < RCOF, slipping occurred, causing the foot to shifted forward and the center of mass (COM) to moved backward, resulting in gait instability. To regain balance, it is crucial to reduce the stability angle and maintain the center of gravity within the base of support (BOS). This was achieved by decreasing LF and increasing VF and PF, effectively moving the center of gravity forward and upward. Additionally, the adjustment of the RCOF was necessary to ensure ACOF > RCOF, thus restoring balance. Notably, after an unexpected slip, the RCOF reached its peak at the moment the heel touched the ground and the toe left the ground, indicating that these moments were critical for gait stability, supporting the observation that the foot often moved backward when the toe lifted off.

Research [46,47] had demonstrated that muscle action played a crucial role in predicting the ability to restore balance after an unexpected slip, with muscle weakness contributing to post-slip instability and an increased risk of falls [48]. This study specifically examined the changes in muscle force magnitude in the slipping leg during the autonomous recovery of balance following an unexpected slip, shifting away from the qualitative analyses typically performed using surface electromyography (sEMG) [49]. The findings regarding muscle force provided insights into previously observed adaptations to slippery surfaces, highlighting their impact on kinematics (joint angles) and kinetics (joint moments) [15].

The stability of individuals during walking referred to their ability to maintain balance against various disturbances, particularly on slippery surfaces such as oiled trails. Research [50] indicated that when the heel struck such surfaces, the RCOF can exceeded the available ACOF, leading to slips. When a slip occurred, the ground moved forward relative to the body, altering the BOS and increasing the distance between the COM and the center of pressure (COP). This imbalance created a backward-tilting moment, causing the body to slip backward [15,51]. To regain balance, the body typically employed a hip-jacking action, raising the hip joint. This movement engaged the hip extensors (such as the BF and ST) [7,52], resulting in changes to hip joint torque and angle [17]. The increased extension helped prevent further backward motion and enlarged the BOS. As the COM shifted forward, muscle forces in the knee flexors (BF, ST) and ankle plantar flexors (LG, MG) increased, while the dorsiflexor muscle force (TA) decreased. This coordinated action enhanced knee flexion and ankle plantar flexion [7,52], facilitating quicker heel-ground contact and optimizing the foot's angle with the ground [44]. As recovery progresses, the RCOF adjusted to remain less than or equal to the ACOF [50], preventing further slipping and keeping the COM within the BOS [52]. Once the slip had ceased, movements of hip extension, knee flexion, and ankle plantar flexion helped shift the COM from the supporting leg to the sliding leg [53,54]. This transition allowed the non-sliding leg to disengage from the ground as the body stabilizes. Research indicated that the recovery phase, comprising 15% to 45% of the gait cycle, shortened the single support phase [32]. Overall, balance regulation after an accidental slip relied on the coordinated action of multiple muscle groups, rather than the effort of a single muscle.

During the second double support phase, as the COM shifted to the non-slip leg, the heel of the slipping leg began to lift off the ground. Consequently, the GRF through the slipping leg decreased, leading to a reduction in muscle force across various joint muscle groups in that leg. Since the COM had not yet fully transferred to the non-slip leg [55], the slipping leg continued to exert force on the oiled surface to provide the RCOF. At this stage, the body had adapted to the slippery conditions and reached a new equilibrium. The adjustments in muscle force across different groups resulted in a decrease in hip flexion torque, transitioning the knee joint from flexion to extension and the ankle from plantar flexion to dorsiflexion [56]. This change reduced the contact area between the foot and the ground, which in turn altered the ground reaction force. Through these coordinated actions, the slipping leg could still generate enough RCOF to meet the body's stability needs, allowing the COM to shift fully to the non-sliding leg until the toe of the slipping leg finally left the ground.

In the early swing phase (62% to 80% of the gait cycle), the muscle force curves for both slip and non-slip conditions displayed significant overlap, indicating similar muscle activation patterns. However, in the later swing phase (80% to 100%), the muscle force in the slipping leg adjusted earlier, aligning closely with the normal walking pattern. This change suggested that the body anticipates the next movement, activating hip, knee, and ankle muscles proactively to enhance stability for the upcoming step [57].

Studies indicated that specialized training programs targeting slip recovery can significantly benefit adults at risk of falling by enhancing their movement responses [55,57]. Activities like shadowboxing have been shown to improve balance and gait stability [58]. This research aimed to provide accurate muscle parameters that can inform training for slip perturbations, ultimately enhancing gait stability through targeted muscle force and joint mobility exercises. However, the current study focused solely on the superficial muscles of the slipping leg. Future research should incorporate both the slipping and non-slipping legs, as well as deeper muscle groups, to gain a comprehensive understanding of biomechanical responses during recovery from unexpected slips.

5. Conclusions

This paper investigated the RCOF and magnitude of muscle force from different muscles acting on balance recovery following an unexpected slip. During the recovery from an unexpected slip, the human coordinated control of increasing the muscle force (ST (27%), TA (103 %), MG (34 %), LG (61 %) ($p < 0.001$)) to increase the magnitude of ground forces which include vertical force, posterior shear force and medial shear forces to adjust the RCOF. Meanwhile, the change of muscle forces contributed to the body shifting the COM from the sliding leg to the dragging leg rapidly and maintained the COM in the BOS to promote the body's recovery from an unexpected slip. After the balance was restored, the maximum values of VL, RF, and LG muscle forces all increased which would influence the values of ground force and RCOF to avoid the slip. In the later stage of the swing phase, people will make a pre-judgment in advance, by generating muscle force in advance and changing the magnitude of muscle force, especially the ankle joint muscles, adjusting the lower limb posture as well as the angle of the ankle joint contact with the ground to prevent the recurrence of slip. The research will provide a new approach to the study of muscle force response characteristics in human locomotion, as well as assist in rehabilitation training, development of effective anti-slip strategies, and bipedal robot stability studies.

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