

Original Article

Effects of Vibration Frequency and Loading on Lower Limb Muscle Activation and Time-Varying Multi-Muscle Coactivation in Whole-Body Vibration Training among Recreational Runners

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Abstract

Objective: This study examined the effects of vibration frequency and additional loading on lower limb muscle activation and coactivation during whole-body vibration (WBV) training in recreational runners. **Methods:** Twenty-five male runners performed WBV training at six frequencies (0–50 Hz) under two loading conditions (no load vs. 20% body mass). Surface electromyography was used to record electromyography root mean square (EMGrms) in 11 lower limb muscles. Coactivation of six muscles was analyzed using the time-varying multi-muscle coactivation function. 1D statistical parametric mapping and two-way repeated-measures ANOVA were used to assess differences. **Results:** EMGrms was significantly influenced by vibration frequency and loading. Additional loading increased activation in the gluteus maximus, quadriceps, hamstrings, and soleus ($p < 0.001$). Frequency effects were notable in the gluteus maximus and hamstrings, with 10–30 Hz inducing higher muscular activation than the no-vibration case ($p < 0.01$). Coactivation was significantly high with loading ($p < 0.001$), particularly during the eccentric squat phase, but frequency had no significant effect. **Conclusion:** Additional loading enhances muscle activation and coactivation, and vibration frequency primarily affects muscular activation. These findings highlight the importance of external load and frequency in optimizing WBV training for enhancing sports performance and preventing injuries in runners.

Keywords: Lower Limb, Muscle Coactivation, Surface Electromyography, Time-Varying Multi-muscle Coactivation Function, Whole-body Vibration

Introduction

Whole-body vibration (WBV) is a training technique that stimulates the muscles and nervous system by transmitting mechanical vibrations to the body through a vibrating platform¹. Although the exact mechanism of WBV training remains unclear, the most widely discussed theory is tonic vibration reflex. Tonic vibration reflex is a process in

which mechanical vibration stimulates muscle spindles, which leads to the activation of alpha motor neurons via group Ia afferents, thereby inducing muscle contraction^{2,3}. Previous studies have reported that WBV training has beneficial effects on balance and postural control^{4,5}, muscle strength^{6,7}, explosive power of the lower extremity^{8,9}, neuromuscular activation⁹, and body fat reduction¹⁰ among different groups. Given these advantages, WBV training is widely applied in rehabilitation, fitness, and elite athletic training^{11,12}.

Neuromuscular activation is the process by which motor neurons transmit signals to muscle fibers, resulting in muscle contraction¹³. This mechanism is essential for executing controlled and precise movements and maintaining human movements¹⁴. Previous studies have demonstrated the effects of WBV training on neuromuscular activation. Pollock et al.¹⁵ reported that

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WBV training increases muscle activity by 5%–20% in the thigh muscles (rectus femoris, biceps femoris, and gluteus maximus) and by 5%–50% in the lower leg muscles (soleus, gastrocnemius, and tibialis anterior) compared with no-vibration conditions. Hazell et al.¹⁶ found that WBV training increases vastus lateralis activation by 3.7%–8.7% and biceps femoris activation by 0.4%–2.0% compared with training without vibration. Moreover, the combination of WBV training with external loading has been explored to enhance these effects. Ritzmann et al.¹⁷ reported that WBV training with additional loading results in a 10%–20% increase in muscle activity in the lower extremity muscles (rectus femoris, vastus medialis, tibialis anterior, gastrocnemius, biceps femoris, and soleus) compared with WBV without additional load. However, these studies focused on the effect of WBV training on the activation levels of individual muscles and did not examine the effects of WBV training on muscle coactivation.

Muscle coactivation refers to the simultaneous activation of two or more muscles and has a substantial effect on enhancing joint stability and movement accuracy^{18,19}. Muscle coactivation increases joint stiffness, which enhances joint stability by elevating ligament tension and improving the contact between joint surfaces during movement²⁰. In particular, the coactivation of lower extremity muscles provides stabilization of the knee and ankle during exercise²¹, which is crucial for the reduction of the risk of lower limb injury and the enhancement of sports performance²². Therefore, understanding and optimizing muscle coactivation patterns are essential for injury prevention and athletic performance maximization.

Muscle coactivation is quantified via surface electromyography (sEMG) and subsequently examined using mathematical equations. Traditionally, coactivation was assessed by calculating the ratio of antagonist muscle activity to agonist muscle activity. However, this traditional approach, which relies on sEMG signals from individual muscles (i.e., one antagonist and one agonist), has limited capacity to accurately capture coactivation patterns among multiple muscles during dynamic exercise. To address these limitations, Ranavolo et al.²³ introduced a novel analytical approach called time-varying multi-muscle coactivation function (TMCf). Compared with traditional coactivation analysis, the TMCf approach provides a more comprehensive analysis by assessing the coactivation patterns of multiple muscles over time, thereby providing deeper insights into motor control.

Despite the growing interest in WBV training, only a few studies have comprehensively examined the activation of multiple lower limb muscles and their coactivation under varying WBV parameters. Previous research has shown that vibration frequency and external loading can modulate neuromuscular responses^{17,24,25}, highlighting the need to investigate how these variables influence muscle activation and coactivation during WBV exercises. Therefore, this study aims to evaluate lower limb muscle activation and coactivation patterns during WBV training under different

vibration frequencies and loading conditions. On the basis of prior evidence demonstrating that external loading and increased vibration frequency enhance neuromuscular activation^{15,16,17}, we hypothesize that high vibration frequencies and the presence of additional loading lead to increased muscle activation and coactivation. The findings can enhance the understanding of neuromuscular control during WBV and offer valuable insights for optimizing training protocols.

Materials and Methods

Participants

A priori power analysis was performed using G*Power 3.1.9.7²⁶, which revealed that at least 20 participants were needed, with a significance level (α) of 0.05, effect size of 0.25, and power of 0.8. Hence, 25 healthy young male recreational runners (weekly running distance ≥ 15 km) were recruited for the study (age: 21.4 ± 3.0 years; height: 177.9 ± 0.1 cm; body mass: 71.7 ± 8.3 kg; BMI: 22.7 ± 2.4 kg/m²). Individuals with known contraindications to WBV exercise, such as cardiovascular diseases (e.g., coronary artery disease and uncontrolled hypertension), neurological, musculoskeletal, or rheumatologic disorders that could impair safe performance, were excluded^{27,28}. Those with a history of lower extremity injury or had surgery within the past six months were also not included.

Experimental Procedures

A sonic vertical-rhythm vibration machine (FL-T801, Frosino, Guangdong, China) was used to conduct WBV training. The participants underwent WBV training under 12 different conditions, which combined six frequencies (no vibration, 10 Hz, 20 Hz, 30 Hz, 40 Hz, and 50 Hz) and two loading conditions (no additional loading and with additional loading), to investigate the effects of frequency and loading conditions on lower extremity muscle activation and coactivation. A weighted vest corresponding to 20% of a participant's body mass was used to simulate additional loading. This magnitude was based on previous weighted vest training studies, which typically applied 10%–20% of body mass^{29,30}, and on our pilot testing, which confirmed that 20% loading allows for safe and feasible squatting on the vibration platform. The vibration amplitude was set to the maximum of 6 mm because previous studies have shown that high amplitudes are effective for WBV training^{31,32}. Notably, the corresponding peak accelerations were measured at 1.13 (no vibration), 1.66 (10 Hz), 1.74 (20 Hz), 1.58 (30 Hz), 1.47 (40 Hz), and 1.49 g (50 Hz). Each participant's exercise conditions were assigned in a randomized order.

To ensure a consistent exercise posture, the participants attended practice and warm-up sessions upon arriving at the laboratory. The basic exercise was a dynamic squat with a 2 s eccentric phase (descent) and a 2 s concentric phase (ascent), with the squat rhythm controlled by a 60



Figure 1. Exercise posture on the whole-body vibration platform

bpm metronome (Figure 1). In the warm-up session, the foot position during the squat exercise was marked on the WBV platform. The participants were instructed to maintain a shoulder-width stance to facilitate a smooth squatting movement and to keep their knee angle as parallel to the WBV platform as possible (average knee angle: 79.9 ± 7.9 degrees). They were also asked to stare straight ahead and hold the handle lightly with both hands for balance. The participants performed all exercises barefoot to ensure consistency and avoid potential damping effects from footwear. In each condition, the participants performed 10 squats (total of 40 s), with a 5 min rest interval between conditions.

Data Acquisition

sEMG and acceleration data from the lower extremity muscles were collected as the participants performed squats. Data acquisition was performed with a wireless Noraxon Ultium surface electromyograph (Noraxon USA Inc., Scottsdale, the USA), which includes an integrated inertial measurement unit (IMU). The sEMG signals were sampled at a frequency of 2,000 Hz, and the acceleration data were sampled at 200 Hz.

The sEMG data were collected from the participants' dominant leg, which is the leg they prefer to use for kicking a ball. Data were collected from 11 lower limb muscles: gluteus maximus, rectus femoris, vastus medialis, vastus lateralis, semitendinosus, biceps femoris, tibialis anterior,

peroneus longus, soleus, gastrocnemius medialis, and gastrocnemius lateralis. The attachment area was prepared by removing hair, cleaning it with alcohol, and ensuring that it was dry before electrode implantation to reduce impedance. Ag/AgCl bipolar surface electrodes were then affixed to the prepared sites. The SENIAM (Surface EMG for Non-Invasive Assessment of Muscles) guidelines were adhered to for electrode placement³³.

Data Processing

sEMG and Acceleration Data

Initially, the z-score method was employed to eliminate anomalies in the raw sEMG and acceleration signal. Data points that exceeded the z-score threshold of 3 (i.e., deviating from the mean by more than three times the standard deviation) were classified as outliers and subsequently designated as missing values. These missing values were replaced via linear interpolation. A fourth-order Butterworth filter with a bandpass of 10–500 Hz was employed to process all raw sEMG and acceleration signals and eliminate motion artifacts. Full-wave rectification was performed, and the signals were smoothed with a low-pass filter with a cutoff frequency of 6 Hz. For each subject, the sEMG signals from each muscle were normalized to the maximum value observed across all 12 WBV training conditions^{34,35,36}.

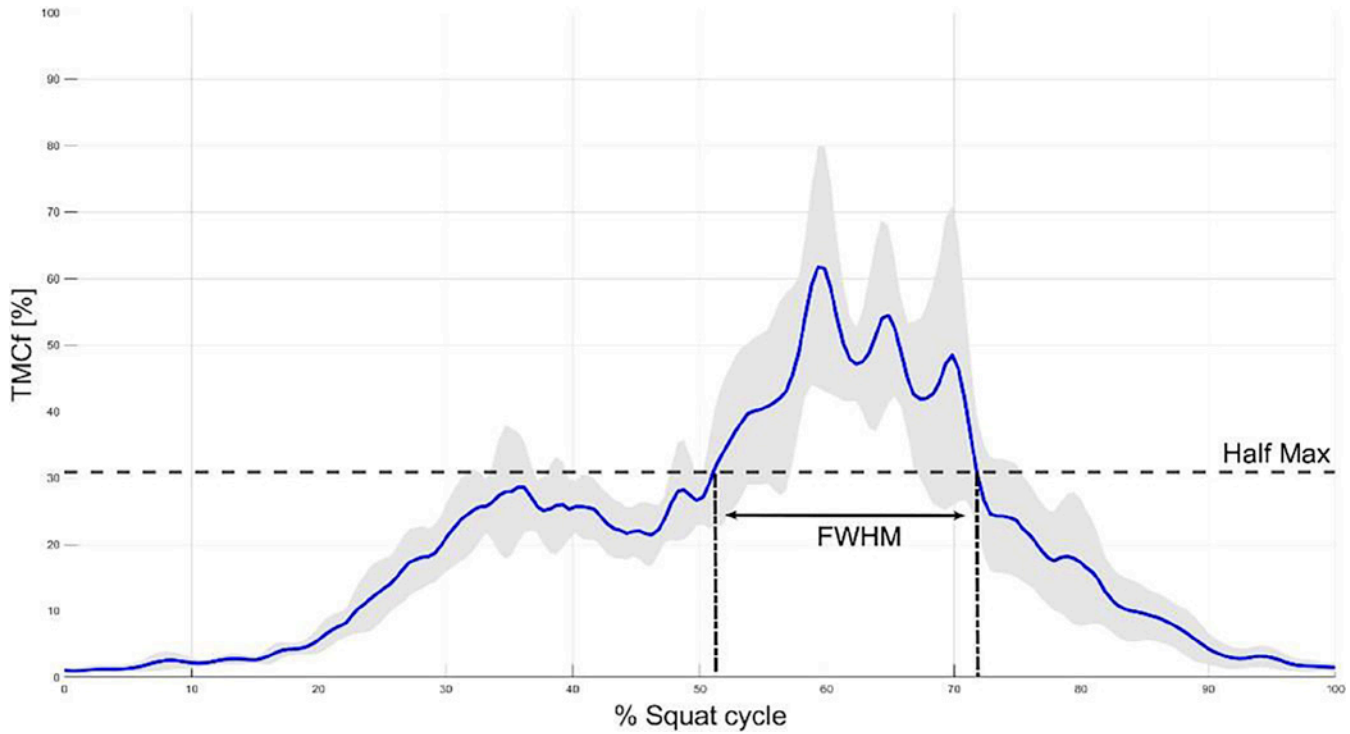


Figure 2. Full width at half maximum (FWHM).

Squat Cycle Detection and Analysis

The acceleration data from IMU were utilized to identify and define the squat cycle. The period during which the Z axis of the IMU acceleration value increased from the minimum to the maximum was identified as the eccentric phase, and the period during which the value decreased from the maximum to the minimum was identified as the concentric phase of the squat.

The sEMG data were time-normalized to the duration of each squat cycle and interpolated to 200 samples to allow a comparison of multiple squat cycles with varying durations³⁷. For each condition, six squat cycles were analyzed, excluding the first two and last two repetitions.

Data Analysis

The EMG root mean square (EMGrms) of 11 muscles was calculated separately to determine muscle activation during WBV training under various conditions. The EMGrms formula is as follows:

$$EMGrms = \sqrt{\frac{1}{N} \sum_{i=1}^N EMG(i)^2}, \quad (1)$$

where $EMG(i)$ represents the original EMG signal of the i -th sample and N denotes the number of samples for calculating RMS.

The coactivation of the lower extremity muscles over

time during the squat was analyzed using the TMCf method^{35,36,37,38}. In TMCf analysis, only the muscles that produce synergistic effects during exercise movements must be included²³. Therefore, TMCf analysis was performed with six muscles, namely, gluteus maximus, rectus femoris, vastus medialis, vastus lateralis, semitendinosus, and biceps femoris, which are known to substantially contribute to squat movement^{39,40}. The TMCf formula is:

$$TMCf(d(k), k) = \left(1 - \frac{1}{1 + e^{-1.2(d(k) - 0.5)}}\right) \cdot \frac{(\sum_{m=1}^M sEMGm(k)/M)^2}{\max_{m=1 \dots M} [sEMGm(k)]}, \quad (2)$$

where $d(k)$ is the average of the differences between the “ k ”-th samples of each sEMG signal pair and calculated as:

$$d(k) = \frac{\sum_{m=1}^{M-1} \sum_{n=m+1}^M |sEMGm(k) - sEMGn(k)|}{J \left(\frac{M!}{2!(M-2)!} \right)}, \quad (3)$$

where J represents the length of the signal and M is the number of muscles considered for analysis. $sEMGm(k)$ and $sEMGn(k)$ represent the k -th sample values of the sEMG signal envelope for the m -th and n -th muscles, respectively.

On the basis of the TMCf results, the full width at half maximum (FWHM) and the center of activity (CoA) were analyzed. FWHM is defined as the total duration of the squat cycles during which the TMCf values exceed half of their maximum. The maximum value of the TMCf curve was initially identified and subsequently halved to calculate

Table 1. Results of EMGrms (Mean ± SD). *GLU* gluteus maximus, *RF* rectus femoris, *VM* vastus medialis, *VL* vastus lateralis, *ST* semitendinosus, *BF* biceps femoris, *TA* tibialis anterior, *PL* peroneus longus, *SOL* soleus, *GM* gastrocnemius medialis, *GL* gastrocnemius lateralis.

Muscle	Loading	0 Hz	10 Hz	20 Hz	30 Hz	40 Hz	50 Hz	Loading effect	Frequency effect	Interaction
GLU	No loading	18.99±5.69	23.02±6.27	22.80±5.49	21.58±6.47	20.05±6.28	20.15±6.14	F(1,24)=16.51, p<0.001***	F(5,120)=13.02, p<0.001***	F(5,120)=2.07, p=0.105
	Additional loading	23.51±7.96	25.95±7.81	25.79±7.65	26.54±9.05	25.07±7.91	24.35±7.37			
RF	No loading	24.05±9.03	25.49±8.11	25.12±7.67	24.99±7.68	24.62±6.85	24.58±7.33	F(1,24)=39.26, p<0.001***	F(5,120)=0.97, p=0.440	F(5,120)=0.62, P=0.689
	Additional loading	31.47±5.23	32.47±5.50	32.38±4.85	31.71±5.05	30.93±5.62	32.06±4.89			
VM	No loading	26.95±6.68	29.60±5.59	29.10±5.71	29.31±6.39	28.93±6.22	29.18±5.84	F(1,24)=59.99, p<0.001***	F(5,120)=2.44, p=0.038*	F(5,120)=2.02, p=0.081
	Additional loading	34.87±5.83	35.66±5.60	34.66±6.01	35.80±5.83	35.05±6.53	34.57±6.65			
VL	No loading	31.71±6.08	31.94±5.80	31.63±5.44	31.31±6.53	31.33±6.41	31.55±5.78	F(1,24)=43.87, p<0.001***	F(5,120)=0.42, p=0.834	F(5,120)=0.58, p=0.716
	Additional loading	37.27±6.24	37.30±6.41	36.58±6.47	37.33±6.15	36.82±6.71	37.15±6.86			
ST	No loading	23.98±8.93	26.65±8.50	26.72±8.76	25.12±9.23	24.22±8.05	25.54±8.45	F(1,24)=19.24, p<0.001***	F(5,120)=3.29, p=0.008**	F(5,120)=0.63, p=0.675
	Additional loading	28.15±8.36	30.48±7.24	31.23±8.04	29.93±8.75	30.42±7.76	29.92±8.17			
BF	No loading	24.71±7.44	26.37±6.85	25.54±6.86	26.46±7.51	26.12±7.64	26.44±7.45	F(1,24)=18.64, p<0.001***	F(5,120)=5.01, p<0.001***	F(5,120)=0.98, p=0.431
	Additional loading	32.55±7.73	34.87±7.08	34.54±7.24	33.61±7.27	33.90±7.35	33.06±7.72			
TA	No loading	23.33±10.91	24.69±11.54	24.56±9.90	24.02±10.05	23.51±9.78	24.68±10.25	F(1,24)=0.22, p=0.643	F(5,120)=0.65, p=0.664	F(5,120)=0.66, p=0.657
	Additional loading	24.22±9.01	23.52±9.08	24.25±10.04	25.50±9.87	24.74±9.74	26.60±9.23			
PL	No loading	27.58±8.07	29.20±10.66	31.10±9.49	29.71±9.19	30.41±9.19	29.99±9.77	F(1,24)=0.83, p=0.373	F(5,120)=0.82, p=0.537	F(5,120)=0.64, p=0.603
	Additional loading	31.01±7.98	30.59±8.23	31.02±8.86	31.84±8.04	32.16±7.66	31.75±9.85			
SOL	No loading	23.03±7.31	26.92±7.65	26.34±8.02	24.48±7.16	26.59±5.85	25.07±7.00	F(1,24)=28.19, p<0.001***	F(5,120)=3.13, p=0.011*	F(5,120)=1.61, p=0.163
	Additional loading	28.82±7.15	29.22±6.70	32.37±7.24	30.29±7.84	31.00±6.74	29.91±7.14			
GM	No loading	18.23±9.42	20.61±8.68	20.39±8.98	18.02±9.44	17.85±8.50	16.87±9.85	F(1,24)=1.36, p=0.255	F(5,120)=2.52, p=0.033*	F(5,120)=0.67, p=0.597
	Additional loading	19.63±10.03	20.18±9.26	21.22±9.21	19.59±9.43	19.84±8.46	18.84±8.98			
GL	No loading	22.31±8.84	25.07±8.64	25.90±9.49	24.42±10.42	23.97±7.67	22.96±9.17	F(1,24)=2.71, p=0.113	F(5,120)=2.93, p=0.016*	F(5,120)=0.58, p=0.643
	Additional loading	25.43±9.46	26.34±8.99	27.11±10.72	26.31±10.21	26.04±9.34	24.97±10.08			

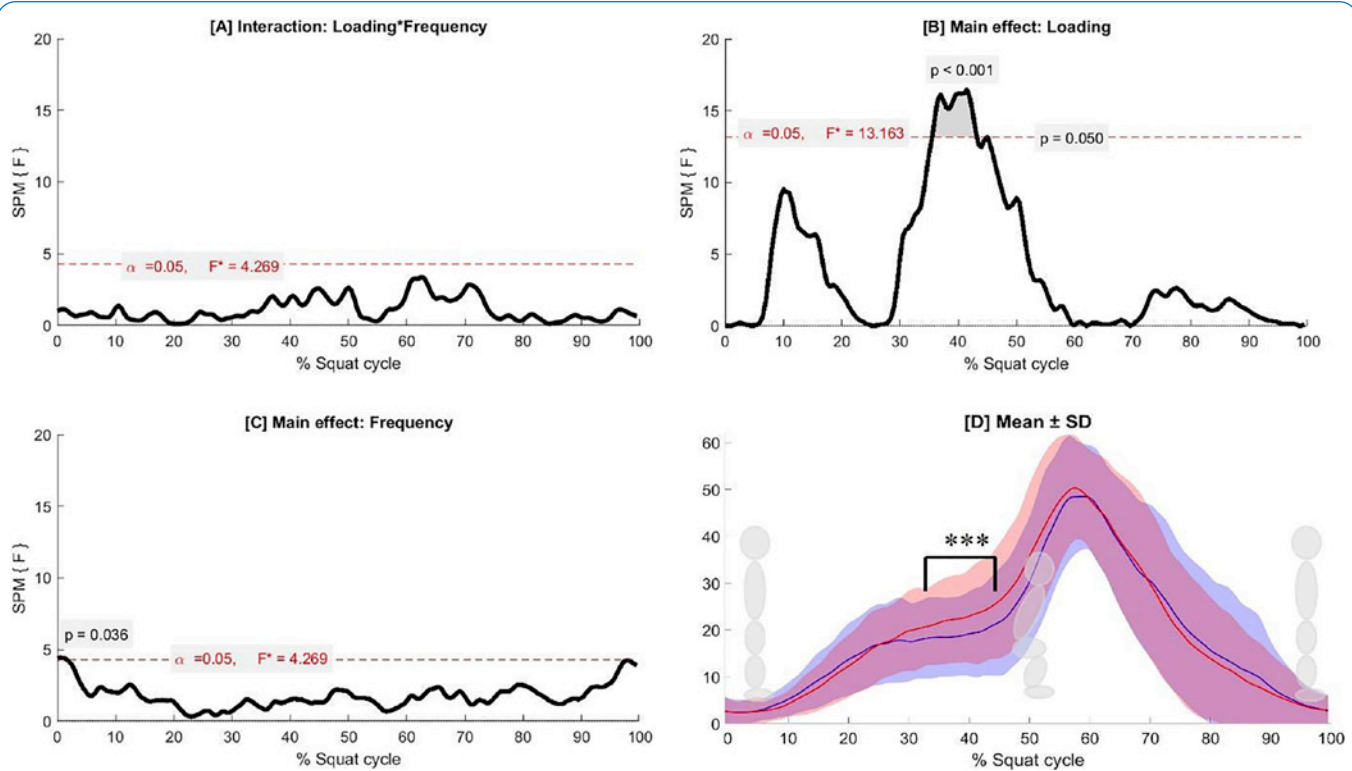


Figure 3. SPM analysis results for TMCf. [A] Interaction effect of loading and frequency, [B] main effect of loading, [C] main effect of frequency, and [D] mean±SD under no-loading and additional-loading conditions. The red line represents the mean of the additional-loading condition, and the blue line is the mean of the no-loading condition. The red and blue shaded areas indicate the SD for additional-loading and no-loading conditions, respectively.

FWHM. The time points corresponding to this half-maximum value were then determined (Figure 2).

During the squat, CoA was calculated using the following formula to determine the point in the squat cycle where coactivation was most concentrated:

$$\begin{cases} A = \sum_{i=1}^{201} (\cos \theta_t \times EMG_i) \\ B = \sum_{i=1}^{201} (\sin \theta_t \times EMG_i) \end{cases}, (4)$$

$$CoA = \tan^{-1} \left(\frac{B}{A} \right)$$

Statistical Analyses

Statistical analyses were performed using SPSS software (version 27.0, SPSS, Inc., Chicago, IL, the USA) and MATLAB (R2023b, MathWorks, the USA). Two-way repeated-measures ANOVA (2 [loading] × 6 [frequency]) was conducted to evaluate the effects of vibration frequency and loading on EMGrms. Post-hoc comparisons were performed using Bonferroni tests. To analyze the continuous data during the squat cycle, a two-way repeated measures ANOVA (2 [loading] × 6 [frequency]) was performed using one-dimensional statistical parametric mapping (1D-SPM), which is available as open-source software in MATLAB.

Post-hoc analyses were conducted using paired t-tests when significant differences were found. Statistical significance was set at 0.05 for all analyses.

Results

EMGrms

The EMGrms results are shown in Table 1. No significant interaction effects were observed between loading and frequency. The gluteus maximus was significantly affected by loading ($F[1,24] = 16.51, p < 0.001$) and frequency ($F[5,120] = 13.02, p < 0.001$), with high EMGrms under additional loading. The EMGrms values at 10, 20, and 30 Hz were significantly higher than that at 0 Hz. With regard to the quadriceps (rectus femoris, vastus medialis, and vastus lateralis), only loading had a significant effect (rectus femoris: $F[1,24] = 39.26, p < 0.001$; vastus medialis: $F[1,24] = 59.99, p < 0.001$; vastus lateralis: $F[1,24] = 43.87, p < 0.001$), with high EMGrms in the additional-loading condition. The hamstrings (semitendinosus and biceps femoris) were significantly influenced by loading (semitendinosus: $F[1,24] = 19.24, p < 0.001$; biceps femoris:

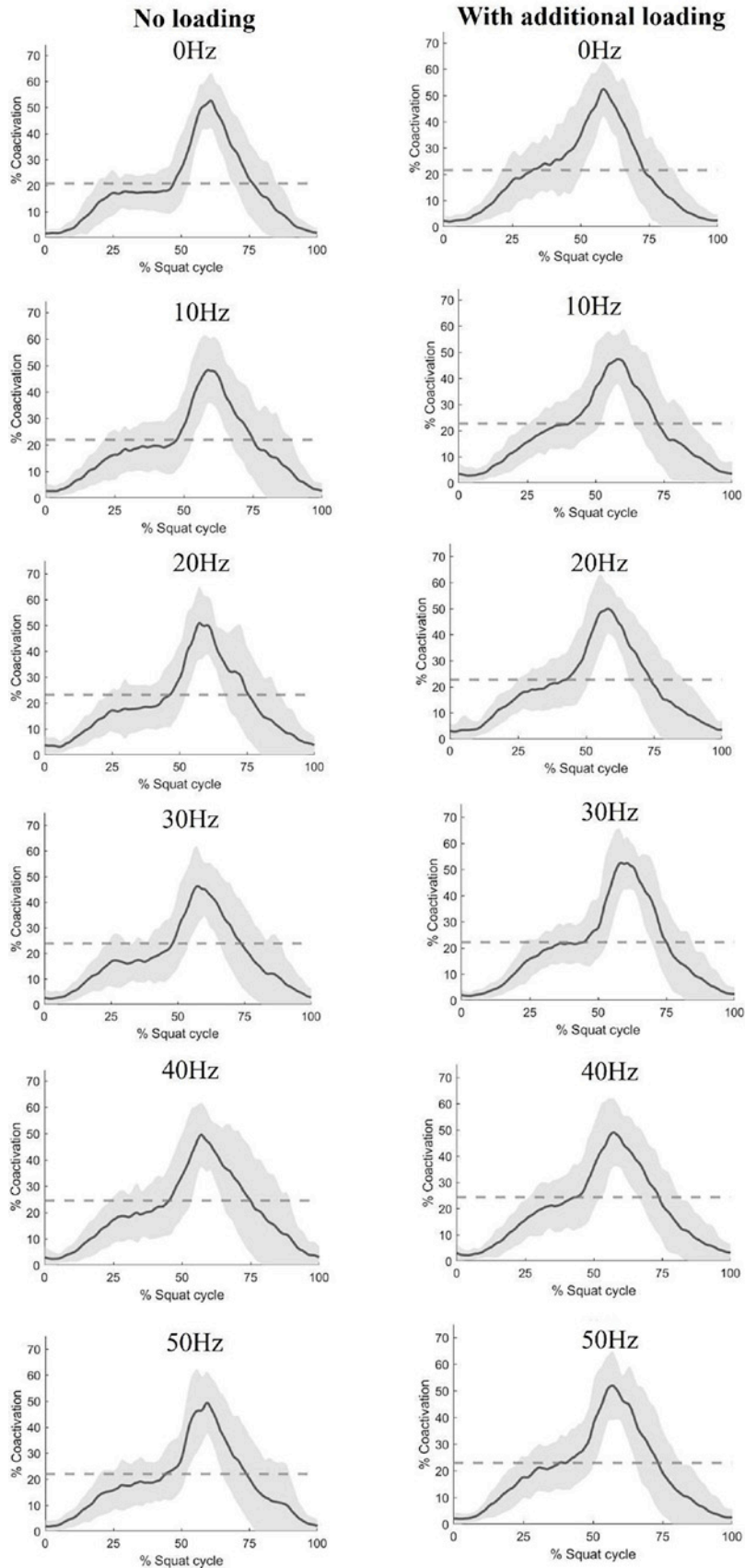


Figure 4. Results of TMCf across all conditions. The solid line represents the mean, and the gray shaded area represents SD. The dotted line indicates half maximum.

Table 2. Mean and SD of FWHM results across different loading and frequency conditions.

Frequency	No loading		With additional loading	
	Mean	SD	Mean	SD
0 Hz	20.92	7.00	21.47	8.33
10 Hz	22.11	10.15	22.81	8.45
20 Hz	23.10	10.42	22.63	8.84
30 Hz	23.84	10.01	22.17	8.46
40 Hz	24.66	10.74	24.28	8.90
50 Hz	22.09	11.35	22.97	8.89

Table 3. Mean and SD of CoA results across different loading and frequency conditions.

Frequency	No loading		With additional loading	
	Mean	SD	Mean	SD
0 Hz	57.08	5.70	54.33	5.71
10 Hz	56.71	6.85	55.09	6.51
20 Hz	56.64	6.15	55.33	6.27
30 Hz	56.81	7.77	56.46	5.25
40 Hz	56.24	8.25	55.74	6.76
50 Hz	55.64	6.56	54.13	6.53

$F[1,24] = 18.64, p < 0.001$) and frequency (semitendinosus: $F[5,120] = 3.29, p = 0.008$; biceps femoris: $F[5,120] = 5.01, p < 0.001$), with high EMGrms in the additional-loading condition. The semitendinosus showed significantly higher EMGrms at 10 Hz than at 0 Hz, and the biceps femoris exhibited significantly higher EMGrms at 10 and 20 Hz than at 0 Hz. The soleus demonstrated a significant loading effect ($F[1,24] = 28.19, p < 0.001$), with high EMGrms in the additional-loading condition. The gastrocnemius lateralis showed a significant frequency effect ($F[5,120] = 2.93, p = 0.016$), with EMGrms at 20 Hz being significantly higher than that at 50 Hz. No significant effects of loading and frequency were observed on the gastrocnemius medialis, tibialis anterior, and peroneus longus.

TMCf

SPM analysis demonstrated that the effect of the interaction between loading and frequency on TMCf was not statistically significant ($F[5,120] = 4.269, p > 0.05$; Figure 3A). However, loading ($F[1,24] = 13.163, p < 0.001$; Figure 3B) and frequency ($F[5,120] = 4.269, p = 0.036$; Figure 3C) exerted significant main effects on TMCf. Given these significant main effects, post-hoc tests were conducted to further investigate the differences. The results indicated a significant difference between no-loading and additional-

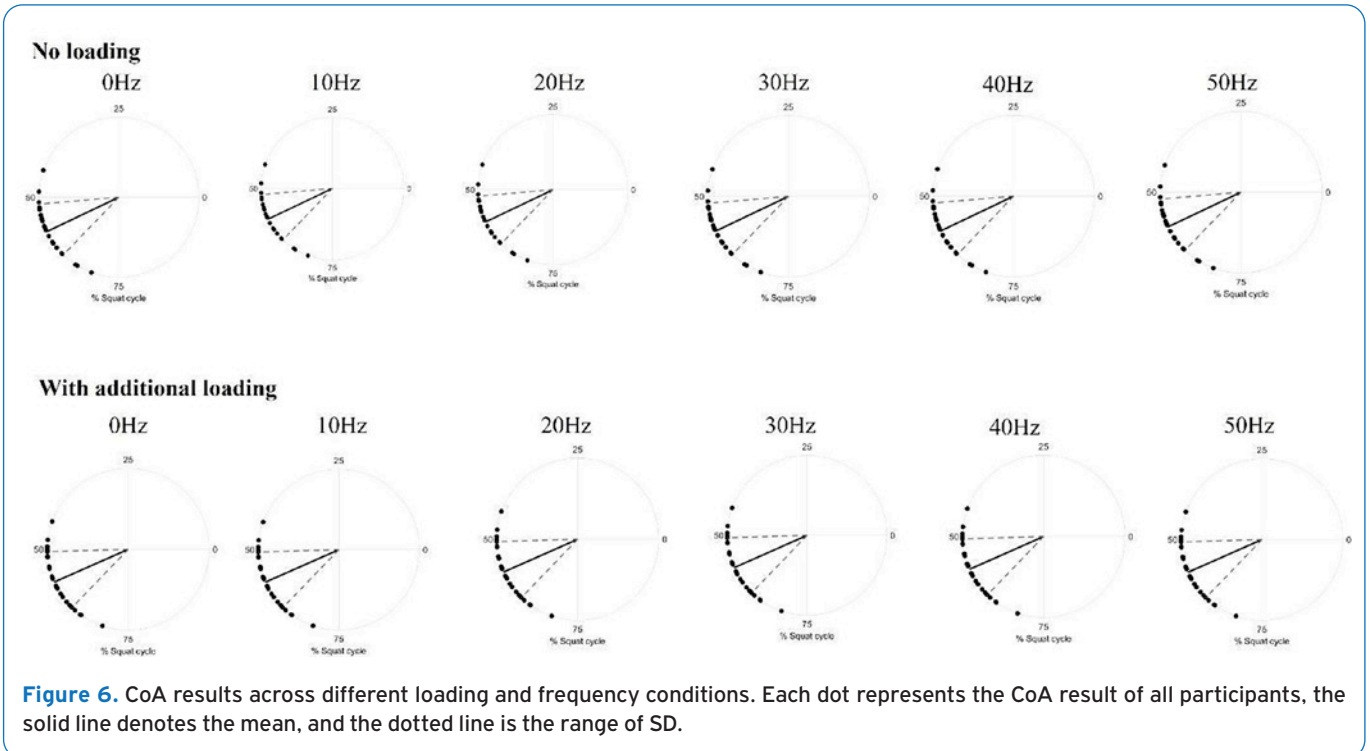
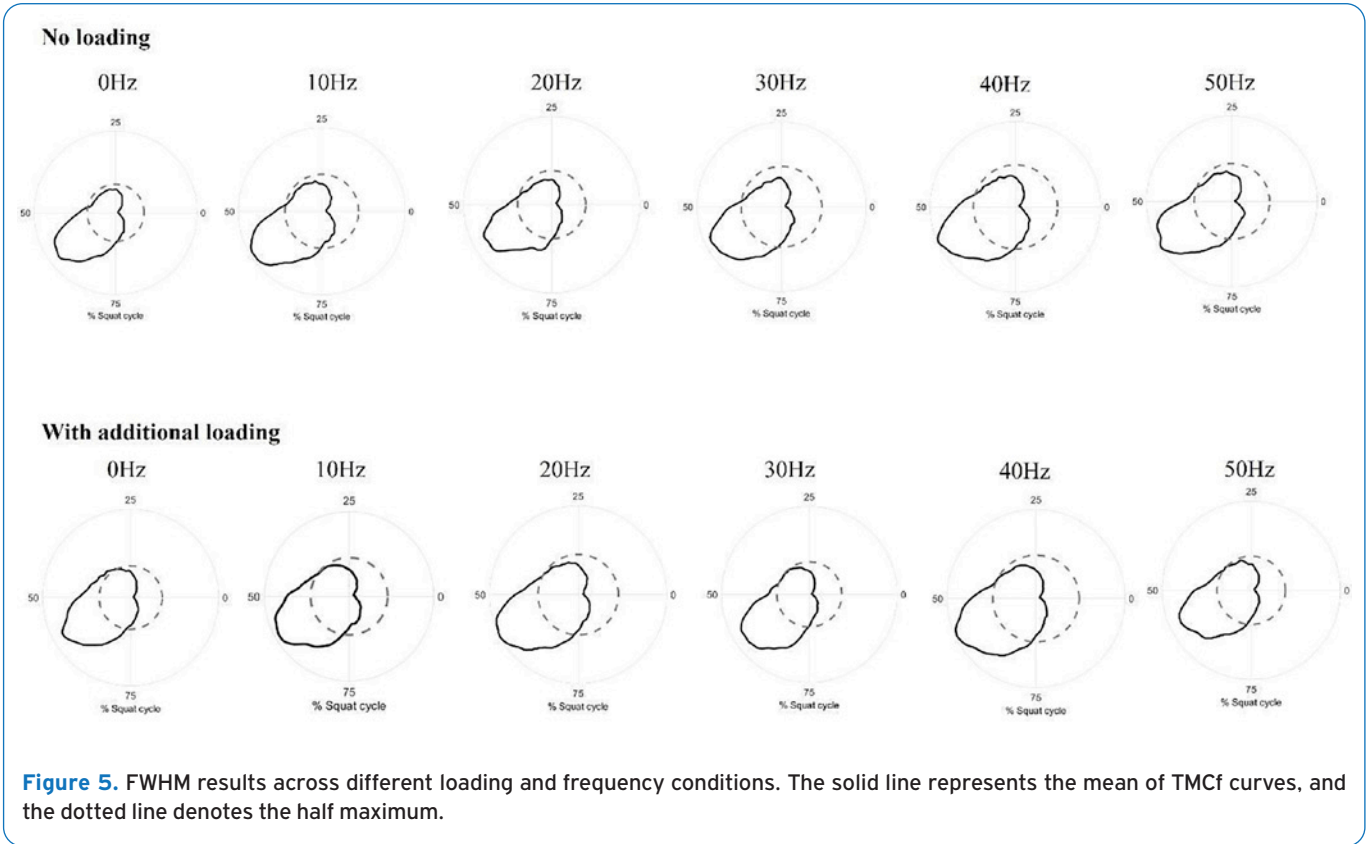
loading conditions during the squat cycle. Specifically, significant differences were found during the 33.5%–43% phase of the squat cycle. In this phase, the additional-loading condition exhibited significantly higher TMCf values compared with the no-loading condition, as shown in Figure 3D. Although the main effect of frequency was significant in the two-way repeated-measures ANOVA, the post-hoc tests did not reveal any significant differences between frequencies. The mean and standard deviation (SD) of TMCf across all conditions are shown in Figure 4.

The FWHM results demonstrated that loading and frequency had no statistically significant interaction effect ($F = 0.507, p = 0.692$). Furthermore, no significant main effects were observed for loading ($F = 0.004, p = 0.951$) and frequency ($F = 2.037, p = 0.078$), as presented in Table 2 and Figure 5.

The CoA results revealed no statistically significant interaction effect between loading and frequency ($F = 1.085, p = 0.372$). The main effects for loading ($F = 3.450, p = 0.076$) and frequency ($F = 1.738, p = 0.131$) were also not significant (Table 3 and Figure 6).

Discussion

This study investigated the effects of additional loading and vibration frequency during WBV training on lower limb



muscle activation and coactivation by using TMCf analysis. EMGrms exhibited statistically significant differences in response to loading conditions in the gluteus maximus, hamstrings, quadriceps, and soleus, and high muscle activation was observed under the additional-loading condition. Furthermore, the gluteal and hamstring muscles exhibited a frequency-dependent response, with 10–30 Hz inducing higher muscle activation compared with the no-vibration condition. Moreover, TMCf was significantly affected by load condition but not by vibration frequency. Specifically, the additional-loading condition elicited higher coactivation than the no-loading condition.

To the best of our knowledge, this study is the first to investigate the effects of lower extremity muscle coactivation during WBV training. Chaltrou et al.⁴¹ examined the coactivation of the hamstrings and quadriceps; however, their study focused only on coactivation during maximum voluntary isometric contraction before and after WBV training. No studies have analyzed coactivation during WBV training. Our study extends extant research by directly examining lower extremity muscle coactivation during WBV training by using TMCf. This approach provides a comprehensive understanding of how WBV parameters influence neuromuscular coordination in real time and offers valuable insights for optimizing WBV training protocols for strength training and injury prevention.

In the present study, we hypothesized that high vibration frequencies increase muscle activation and coactivation. Contrary to our hypothesis, peak muscle activation was observed within the moderate frequency range (10–30 Hz), rather than at high frequencies. This unexpected pattern may be partly attributable to the specific acceleration characteristics of the WBV platform used in our study. The effects of WBV training are largely determined by biomechanical parameters, such as frequency, amplitude, and acceleration^{42,43}. In our setup, the acceleration at moderate frequencies (e.g., 10 Hz: 1.66 g; 20 Hz: 1.74 g; 30 Hz: 1.58 g) was higher than that at high frequencies (e.g., 40 Hz: 1.47 g; 50 Hz: 1.49 g). This mechanical property may have contributed to the enhanced muscle activation at moderate frequencies and may explain the discrepancy with previous findings, such as those of Hazell et al.²⁵, who used a different WBV platform and reported increased muscle activation at 45 Hz.

Despite the substantial frequency-dependent modulation of muscle activation, coactivation did not exhibit any statistically significant variation across the vibration conditions. The present findings extend prior research by demonstrating that while vibration frequency influences the recruitment of individual muscles, it does not substantially affect the underlying neuromuscular coordination strategies responsible for muscle coactivation. A possible explanation lies in the differing sensitivity thresholds of these mechanisms; muscle activation may respond to moderate mechanical stimuli, whereas coactivation (primarily a stabilization strategy) may require intense stimuli to be effectively engaged. Supporting this

interpretation, previous studies that used powerful WBV platforms reported substantially high acceleration levels. For example, Pel et al.⁴⁴ observed accelerations up to 7.7 g at 50 Hz, and Lienhard et al.⁴⁵ found that accelerations ≥ 1.8 g considerably influence lower limb muscle activation. By contrast, the maximum acceleration in our study was only 1.74 g at 20 Hz, which may have been insufficient to elicit measurable changes in coactivation.

Consistent with our hypothesis, muscle activation and coactivation of the lower limb muscles were influenced by the loading condition. Specifically, additional loading led to increased EMGrms values, reflecting heightened activation of individual muscles and substantially elevated TMCf values, which are indicative of enhanced agonist–antagonist coactivation. Similar to our research, previous studies have reported that the combination of additional loading and WBV training increases muscle activation, and extra loading is a variation that can further maximize the exercise effect of WBV training^{18,44}. Hazell et al.²⁵ demonstrated that incorporating an additional load equivalent to 30% of body mass during WBV training results in increased EMGrms of lower extremity muscles, including the vastus lateralis (0%–3.6%), biceps femoris (0.2%–2.0%), tibialis anterior (2.6%–3.8%), and gastrocnemius (6.3%–9.8%). Similarly, Ritzmann et al.¹⁷ compared WBV training with and without an additional loading of 30% body weight and found a substantial increase in muscle activity (integrated EMG) in the rectus femoris (21%), vastus medialis (15%), tibialis anterior (13%), gastrocnemius (14%), and soleus (11%). The primary explanation for the increased muscle activation that results from the additional stimulus is the increased recruitment and synchronization of motor units^{47,48,49}. The weighted vest may enhance this process by providing external resistance.

The explanation for the increased coactivation of lower limb muscles under the additional-loading condition can be attributed to the increased exercise difficulty. Wearing a weighted vest increases the need for trunk stability, thereby increasing the overall difficulty of squat exercise. According to Latash⁵⁰, healthy individuals have high degrees of muscular coactivation when they exercise in unstable or challenging conditions. Challenging task demands necessitate additional corrective movements because of the inherent difficulty in anticipating changes in external or body conditions. This situation enables muscles to coactivate effectively in adapting to unexpected changes. In addition, unstable motions can limit the capacity to accurately determine the location of joints. However, simultaneous activation of muscles, known as muscle coactivation, aids in maintaining joint stability and enhancing control during movement⁵⁰. Therefore, the addition of loading through a weighted vest creates challenging exercise conditions, resulting in increased coactivation of lower extremity muscles. This increased coactivation helps maintain joint stability and allows for uninterrupted corrective motions during challenging activities.

SPM analysis showed that the additional-loading condition

had higher values of coactivation than the no-additional-loading condition, particularly between 33.5% and 43% of the squat phase (i.e., eccentric phase). This phase relates to the moment when the leg is descending right before it reaches the same level as the floor. During this phase, the center of gravity of the body shifts, and the angles of the hip, knee, and ankle increase simultaneously, resulting in increased instability⁵¹. To counteract this instability, muscles coactivate to ensure joint safety. The increase in coactivation during this period may be a response to dissipate stress on the joint and prevent injury.

TMCf showed statistically significant differences depending on the WBV training parameters, but no significant differences were observed in FWHM and CoA. FWHM quantifies the duration of time that a muscle is active during an exercise cycle. The lack of significant differences in FWHM among the various WBV frequencies and loading conditions suggests that the duration of muscle activation remains generally stable regardless of the external stimulus. Possibly, the neuromuscular system maintains consistent timing of muscle activation to ensure stability and efficiency during dynamic exercises, such as squats. In other words, the WBV parameter can affect the intensity of muscle coactivation, but not the length of time that a muscle remains coactive during an exercise cycle. CoA indicates where coactivation is most concentrated within an exercise cycle. The CoA results showed no significant difference, suggesting that the peak of muscle coactivation remains constant even under various WBV parameter conditions. This finding may mean that the WBV training parameter primarily modulates coactivation intensity rather than changing the timing of muscle coactivation within an exercise cycle.

This study systematically analyzed the effects of additional loading and vibration frequency during WBV training on lower limb muscle activation and coactivation, providing critical insights for optimizing WBV applications. The substantial increase in muscle coactivation under additional loading suggests that combining WBV with external loading may be an effective strategy for enhancing joint stability and neuromuscular control. Furthermore, in this study, vibration frequency (10–30 Hz) played a crucial role in increasing muscle activation, particularly in the gluteal and hamstring muscles. Therefore, when designing WBV-based strength and balance training programs for sports performance enhancement and rehabilitation, precisely adjusting vibration frequency and external loading is essential to maximize neuromuscular activation and achieve optimal training effects.

This study has several limitations. First, this study included only healthy young male recreational runners. Therefore, the findings cannot be generalized to females, old individuals, and those participating in different sports activities. Future studies must address these limitations and investigate the effects of WBV training on different populations. Second, this study focused on analyzing the effects of vibration frequency and additional loading

conditions on lower extremity muscle coactivation. However, in addition to vibration frequency and additional load, other factors, such as exercise posture, vibration direction of the vibration platform, and vibration exposure duration, may change the effect of WBV training. These variables were not considered in this study. Therefore, future research should investigate the effects of various WBV parameters on muscle coactivation.

Conclusion

The findings of this study suggest that vibration frequency and additional loading are key components of WBV training protocols. Lower limb muscle activation was substantially influenced by vibration frequency, with the hip and hamstring muscles showing the highest activation between 10 and 30 Hz. Muscle coactivation was considerably affected by additional loading. These findings highlight the importance of additional loading as a stimulus for enhancing lower extremity muscle coactivation, which may contribute to improved joint stability and movement accuracy.

Ethics approval

The study was approved by the Human Ethics Committee of Shanghai University of Sport (102772024RT052).

Consent to participate

All participants provided written informed consent prior to participation.

Authors' contributions

Conceptualization: Hyeri Nam. Formal Analysis: Hyeri Nam, Lulu Yin, Peng Chen.

Investigation: Hyeri Nam, Lulu Yin, Peng Chen. Methodology: Hyeri Nam, Lin Wang.

Supervision: Lin Wang. Writing – original draft preparation: Hyeri Nam. Writing – review & editing: Lin Wang, Lulu Yin, Peng Chen.

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